To develop a clinically friendly multi-segment kinematic and kinetic foot model and test it on pre-/post-HTO patients with medial knee osteoarthritis and knee varus

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A thesis submitted in partial fulfillment of the requirements for the Master of Science degree in Kinesiology
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Abstract

This study aims to develop a clinically useful multi-segment foot model that will enable the analysis of foot kinematics and kinetics in an optical motion capture laboratory setting. This study will also test the new multi-segment foot model on patients with knee osteoarthritis and varus knee alignment that is corrected by high tibial osteotomy (HTO). The multi-segment foot model divides the foot into four functional segments: the hindfoot, midfoot, forefoot and hallux. An X-Z-Y-Cardan angle rotation convention was used to determine intersegmental dorsi/plantar flexion, inversion/eversion, and internal/external rotation. Joint moments, joint powers and medio-longitudinal arch (MLA) height/length ratio were also measured. Ten healthy adults and four HTO patients were tested with an optical motion capture system as they walked barefoot in their self-selected speeds. Repeatability of joint motions was calculated using coefficients of multiple correlation. Outcome measures were compared with other multi-segment foot models found in the biomechanics literature to assess validity.

This novel multi-segment foot model showed strong test-retest and within-subject reliability (R>0.7) for most joint motions (24/27) in healthy adults. The model was sensitive enough to detect abnormal foot motions, including lower MLA, increased hallux abduction, in patient gait compared to the normal cohort. This novel model has been shown to be a clinically useful tool for research and assessment on clinical populations.

Keywords

Optical motion capture, multi-segment foot model, gait analysis, intersegmental joint motions, foot kinematics, foot kinetics, osteoarthritis, high tibial osteotomy
Summary for Lay Audience

The foot has been traditionally treated as a single, rigid segment in gait analysis. For this reason, the loads and motions of the segment that make up the foot have not been studied. The study aims to develop a multi-segment foot model that can measure the motion and biomechanical load within the foot in patients in a clinical setting. This model functionally divides the foot into four segments: the hindfoot, midfoot, forefoot and hallux. Relative motions between these segments and the load carried between these segments is measured. The novel model was tested on patients with knee osteoarthritis and a varus knee alignment. These patients were undergoing a corrective realignment surgery at the knee called a high tibial osteotomy (HTO). It is hypothesized that due to the varus knee alignment, these patients would experience abnormal foot/ankle motions as well. This novel multi-segment foot model showed strong reliability and validity in a cohort of ten healthy adults and was sensitive enough to detect abnormal foot/ankle motions in four HTO patients. This model has been shown to be a clinically useful tool in a gait analysis laboratory for use in research and for the assessment on clinical populations.
Co-Authorship Statement

I acknowledge my supervisor Thomas Jenkyn who helped me design the study and edit the writing of this thesis.
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I would like to thank my supervisor and mentor Thomas Jenkyn for all his help and support over the past two years. He supported me with abundant resources and connections I needed for completing the thesis. His patient guidance inspired me a lot during difficult times. I learned so much from him in many ways and could not be more grateful to be his student.

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I would also like to thank Kristyn Leitch and other members of WOBL for teaching me how to use different equipment, collecting data from patients, and answering my questions. It is my pleasure to meet and work with these friendly, knowledgeable, and amazing people.

The past two years have been the most challenging time ever for people all over the world. It has also been significant for me personally as I have profoundly experienced my inner struggles and fears more so than ever. No growth comes without pain. I would like to thank my parents, who are one Pacific Ocean away, for being my best friends and giving me unconditional love and understanding along the way.

Finally, I would like to thank all frontline healthcare workers and researchers in this global battle with COVID-19. They are real heroes. It is their sacrifices that let us live a life. May all be healthy.
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Glossary of Acronyms

LLA: lateral longitudinal arch
MLA: medial longitudinal arch
MP: metatarsophalangeal
DOF: degree of freedom
ROM: range of motion
HS: heel strike
FF: foot flat
MS: midstance
HF: heel off
TO: toe off
BSIP: body segment inertial parameter
CM: center of mass
RG: radius of gyration
GRF: ground reaction force
OA: osteoarthritis
HKA: hip-knee-ankle
HTO: high tibial osteotomy
LCWO: lateral closing wedge osteotomy
MOWO: medial opening wedge osteotomy
LM: lateral malleolus
MM: medial malleolus
TC: top calcaneus
BC: bottom calcaneus
TH: talar head
NV: navicular
CB: cuboid
MidNC: midpoint between navicular and cuboid
1MTH: first metatarsal head
5MTH: fifth metatarsal head
1-5MTH: midpoint between 1MTH and 5MTH
23T: between the heads of second and third metatarsals
CMC: coefficient of multiple correlation
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1.1 Anatomical planes and movements

When describing the human body and motions that occur within the body, three anatomical planes and three anatomical axes are used (Figure 1-1). These planes, known as the cardinal planes, are the transverse plane, the frontal plane and the sagittal plane. The transverse plane separates the body top from bottom. The axis normal to the transverse plane is the long axis. Rotations around the long axis generate motions in the transverse plane to the left or the right. The frontal plane divides front from back and the axis normal to this plane is the anterior-posterior axis. Rotations around the anterior-posterior axis generate motions of abduction and adduction in the frontal plane. The sagittal plane divides the body into left and right sides and the axis normal to this plane is the mediolateral axis. Rotations about this axis are flexion and extension. Flexion and extension motion describe movements that decrease and increase joint angles, respectively. Abduction and adduction describe motions in the frontal plane that move away and towards the midline of the body, respectively. In the limbs, internal and external rotation describe movements in the transverse plane that rotate along the long axis of the body segment toward or away from the anterior surface of the body (Oatis, 2009).

The rotation axes of joints of the foot and ankle tend to not be perpendicular to the cardinal planes. So motions at these joints result in rotations within three planes. As a result, foot supination involves inversion, internal rotation, and plantar flexion and foot pronation involves eversion, external rotation and dorsiflexion (Brockett et al., 2016). To keep the terminology consistent in describing foot and ankle motions in this study, the terms used are specified to each joint and segment to avoid confusion.
Figure 1-1: Anatomical planes (left) and movements (right). Left adapted from “Goniometer crosstalk compensation for knee joint applications.” by De Oliveira Sato, T., Hansson, G.-Å., & Coury, H. J. C. G., 2010, Sensors (Basel, Switzerland), 10(11), 9994–10005. CC BY 3.0. Right adapted from Kinesiology: the mechanics and pathomechanics of human movement, Figure 7.1, by Oatis, C. A., 2009, Lippincott Williams & Wilkins. Copyright 2009 by © Wolters Kluwer Health, Inc.

1.2 The foot and ankle: anatomy and movements

The foot is made up of twenty-eight bones and is typically subdivided into three segments: the hindfoot, the midfoot, and the forefoot as shown in Figure 1-2. The hindfoot consists of the calcaneus and talus. The calcaneus forms the heel. The talus sits superior to the calcaneus and articulates with the distal end of the lower leg. The ankle (talocrural joint), formed by the talus, the tibia, and the fibula, enables dorsiflexion and plantar flexion of the foot with respect to the lower leg in the sagittal plane. The subtalar joint is the articulation between the talus and the calcaneus. Inversion and eversion of the foot relative to the lower leg occurs at the subtalar joint. The midfoot consists of the navicular, cuboid and three cuneiforms. These bones have a small range of motion relative to each other. The midfoot articulates with the hindfoot at the transverse tarsal joints including the talonavicular joint and the calcaneocuboid joints (the Chopart joint complex). The forefoot consists of five metatarsal bones. Joints in the forefoot are
interphalangeal joints and tarsometatarsal joints (the Lisfranc joint). Only the first and fifth metatarsal bones have a range of motion (ROM) to adapt to uneven surfaces (Chan & Rudins, 1994). The first metatarsophalangeal (MP) joint is the articulation between the hallux and the first metatarsal bone. The shape of foot bones combined with the support of ligaments forms the lateral longitudinal arch (LLA), the medial longitudinal arch (MLA), and the transverse arch as shown in Figure 1-3. The LLA consists of the calcaneus, cuboid and the lateral two metatarsal bones. The MLA consists of the calcaneus, talus, navicular, three cuneiforms and the medial three metatarsal bones and is usually not in contact with the ground in static loading due to its greater height than LLA. Plantar aponeurosis and the calcaneonavicular ligament (spring ligament) are significant to the maintenance of MLA. The transverse arch is across the tarsometatarsal joints. The arches support and protect the foot by changing its flexibility and rigidity to redistribute pressure during dynamic loading (Drake et al., 2005). Movements of the foot and ankle can be described as dorsi-/plantar flexion in the sagittal plane, inversion/eversion in the frontal plane, and internal/external rotation in the transverse plane as shown in Figure 1-4.

Figure 1-2: Bony structures of the foot. Adapted from “Finite element modeling and analysis of human foot,” by Vigneshwar, R., Jeevamalar, J. & Vijayaragavan, E., 2015, International Conference on Advances in Materials and Mechanical Engineering (ICAMME 2015),
Figure 1-3: Arches of the foot. Red: LLA. Green: MLA. Blue: transverse arch.
1.3 Optical Motion Capture

Optical motion tracking systems allow for the three-dimensional (3D) motion of the segments of the body to be captured. Auto-reflective markers are attached to each segment. Their reflections are captured by multiple cameras and the 3D position of each marker is then triangulated by motion analysis software. Multiple cameras are positioned to film the moving objects at various angles. Passive (retroreflective) or active (blinking) markers are attached to specific landmarks on the surface of the body. A marker must be visible to at least two cameras to be located in the global reference frame of the laboratory. At least three non-collinear markers are required on each body segment in order to measure its 3D pose. Pose is its position in the global reference frame and its angular orientation. The three segment markers are used to construct a segment-fixed reference frame, which enables intersegmental kinematic measures including joint angles,
velocities, and accelerations in 3D space. Virtual markers can be defined within the segment-fixed reference frame at points of anatomical interest, (i.e., at the center of joints) and tracked optically along with the actual markers. Virtual markers can also be used to construct anatomical coordinate systems for a body segment.

Retroreflective markers were used in the current study. Most passive systems use either visible red lights or infrared to illuminate the markers for the cameras. Each marker’s 3D coordinates were measured, and the segment pose was determined in six degrees of freedom (DOFs) as the test subject moved through the lab space. Data accuracy depended on the quality of cameras and markers, calibration of cameras each day, and marker placement. In addition, since markers are attached to the skin, errors in segment pose determination arose from the movement of the soft tissues to which the markers were attached.

1.3.1 The Walking Gait Cycle

Walking has a cyclic movement pattern as two feet alternate to move the body forward. A gait cycle starts with a foot-strike on the ground and ends with the next foot-strike on the same side. The cycle of one leg and foot can be divided into the stance phase and the swing phase. The stance phase is when the foot is in contact with the ground. The swing phase follows as the foot rises in the air, until it is in contact with the ground again and another gait cycle begins. Since different gait events happen in sequence regardless of the walking speed, they are described in terms of the time percentage of a gait cycle for comparison between individuals and between trials (Figure 1-5). Five gait events can be identified in the stance phase: heel strike (HS), foot flat (FF), midstance (MS), heel off (HF), and toe off (TO). HS is the initial contact with the ground. FF, happening at about 15% into the cycle, is when the sole of the foot comes into full contact with the ground and supports the full body weight. MS is defined when the hip joint is right above the ankle joint center (between 20-30%). HF is when the heel lifts off the ground and is ready to propel forward. TO is when the toe lifts off the ground, representing the termination of the stance phase. The period between HF and TO is identified as pre-swing (between 50-60%). The swing phase is divided into swing-through and deceleration of the leg. The initial HS is represented by 0% while the subsequent HS of the same foot by 100%. The
stance phase makes up for about 60% of a gait cycle while the swing phase accounts for the remaining 40% (Chan & Rudins, 1994).

Figure 1-5: A gait cycle. Adapted from “Towards effective non-invasive brain-computer interfaces dedicated to gait rehabilitation systems,” by Castermans, T., Duvinage, M., Cheron, G., & Dutoit, T., 2013, Brain Sciences, 4(1), 1–48. CC BY 3.0.

The foot and ankle generate motions in all three planes during a gait cycle. For a normal gait cycle of a healthy adult, the foot in the first portion of stance phase is in a flexible configuration due to it evertng and internally rotating during the first 15% of the stance phase (from HS to FF). In the configuration, the MLA is flexible and drops to adapt to the ground. Plantar flexion occurs at HS and continues until MS. Starting from MS until TO, the foot inverts, externally rotates and dorsiflexes to restore a rigid, stable MLA and prepare for forward propulsion. The foot dorsiflexes from MS to approximately 40% of the cycle and then becomes slightly plantar flexed in pre-swing. During the swing phase, the foot is less rigid compared to the stance phase and thus appears slightly externally rotated and everted. Dorsiflexion is sustained in the swing phase to prevent the foot from tripping over the ground until the next HS (Chan & Rudins, 1994).

1.4 Multi-segment foot models

The foot has been treated as a single rigid segment for most of the history of human gait analysis (Burnett et al., 2015; Kabada et al., 1990). While a single-segment foot model is simple and easy to implement in motion capture, it has several problems in clinical gait analysis, since clinically important motion of the foot cannot be measured. Biomechanical load in the hindfoot, midfoot and forefoot, differences between foot types
(i.e., normal vs. flat feet) and the overestimation of ankle power (De Ridder et al., 2015; Dixon et al., 2012; Pothrat et al., 2015) have all been identified in the literature. To address these problems, a few multi-segment kinematic foot models have been proposed and tested on different clinical populations (Arampatzis, 2002; Carson et al., 2001; Jenkyn & Nicol, 2007; Kidder et al., 1996; Leardini et al., 1999; MacWilliams et al., 2003). But each of these multi-segment foot models have their own drawbacks.

By their nature, multi-segment foot models require the foot to be segmented so the motion between segments can be measured. Each foot model defines a different number of segments, or defines segments in different ways (Arampatzis, 2002; Jenkyn & Nicol, 2007; MacWilliams et al., 2003). An excessive number of foot segments requires an increased number of markers that must be placed on the foot (Arampatzis, 2002; MacWilliams et al., 2003). The foot may not have a big enough surface area to accommodate so many markers. Having too many markers and too many segments may also increase complexity of the model and increase the risk of errors in marker placement and post-processing.

Defining too few segments means some clinically important motions cannot be captured at all. Some foot models did not define a midfoot or hallux segment (Carson et al., 2001; Kidder et al., 1996;). But both of these segments are functionally important during walking because the midfoot serves as a shock absorber and transmitter for stabilization and the hallux pushes off at the end of the stance phase. The lack of a direct representation of the midfoot and hallux can lead to loss of information.

1.5 Human kinetics

Human kinetics is the study of the external and internal forces acting on the human body. External forces, such as ground reaction forces (GRFs) arise from sources external to the body. Internal forces are created within the body by muscles, ligaments, and joint contact (Zatsiorsky, 2002). Since it is difficult to measure tissue-specific internal forces in vivo, the summed effect on a joint is calculated using inverse dynamics, by which internal joint moments are calculated using Newton-Euler equations from the external forces acting on the body, as measured by force plates, segment accelerations and body segment inertia.
parameters (BSIPs) including the mass, the center of mass (CM), and the radius of
gyration (RG) or the moment of inertia. Joint powers describe energy generation and
absorption of muscles and ligaments around a joint. It is calculated as the dot product of
the joint force and the angular velocity.

Dempster’s cadaveric study (1955) provided the foundation of BSIPs and Zatsiorsky and
Seluyanov (1983) expanded these measures on cadavers. In 1990, Zatsiorsky et al. used
the gamma-ray scanning technique to determine BSIPs on living subjects. However, body
segments were defined by bony landmarks, which were sometimes distant from the
commonly used landmarks like the joint centers. Thus, De Leva (1996) adjusted the mean
relative CM positions and RGs reported by Zatsiorsky et al. and referenced them to joint
centers or commonly used reference points (Table 1-1). Both CM positions and RGs are
reported as percentages of the respective segment lengths.

Table 1-1: Adjusted mass percentages, CMs and RGs for body segments. Adapted
from “Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters” by De
Elsevier.

<table>
<thead>
<tr>
<th>Segment</th>
<th>Endpoint</th>
<th>Mass (%mass)</th>
<th>CM (%length)</th>
<th>Sagittal k (%length)</th>
<th>Transverse k (%length)</th>
<th>Longitudinal k (%length)</th>
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</thead>
<tbody>
<tr>
<td></td>
<td></td>
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<td>31.8</td>
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<td>36.2</td>
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</tr>
<tr>
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<td>41.51</td>
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1.5.1 Multi-segment foot kinetic analysis

While multi-segment kinematic foot analysis has been increasingly used in clinical gait studies, there are very few uses of multi-segment foot models for kinetic analysis of the forces acting on foot segments. This has been due to measurement and equipment limitations (Bruening et al., 2010; Buczek et al., 2006). Eerdekens et al.’s study (2019) pointed out that one-segment kinetic foot modelling may overestimate ankle joint kinetics in both healthy and pathological gait, leading to clinically relevant misinterpretations. This calls for the development of multi-segment kinetic foot models.

Scott and Winter (1993) proposed an eight-segment foot model with eight monocentric, single-DOF joints, and the force at each loading site were calculated from walking trials in which only a chosen portion of foot landed on the force plate. Abuzzahab and colleagues (1997) proposed a four-segment model with four joints: first MP, talonavicular, subtalar and talocrural joints, among which only the talonavicular joint was modeled with three DOFs, while the others were set up as single-DOF joints. Both models were complicated for clinical gait analysis and potentially oversimplified joint motions. MacWilliams et al. (2003) later proposed a nine-segment model to determine three-dimensional angles, external moments, and powers in eight joints. This model generated the most comprehensive kinetic data but required 19 markers on the foot and additional pedobarographic data. Bruening and colleagues (2012) described a method of using three different foot models to separate the ankle, the mid-tarsal joint and the first MP joint and measured GRFs by two adjacent force plates. However, the required visual targeting is likely to alter normal gait and thus limits the use of this method to healthy and minimally impaired populations (Bruening et al., 2012). Saraswat et al.’s study (2014) later proposed a three-segment kinetic foot model that used a force plate and pedobarographic data and applied it to pediatric planovalgus feet.

1.6 Knee osteoarthritis

Osteoarthritis (OA) is one of the most common chronic diseases worldwide, especially in the aging population. It is a whole-joint disease with a primary characteristic of cartilage degeneration. Symptoms of OA include pain, joint stiffness, crepitus, instability, and joint
deformity. Pain is activity-related but can become constant and more severe in advanced stages, leading to reduced quality of life and even functional disability. Joint space narrowing, subchondral bone sclerosis, osteophytes, periarticular ossicles, and altered shape of bone ends are considered evidence of radiographic OA (Kellgren & Laurence, 1057). Figure 1-6 shows an X-ray image of knee with and without osteoarthritis. The quantification of joint space narrowing combined with measurement of osteophytes prompted a widely used grading standard, the Kellgren-Lawrence (KL) grading scheme. OA can be categorized in to five KL grades: none (0), doubtful (1), minimal (2), moderate (3), severe (4). Patients in Grade 2 and above are diagnosed with radiographic OA (Kellgren & Lawrence, 1957). The exact etiology of OA remains unknown. Yet a great amount of evidence showed that multiple risk factors interact and contribute to OA onset and progression, including genetics, age, gender, excessive joint activity, previous trauma, overweight/obesity, the presence of other co-morbidities (Felson, 2009; Pereira et al., 2015). In general, the pathophysiology of OA is considered the outcome of a combination of mechanical, structural, biochemical processes.

OA at the knee is particularly common. The prevalence of knee OA is about 24% in the 60-64-year-olds and reaches 40% a decade later (Michael et al., 2010). The number continues to increase with age. Knee OA can happen at the tibiofemoral joint and/or the patellofemoral joint. The tibiofemoral joint contains the medial and lateral compartments. Medial compartmental OA is overall more prevalent than OA in the lateral compartment or the patellofemoral compartment, likely due to heavier medial loading (McAlindon et al., 1992). Medial compartment OA and the mechanical driver of disease were of particular interest in the study.
1.7 Knee mechanics

The mechanical axis of the femur is from the center of the femoral head to the center of the knee. The mechanical axis of the tibia is from the center of the knee to the center of the ankle. The anatomical axis of the femur is from the piriformis fossa to the center of the knee. The hip-knee-ankle angle (HKA) measures the angle between the mechanical axes of the femur and the tibia in the frontal plane. It is an indicator of the lower limb alignment. Genu valgum at the joint is when the apex of the joint points more medially, and genu varum is when the joint points more laterally. For a neutral knee alignment, the mechanical axis of the lower limb (from the center of femoral head to the center of ankle) passes through the center of the knee. Figure 1-7 shows the load bearing axis (mechanical axis) and HKA of varus, neutral and valgus knee alignment. HKA is 1-1.5° varus and the angle between the mechanical and anatomical axes of the femur is about 5° (Hsu et al., 1990; Moreland et al., 1987). When in a neutral standing position, approximately 60% of the body weight passes through the medial compartment of the knee. For a varus knee alignment, however, the mechanical axis shifts medially from the center of knee and thus increases load distribution on the medial side, driving cartilage degeneration in the medial compartment. Many studies show that varus knee alignment is associated with
increased risk of incident medial compartment OA and structural progression in the medial compartment (Palmer et al., 2020; Sharma et al., 2010).

![Figure 1-7: The mechanical axis (LBA) and HKA in knee varus and in knee valgus. Adapted from “Static knee alignment and its association with radiographic knee osteoarthritis,” by Cooke, T. D., 2007, Osteoarthritis and cartilage, 15(7), 844–847. Copyright 2007 by © Elsevier.]

1.8 High tibial osteotomy

Among many orthopaedic treatments for knee OA, high tibial osteotomy (HTO) is frequently implemented in young and active patients (40-60 years old) to treat unicompartmental knee OA. HTO is a surgical procedure to realign the tibia and restore loading balance of the knee. It is used to correct knee varus into a slight valgus alignment (3-5° valgus) such that the mechanical axis of the lower limb passes through the Fujisawa point, which is 62% of the tibial plateau width when measured from the edge of the medial tibial plateau. The correcting angle is calculated preoperatively (Sabzevari et al., 2016). Lateral closing wedge osteotomy (LCWO) and medial opening wedge osteotomy (MOWO) are two commonly used surgical techniques. A LCWO removes a wedge of bone from a lateral incision on the fibula and the tibia. A MOWO inserts a spacer plate or autograft through a medial incision on the tibia. Figure 1-8 is an illustration of LCWO (a) and MOWO (b). According to Wang et al.’s meta-analysis (2018) on HTO, LCWO and
MOWO are similar to each other in terms of post-operative knee pain, HKA over five years, knee functions, complications, and survival rates. However, LCWO involves two bone cuts and can result in higher risk of peroneal nerve injury, leg shortening, and frontal-plane-only correction. MOWO, on the other hand, has only one bone cut but may lead to higher risk of non-union or delayed union due to the use of bone substitutes. It may also increase posterior tilt of the tibial plateau and decrease patellar height.

![Figure 1-8: LCWO (a) and MOWO (b). Adapted from “Experimental modular stand used for studies of the High Tibial Osteotomy,” by Cofaru, I. I., 2017, MATEC Web of Conferences, 121, 4004–. CC BY 4.0.](image)

1.9 Foot/ankle alignment and movements pre-/post-HTO

1.9.1 Foot/ankle alignment with knee varus

Knee malalignment in the frontal plane has a considerable influence on foot and ankle alignment. In cadavers, Olerud (1985) and Ting et al. (1987) discovered that varus or valgus angulation of the tibia was compensated by pronation or supination of the subtalar joint, respectively. Ting and colleagues (1987) further pointed out that such compensation led to subtalar joint restriction, or reduced ROM of the subtalar joint. This phenomenon was also discovered in the clinical population as Gao et al.’s radiographic study (2016) on 149 patients with symptomatic knee OA discovered that knee varus and valgus can induce ankle tilt directly on the pathologic side and indirectly on the non-
pathologic side. Figure 1-9 is an example of knee varus and induced ankle valgus on a knee OA patient’s X-ray image.

A case-control study in elderly Japanese women found that compared to healthy controls, individuals with medial knee OA showed significant varus knee alignment, pronated feet, as well as foot eversion with a larger passive ROM, suggesting that ankle valgus was closely related to knee varus (Matsumura et al., 2020). In patients with end-stage knee OA and receiving total knee replacement, a linear correlation was present between knee alignment and ankle/hindfoot alignment, meaning the ankle valgus increased with the progression of knee varus (Norton et al., 2015; Xie et al., 2019). Unfortunately, data from the Osteoarthritis Initiative showed that 25% of knee OA patients experience either ipsilateral or bilateral foot pain, negatively affecting the overall health and function (Paterson et al., 2015). The malalignment of knee and ankle is also associated with ankle degeneration and may contribute to the development and progression of ankle OA (Gao et al., 2016; Xie et al., 2018).

Figure 1-9: X-ray image showing the lower limb alignment. Adapted from “The effect of varus knee deformities on the ankle alignment in patients with knee osteoarthritis,” by Xie, K., Han, X., Jiang, X., Ai, S., Dai, K., Yu, Z., Wu, H., Qu, X., & Yan, M., 2019, Journal of Orthopaedic Surgery and Research. CC BY 4.0.
1.9.2 Foot/ankle motion with knee varus

Knee varus alters foot and ankle mechanics during walking. Young adults with asymptomatic knee varus showed greater peak hindfoot eversion than healthy controls, which was also a sign seen in medial knee OA patients (Barrios et al., 2009). Levinger and colleagues (2012) used the Oxford Foot Model to study gait kinematics in medial OA patients and found that compared with controls, medial OA patients contacted the ground with a more everted hindfoot position relative to the tibia and demonstrated less foot mobility throughout the gait cycle. Using the IOR multi-segment foot model, Arnold et al.’s study (2014) on medial OA patients with knee varus for the first time reported that the increased foot rigidity was due to the reduced frontal plane ROM of the midfoot. The hallux of medial OA patients also had a reduced ROM in the sagittal plane compared with the controls. Data from the same cohort indicated that greater hindfoot eversion was associated with lower knee adduction moments in the stance phase, indicating reduced medial compartment loading in the knee (Levinger et al., 2013).

In summary, foot eversion seems to be a compensatory motion for varus knee alignment adopted by medial OA patients and asymptomatic individuals to reduce medial loading of the knee in walking.

1.9.3 Foot/ankle alignment and movements post-HTO

Radiographic and gait studies of patients receiving HTO show that HTO is successful in correcting knee varus, increasing walking speed and stride length, decreasing knee adduction moment, and normalizing dynamic knee function parameters such as load distribution (Birmingham et al., 2009; Ivarsson et al., 1989; Lind et al., 2013; Wada et al., 1998; Wang et al., 1990). However, research examining the relationship between lower limb alignment and ankle joint biomechanics pre-/post-HTO is limited. Based on the current understanding, the change of foot/ankle alignment following HTO is less predictable than knee alignment. A few studies showed that the hindfoot valgus deviation in medial knee OA patients significantly decreased post-HTO and the ankle joint orientation was more parallel to the ground (G. W. Choi et al., 2017; J. Y. Choi et al., 201; Takeuchi et al., 2008). On the contrary, a case study of a 50-year-old female
reported aggravated ankle valgus and lateral ankle pain post-HTO (Jeong & Soohoo, 2014). Choi et al.’s study (2017) later found that when preoperative knee varus is mild, ankle valgus may even increase post-operation.

In addition, Suero et al. (2014) described in a cadaveric study that the intraarticular contact pressure of the ankle may progressively increase with the increase of HTO correction angle. The redistribution of pressure over a smaller contact surface may have implications on ankle biomechanics and potentially lead to ankle degeneration and pain.

Recently, Kyung and colleagues (2021) used a multi-segment foot model to study the foot and ankle motion in pre-/post-HTO patients with medial knee OA. In the pre-HTO state, the forefoot was pronated relative to the hindfoot but the hindfoot was supinated relative to the tibia throughout the gait cycle. It suggested that the valgus compensation for knee varus did not happen at the ankle but in the midfoot. And patients’ gait improved post-HTO. This finding contradicted the aforementioned findings and challenged the current understanding of the location of compensatory movements within the foot.

1.10 Rationale and objectives

Existing multi-segment foot models have several problems that limit their clinical use. These include complexity of marker placement, gait-altering testing requirements (e.g., visual targeting), too many segments, or segments with limited clinical relevance. These problems impede the use of multi-segment foot models in large clinical populations either as a research tool or as a clinical assessment tool. The primary objective of this thesis research is to develop a novel multi-segment foot model that is clinically feasible and enables both kinematic and kinetic analysis of large patient groups. This research study will develop such a model and test its reliability and validity.

The DuPont multi-segment foot model, of which the repeatability has been tested in healthy adults and children (Seo et al., 2014; Kim et al., 2018), was chosen as the foundation and modified for this study to create a midfoot segment in addition to the hindfoot, forefoot, and hallux. The Cardan convention, which describes joint motions by sequential rotations of a segment about body-fixed axes of its proximal segment, was
used to describe the relative orientations between segments (Tupling & Pierrynowski, 1987).

The novel multi-segment foot model is created to measure the following motions during the gait cycle:

- Hindfoot dorsi/plantarflexion
- Hindfoot inversion/eversion
- Hindfoot internal/external rotation
- Midfoot inversion/eversion with respect to the hindfoot
- Midfoot internal/external rotation with respect to the hindfoot
- Midfoot dorsi/plantarflexion with respect to the hindfoot
- Forefoot inversion/eversion with respect to the midfoot
- Forefoot internal/external rotation with respect to the midfoot
- Forefoot dorsi/plantarflexion with respect to the midfoot
- Hallux dorsi/plantarflexion with respect to the forefoot
- Hallux ab/adduction with respect to the forefoot
- Rise and fall of the medial longitudinal arch (MLA)

For each motion, the outcome measures were the intersegmental rotation angle, the joint moment, the joint powers, and the MLA height-to-length ratio. It is hypothesized that this model will be able to measure all these outcome measures within an existing optical motion capture laboratory with existing equipment.

The second objective is to evaluate the reliability and the validity of the novel multi-segment foot model. The coefficient of multiple correlation (CMC) will be calculated as an indicator of the model’s repeatability. It is hypothesized that for all outcome measures, test-retest CMCs and within-subject CMCs will be greater than 0.7, and outcome measures of this model will be in accordance with the other multi-segment foot models.

The third objective is to test the novel multi-segment foot model in a patient group and compare the results to a normal population. Pre-/post HTO patients with medial knee OA and knee varus is a good clinical context where we can apply the multi-segment foot model because it is anticipated that their foot/ankle kinematics and kinetics are compromised during walking. It is hypothesized that pre-HTO patients will demonstrate greater forefoot and hindfoot eversion and lower MLA height/length ratio compared to
the healthy population, and post-HTO patients will demonstrate improved or close-to-normal foot/ankle movements due to the correction of knee varus.

The kinematic and kinetic information of patients enabled by this novel multi-segment foot model will be important to future research and clinical management of pre-/post-surgery knee OA patients. Based on the current knowledge, very few multi-segment kinetic foot analysis has been done in clinical populations and none in patients with medial knee OA and knee varus. Moreover, multi-segment foot models used in previous studies only divided the foot into three segments (hallux, forefoot and hindfoot) and failed to directly measure midfoot motions. As a result, this foot model will be applied to pre-/post-HTO patients with medial knee OA and knee varus and compare their foot/ankle motions with healthy controls.
Chapter 2

2.1 Equipment
The testing took place in the Wolf Orthopaedic Biomechanics Laboratory at the Fowler Kennedy Sport Medicine Clinic. Within the lab, twelve three-dimensional (3D) real-time, digital tracking cameras (Kestrels & Raptor-12HS, Motion Analysis Corporation, Santa Rosa, CA, USA) were placed symmetrically around a ten-meter walkway. The sampling rate of the cameras was 60 frames per second. A force plate mounted into the floor was used to measure the ground reaction forces (GRFs) of a foot-strike with a sampling rate of 1200 Hz. Raw data were filtered by a low-pass Butterworth filter at a cut-off frequency of 6 Hz. The motion capture acquisition and editing software used was Cortex 7.0 (Motion Analysis Corporation, Santa Rosa, CA, USA), which is a real-time monitoring and post-processing application. It offers tools including Skeleton Builders for creating and defining body segments from joint to joint and KinTools RT for kinematic and kinetic analysis.

2.2 Subjects
Ten young, healthy adults (3 males; average age 22±2 years old; average height 169.2±8.5cm; average weight 64±11kg; average BMI 22.3±2.6kg/m²) volunteered for the study. None of the participants had any musculoskeletal disorders or previous musculoskeletal injuries or any ongoing symptoms. None of the participants had any obvious lower limb malalignment. The demographics of the normal participants are listed in Table 2-1 below.

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This study was approved by Western University’s Research Ethics Board. Written and verbal informed consents were obtained from each participation before testing.

### 2.3 Data collection

#### 2.3.1 Calibration

The standard calibration process was performed before each data collection. The twelve cameras were turned on and connected to Cortex. After starting the “Initial Calibration” program, an L-frame object with auto-reflective markers on it was positioned at a designated location on the ground at the centre of the laboratory to define the origin and the x, y, z axes of the global coordinate system. After this, a wand with four auto-reflective markers was waved through the laboratory volume to calibrate the volume within which the test subjects would be moving. Once these two steps had been completed, the laboratory was ready for data collection.

#### 2.3.2 Marker placement

A set of auto-reflective, single markers were attached to the skin of each test participant in this study. The configuration of the marker set was the full body modified Helen Hayes marker set and the DuPont foot model marker set, which comprised a total of 40 markers. These markers enabled full-body tracking. The landmarks identified on the foot were as follows: the center of the proximal posterior calcaneus (top calcaneus), the center of the distal posterior calcaneus (bottom calcaneus), medial and lateral malleoli, navicular, talar head, cuboid, heads of the first and fifth metatarsals, between the heads of second and third metatarsals, and the hallux nail bed. The marker on talar head was added to the original DuPont model so that the midfoot had three non-colinear markers on it. These landmarks were used to construct a segment-fixed anatomical coordinate system to each of the segments of the DuPont foot model using the Cortex software. The complete auto-
reflective marker set is shown on a test participant in Figure 2-1 below. Figure 2-2 shows a close-up view of the marker set used for the DuPont foot model.

Figure 2-1: A participant with full markers attached. Photo courtesy of W. Li.

Figure 2-2: Modified DuPont foot model with annotations. LM: lateral malleolus, MM: medial malleolus, TC: top calcaneus, BC: bottom calcaneus, TH: talar head, NV: navicular bone, CB: cuboid, 1MTH: first metatarsal head, 5MTH: fifth metatarsal head, 23T: between the heads of second and third metatarsals. Photo courtesy of W. Li.
2.3.3 Initial static trial

After applying all markers, a static standing trial was performed. Each participant stood statically on the force plate in the anatomical standing position (shown in Figure 2-1 above) with the body standing up straight, arms by sides of the body and palms facing forward, feet shoulder width apart. The height and weight of each participant was recorded. The static trial was used to determine the neutral or “zero” position of all body segments for that individual.

2.3.4 Dynamic trials

Only one force plate was used in the dynamic trials. After the static trial, each participant was instructed to walk at their self-selected pace along the 10 m walkway as naturally as possible without looking down at the ground or the force plate. Before recording each trial, the line from which the participant started walking was adjusted so that the right or the left foot would strike the center of the force plate. Six dynamic walking trials were captured for each subject, three trials with the left foot striking the force plate and three with the right foot striking the force plate.

2.4 Data processing

Post-processing of data consisted of two steps. The first step was performed in Cortex and the second step was in Microsoft Excel.

2.4.1 Processing in Cortex

After creating a template for the modified Helen Hayes plus the modified DuPont foot model, markers in both the static and dynamic trials were automatically identified by the template within Cortex. Tracking data were then manually cleaned to make sure all markers were correctly identified and tracked. Small gaps in marker trajectories (i.e., fewer than five frames) were fixed by Cubic Join and larger gaps were filled by Virtual Join. Virtual markers at the joint centers were calculated from existing surface markers. For example, the ankle joint center was defined as the midpoint of the medial and lateral malleoli. After cleaning the marker trajectories, gait events were automatically identified by BioFeedTrak Event Timeline and then manually adjusted for accuracy. The data
capture was trimmed down to one gait cycle from the HS on the force plate to the next HS of the same foot.

2.4.2 Segmentation
In this study, we used the Skeleton Builders in Cortex to create body and foot segments. To accurately describe a foot segment’s motion at a joint, a Parent Segment, which was conventionally set as the proximal segment at that joint, was defined as the reference. Thus, hallux moves relative to forefoot, the forefoot relative to midfoot, midfoot relative to hindfoot, and hindfoot relative to the tibia. Cortex defines a local coordinate system by the Original Marker, the Bone (Long) Axis Marker and the Plane Marker. The Bone axis (from the Original Marker to the Bone Axis Marker) was set as the z axis, and the length of a segment was determined by the distance between the Original Marker and the Bone Axis Marker. The Plane Marker, together with the Original Marker and the Bone Axis Marker, defined the YZ plane, and thus the x axis was perpendicular to the YZ plane according to the right-hand rule. The y axis was orthogonal to the x and z axes. For all foot segments, the x axes were medio-lateral pointing to the left, the y axes were vertical pointing upwards, and the z axes pointed anteriorly. RX, RY and RZ Offsets are angles in degrees from -360 to 360 to orient the local coordinate systems so that they align in the same directions. The inter-segmental rotations in the sagittal, frontal, and transverse planes were calculated by the X-Z-Y-Cardan angles in degrees, of which the X-Z-Y rotation sequence corresponded with the following foot motion sequence:

  X: dorsiflexion/plantar flexion
  Z: inversion/eversion
  Y: internal rotation/external rotation

The lower body segment definitions are shown in Table 2-2 and the segment modeling is shown in Figure 2-3 and Figure 2-4.

Table 2-2: Lower body segment definitions in Cortex. Global: the global coordinate system; V_: virtual markers; JC: joint center; BC: bottom calcaneus; TC: top calcaneus; TH: talar head; MidNC: the midpoint between navicular and cuboid; 1-5MTH: the midpoint between the 1st and 5th metatarsal heads; NV: navicular; 23T: between the 2nd and 3rd metatarsal heads.
<table>
<thead>
<tr>
<th>Segment</th>
<th>Parent</th>
<th>Origin Marker</th>
<th>Long Axis Marker</th>
<th>Plane Marker</th>
<th>RX Offset</th>
<th>RY Offset</th>
<th>RZ Offset</th>
</tr>
</thead>
<tbody>
<tr>
<td>pelvis</td>
<td>Global</td>
<td>V_Pelvis_Origin</td>
<td>V_Mid_Hip</td>
<td>V_Sacral</td>
<td>0</td>
<td>0</td>
<td>180</td>
</tr>
<tr>
<td>R.Thigh</td>
<td>pelvis</td>
<td>V_R.Hip_JC</td>
<td>V_R.Knee_JC</td>
<td>R.Knee</td>
<td>0</td>
<td>0</td>
<td>-90</td>
</tr>
<tr>
<td>L.Thigh</td>
<td>pelvis</td>
<td>V_L.Hip_JC</td>
<td>V_L.Knee_JC</td>
<td>L.Knee</td>
<td>0</td>
<td>0</td>
<td>90</td>
</tr>
<tr>
<td>R.Shank</td>
<td>R.Thigh</td>
<td>V_R.Knee_JC</td>
<td>V_R.Akle_JC</td>
<td>R.Akle</td>
<td>0</td>
<td>0</td>
<td>-90</td>
</tr>
<tr>
<td>L.Shank</td>
<td>L.Thigh</td>
<td>V_L.Knee_JC</td>
<td>V_L.Akle_JC</td>
<td>L.Akle</td>
<td>0</td>
<td>0</td>
<td>90</td>
</tr>
<tr>
<td>R.Hindfoot</td>
<td>R.Shank</td>
<td>R.BC</td>
<td>V_R.Akle_JC</td>
<td>R.TC</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>L.Hindfoot</td>
<td>L.Shank</td>
<td>L.BC</td>
<td>V_L.Akle_JC</td>
<td>L.TC</td>
<td>0</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>R.Midfoot</td>
<td>R.Hindfoot</td>
<td>V_R.Akle_JC</td>
<td>R.TH</td>
<td>V_R.MidNC</td>
<td>0</td>
<td>0</td>
<td>180</td>
</tr>
<tr>
<td>L.Midfoot</td>
<td>L.Hindfoot</td>
<td>V_L.Akle_JC</td>
<td>L.TH</td>
<td>V_L.MidNC</td>
<td>0</td>
<td>0</td>
<td>180</td>
</tr>
<tr>
<td>R.Foorefoot</td>
<td>Midfoot</td>
<td>V_R.MidNC</td>
<td>R.1-5MTH</td>
<td>R.NV</td>
<td>0</td>
<td>0</td>
<td>90</td>
</tr>
<tr>
<td>L.Foorefoot</td>
<td>Midfoot</td>
<td>V_L.MidNC</td>
<td>L.1-5MTH</td>
<td>L.NV</td>
<td>0</td>
<td>0</td>
<td>-90</td>
</tr>
<tr>
<td>R.Hallux</td>
<td>R.Foorefoot</td>
<td>1st MTH</td>
<td>R.Hallux</td>
<td>R.23T</td>
<td>0</td>
<td>0</td>
<td>-90</td>
</tr>
<tr>
<td>L.Hallux</td>
<td>L.Foorefoot</td>
<td>1st MTH</td>
<td>L.Hallux</td>
<td>L.23T</td>
<td>0</td>
<td>0</td>
<td>90</td>
</tr>
</tbody>
</table>

Figure 2-3: Foot segmentation modeling in Cortex 7.0.
2.4.3 Mass Model Editor

The relative CMs (centers of mass) and RGs (radii of gyration) of body segments were set according to De Leva’s report on BSIPs (1996). The Head/Neck segment was excluded because head markers were not included in this study. This explained why the segment sums for female and male were 0.9331 and 0.929 respectively rather than 1.0 (as shown in Figure 2-6).
According to Drillis et al.’s study (1964), the foot was subdivided into the base of foot, the middle foot and five toes (shown in Figure 2-5). The volume of the base of foot and the middle foot together took up 13.77/1000 of the total body volume, and the five toes took up 2.70/1000 of the total body volume for males. For the female counterpart, the numbers were 10.92/1000 and 2.40/1000 of the total body volume, respectively.

Figure 2-5: Body segments. Adapted from “Body segment parameters; a survey of measurement techniques,” by Drillis, R., Contini, R., & Bluestein, M., 1964, Artificial limbs, 8, 44–66. (http://www.oandplibrary.org/al/pdf/1964_01_044.pdf). In the public domain.
Although based on De Leva (1996) the CM of the whole foot is 0.4014 of the length, the mass of foot is not evenly distributed. The hindfoot and midfoot consist primarily of bones (calcaneus, navicular, cuneiforms, cuboid) while soft tissues account for a higher proportion in the forefoot and toes. Therefore, without further information, we assumed that “the base of foot” corresponded to the hindfoot, and “the middle foot” equaled to the forefoot plus the midfoot in our study. Drawing on the experience of Dixon et al.’s study (2012) and Bruening et al.’s study (2012), the volume of the hindfoot was arbitrarily set as half of the overall base of foot and middle foot volume, and the volumes of the forefoot and midfoot were further divided into half. The hallux segment represented the five toes. The masses of foot segments were partitioned in proportion to the volume of each segment. The RGs of foot segments were determined according to De Leva (1996). The CMs were arbitrarily set to half the length of segments. Parameters for Mass Model Editor in Cortex 7.0 are shown in Figure 2-7.
2.4.4 Processing in Excel

The second step of post-processing was performed in Excel. Both the static and dynamic data were exported as *.kin and *.trc files into Excel for further analysis. Static data were used as the starting/neutral position for an individual. Relative rotations between foot segments in each dynamic trial were calculated by subtracting static values from dynamic values. The length of the MLA was defined as the magnitude of the vector formed by marker BC (bottom calcaneus) and marker 1MTH (first metatarsal head). The height of MLA was the perpendicular distance from marker NV (navicular) to MLA length. MLA height/length ratio was calculated for a gait cycle. Kinetic measures of joint moments and powers were normalized to the participant’s body mass (Nm/kg and W/kg, respectively). Gait cycles in dynamic trials may contain different numbers of frames for an individual due to different walking speeds. Interpersonal variation may even be greater. To enable comparison between groups, all kinematic and kinetic curves in each gait cycle were normalized to 100% (101 data points) by resampling and interpolating in Excel. Final kinematic and kinetic measures were averaged over three walking trials for all ten participants.
Since the right and left feet showed mirrored movement patterns for all participants as expected, the right foot was chosen to represent the individual’s walking. Within-subject coefficients of correlation (CMCs) and test-retest CMCs were calculated over the three walking trials of each of the ten participants to test the multi-segment foot model’s reliability. The strength scale of CMC was set such that $R<0.3$ indicates a weak correlation, $0.3\leq R<0.7$ indicates a moderate correlation, and $R\geq0.7$ indicates a strong to very strong correlation (Akoglu, 2018).

The multi-segment kinematic foot model was compared with the original DuPont model (Lee et al., 2017; Seo et al., 2014), Leardini’s model (1999) and Jenkyn and Nicol’s model (2007). The multi-segment kinetic foot model was compared with Bruening et al.’s model (2012) and Saraswat et al.’s model (2014).

### 2.5 Results

#### 2.5.1 Kinematic analysis

The averaged walking speed of ten participants was $1.22\pm0.17$ m/s and the average stride length was $1.31\pm0.15$ m. Averaged over ten participants, intersegmental rotation angle curves and the MLA height/length ratio curve in a normalized gait cycle are shown in Figure 2-8.
Figure 2-8: Rotation angle curves of each foot segment and MLA h/l ratio curve in a gait cycle.

For the hindfoot, the ankle joint in the sagittal plane (Figure 2-8, top left) was in slight plantar flexion (-3.15°±2.83) at heel strike, continued to plantar flex until foot flat (-9.03°±3.43), and moved into gradual dorsiflexion until toe off (TO). In the transverse plane (Figure 2-8, top right), the hindfoot was neutral (-0.92°±1.07) at heel strike, slightly internally rotated until midstance, moved into external rotation, and peaked at the start of pre-swing (-2.31°±1.37), and remain mostly externally rotated in the swing phase. In the frontal plane (Figure 2-8, top center), the subtalar joint started the gait cycle in a neutral
position (0.68°±4.01), everted (-2.03°±1.07) during midstance, peaked in inversion (4.52°±3.65) right before TO and remained inverted in the swing phase.

For the midfoot, in the sagittal plane (Figure 2-8, second row left) it was almost neutral (-1.52°±1.16) at heel strike, slightly dorsiflexed until midstance, and reached a peak of plantar flexion (-5.89°±2.64) right before TO. In the frontal and the transverse planes (Figure 2-8, second row center and right), the midfoot stabilized around its neutral position throughout the gait cycle.

For the forefoot (Figure 2-8, third row left), it started at a neutral position (0.73°±1.61) in the sagittal plane and dorsiflexed until TO. In the transverse plane (Figure 2-8, third row right), the forefoot was slightly internally rotated (2.23°±2.78) at heel strike (HS), became neutral during midstance and reached a peak of internal rotation at TO (5.50°±4.51). In the frontal plane (Figure 2-8, third row center), the forefoot was slightly inverted (1.73°±2.05) at HS, became neutral during midstance and reached a peak of inversion (2.06°±2.51) at TO.

The hallux demonstrated large ROMs (Figure 2-8, bottom row). In the sagittal plane, it was dorsiflexed (13.62°±8.17) at HS and reached its peak of dorsiflexion (38.47°±5.08) at TO. In the transverse plane, it started slightly adducted (-6.10°±3.33) and reached its peak of adduction (-11.37°±4.35) at TO.

The MLA height/length ratio (Figure 2-8, bottom row right) gradually declined during the stance phase as the foot became more flexible to support the body weight, reached its valley right before TO, then restored its height after TO into the swing phase.

2.5.2 Kinetic analysis

Averaged over ten participants, intersegmental joint moment and power curves in a normalized gait cycle are shown in Figure 2-9. Joint moments and powers were normalized by the body mass in kilograms.
Figure 2-9: Joint moment curves and joint power curves of each foot segment in a gait cycle.

Peak joint moments for all foot segments in all three planes and maximal and minimal joint powers are summarized in Table 2-3. All measures reached their peaks (the second peak for hallux) in the beginning of pre-swing (at around 50%) and became zero after TO (at around 61%) as the foot was no longer in contact with the ground.

Table 2-3: Peak joint moments and maximal and minimal joint powers.

<table>
<thead>
<tr>
<th>Segment</th>
<th>Joint power Max, Min (W/kg)</th>
<th>Joint motion</th>
<th>Joint moment Max (Nm/kg)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hindfoot</td>
<td>3.14, -1.01</td>
<td>Plantar flexion (+)/Dorsiflexion (-)</td>
<td>2.04</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Internal (+)/External rotation (-)</td>
<td>-0.46</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Inversion (+)/Eversion (-)</td>
<td>0.08</td>
</tr>
<tr>
<td>Midfoot</td>
<td>0.82, -0.09</td>
<td>Plantar flexion (+)/Dorsiflexion (-)</td>
<td>1.32</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Internal (+)/External rotation (-)</td>
<td>-0.17</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Inversion (+)/Eversion (-)</td>
<td>-0.15</td>
</tr>
<tr>
<td>Forefoot</td>
<td>0.55, -0.47</td>
<td>Plantar flexion (+)/Dorsiflexion (-)</td>
<td>0.98</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Internal (+)/External rotation (-)</td>
<td>-0.22</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Inversion (+)/Eversion (-)</td>
<td>-0.14</td>
</tr>
<tr>
<td>Hallux</td>
<td>0.09, -1.51</td>
<td>Plantar flexion (+)/Dorsiflexion (-)</td>
<td>-1.03</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Abduction (+)/adduction (-)</td>
<td>0.34</td>
</tr>
</tbody>
</table>
2.6 Reliability

Two participants were tested for a second time on a different day to calculate the test-retest repeatability. Within-subject CMCs and test-retest CMCs for intersegmental rotation angle curves and joint moment and power curves were calculated over three walking trials from all ten participants (Table 2-4, Table 2-5). For the rotation angle curves, all intersegmental motions showed strong test-retest and within-subject reliability ($R \geq 0.7$), except that hindfoot inversion/eversion, midfoot internal/external rotation, and midfoot inversion/eversion showed moderate test-retest reliability ($0.3 \leq R < 0.7$), and that midfoot inversion/eversion showed moderate within-subject reliability ($R = 0.66$). The reduced repeatability of midfoot motions is likely because the midfoot is more stable than the other foot segments and lacks a typical movement pattern during walking, especially in the frontal and transverse planes. Joint moments for all foot segments in all three planes and all joint powers showed strong test-retest and within-subject reliability ($R \geq 0.8$).

**Table 2-4: Test-retest CMC±SD and within-subject CMC±SD for rotation angle curves for intersegmental foot motions. * denotes $0.3 \leq R < 0.7$ or moderate reliability.**

<table>
<thead>
<tr>
<th>Segment</th>
<th>Joint motion</th>
<th>Test-retest CMC±SD</th>
<th>Within-subject CMC±SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hindfoot</td>
<td>Dorsiflexion (+) / Plantar flexion (-)</td>
<td>0.95±0.05</td>
<td>0.96±0.02</td>
</tr>
<tr>
<td></td>
<td>Internal (+) / External rotation (-)</td>
<td>0.75±0.16</td>
<td>0.89±0.06</td>
</tr>
<tr>
<td></td>
<td>Inversion (+) / Eversion (-)</td>
<td>0.64±0.05*</td>
<td>0.75±0.13</td>
</tr>
<tr>
<td>Midfoot</td>
<td>Dorsiflexion (+) / Plantar flexion (-)</td>
<td>0.80±0.12</td>
<td>0.90±0.04</td>
</tr>
<tr>
<td></td>
<td>Internal (+) / External rotation (-)</td>
<td>0.34±0.01*</td>
<td>0.79±0.08</td>
</tr>
<tr>
<td></td>
<td>Inversion (+) / Eversion (-)</td>
<td>0.49±0.34*</td>
<td>0.66±0.15*</td>
</tr>
<tr>
<td>Forefoot</td>
<td>Dorsiflexion (+) / Plantar flexion (-)</td>
<td>0.94±0.06</td>
<td>0.94±0.03</td>
</tr>
<tr>
<td></td>
<td>Internal (+) / External rotation (-)</td>
<td>0.91±0.05</td>
<td>0.93±0.03</td>
</tr>
<tr>
<td></td>
<td>Inversion (+) / Eversion (-)</td>
<td>0.88±0.02</td>
<td>0.84±0.04</td>
</tr>
</tbody>
</table>
### Table 2-5: Test-retest CMC±SD and within-subject CMC±SD for joint moment and power curves.

<table>
<thead>
<tr>
<th>Segment</th>
<th>Joint power curves</th>
<th>Joint motion</th>
<th>Joint moment curves</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Within-subject CMC±SD</td>
<td>Test-retest CMC±SD</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>Dorsiflexion (+) / Plantar flexion (-)</td>
</tr>
<tr>
<td>Hindfoot</td>
<td></td>
<td></td>
<td>Dorsiflexion (+) / Plantar flexion (-)</td>
</tr>
<tr>
<td>Midfoot</td>
<td>0.83±0.08</td>
<td>0.96±0.02</td>
<td>Internal (+) / External rotation (-)</td>
</tr>
<tr>
<td>Forefoot</td>
<td>0.85±0.09</td>
<td>0.92±0.08</td>
<td>Dorsiflexion (+) / Plantar flexion (-)</td>
</tr>
<tr>
<td>Hallux</td>
<td>0.80±0.11</td>
<td>0.85±11</td>
<td>Dorsiflexion (+) / Plantar flexion (-)</td>
</tr>
</tbody>
</table>

#### 2.7 Validation

##### 2.7.1 Multi-segment foot kinematics

Ranges of motion (ROMs) of each foot segment in the current model were compared with three other multi-segment foot models in the literature as summarized in Table 2-6. The original DuPont foot model did not contain a midfoot segment, and thus forefoot motions were measured relative to the hindfoot (Lee et al., 2017). Leardini et al.’s model (1999) had the same foot segmentation as the current model and reported ROMs in all three planes but did not include MLA index. Jenkyn and Nicol’s model (2007) had a similar segmentation without the hallux and selectively reported intersegmental foot motions. The MLA height/length ratio was normalized to the length of MLA.
Table 2-6: Comparison of ROMs in the current model with other foot models in the literature (Lee et al., 2017; Leardini et al., 1999; Jenkyn & Nicol, 2007)

<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>ROM (Max, Min)</td>
<td>ROM (Max, Min)</td>
<td>ROM (Max, Min)</td>
<td>ROM (Max, Min)</td>
</tr>
<tr>
<td>Hindfoot</td>
<td>Dorsiflexion (+)/Plantar flexion (-)</td>
<td>26.30 (7.65, -18.65)</td>
<td>25.5 (14.0, -11.5)</td>
<td>12.0 (4.7, -7.3)</td>
<td>15 (5, -10)</td>
</tr>
<tr>
<td></td>
<td>Internal (+)/External rotation (-)</td>
<td>2.71 (0.28, -2.44)</td>
<td>12.8 (11.3, -1.5)</td>
<td>10.7 (4.5, -6.2)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Inversion (+)/Eversion (-)</td>
<td>6.56 (4.53, -2.03)</td>
<td>13.2 (9.8, -3.4)</td>
<td>9.5 (2.2, -7.3)</td>
<td>10 (5, -5)</td>
</tr>
<tr>
<td>Midfoot</td>
<td>Dorsiflexion (+)/Plantar flexion (-)</td>
<td>6.46 (0.77, -5.69)</td>
<td></td>
<td>8.0 (2.5, -5.5)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Internal (+)/External rotation (-)</td>
<td>3.71 (2.10, -1.61)</td>
<td>2.8 (2.1, -0.7)</td>
<td>8 (6, -2)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Inversion (+)/Eversion (-)</td>
<td>3.93 (0.27, -3.66)</td>
<td>4.4 (3.0, -1.4)</td>
<td>11 (8, -3)</td>
<td></td>
</tr>
<tr>
<td>Forefoot</td>
<td>Dorsiflexion (+)/Plantar flexion (-)</td>
<td>14.17 (8.23, -5.95)</td>
<td>14.8 (3.5, -11.2)</td>
<td>11.5 (12.6, 1.1)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Internal (+)/External rotation (-)</td>
<td>7.48 (5.50, -1.98)</td>
<td>10.5 (-0.5, -11.0)</td>
<td>6.9 (3.7, -3.2)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Inversion (+)/Eversion (-)</td>
<td>2.82 (2.44, -0.39)</td>
<td>8.7 (11.8, 3.1)</td>
<td>14.3 (13.3, -1.0)</td>
<td>12 (15, 3)</td>
</tr>
<tr>
<td>Hallux</td>
<td>Dorsiflexion (+)/Plantar flexion (-)</td>
<td>37.29 (38.47, 1.18)</td>
<td>40.3 (30.8, -9.5)</td>
<td>25.6 (26.8, 1.2)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Abduction (+)/Adduction (-)</td>
<td>11.63 (0.27, -11.37)</td>
<td>10.4 (0.9, -11.3)</td>
<td>15.7 (1.0, -14.7)</td>
<td></td>
</tr>
<tr>
<td></td>
<td>MLA height/length ratio</td>
<td>0.03 (0.17, 0.14)</td>
<td>0.06 (0.25, 0.19)</td>
<td>0.4 (1.3, 0.9)</td>
<td></td>
</tr>
</tbody>
</table>

Due to the creation of the midfoot segment in the four-segment modified DuPont model, ROMs in the transverse and frontal planes reported in the current study are generally smaller than those reported from the original three-segment DuPont model (except hallux abduction/adduction) on healthy, young females in Lee and colleagues’ study (2017). ROMs in the sagittal plane are comparable for the hindfoot, forefoot and hallux. The midfoot ROMs of the current model were comparable with Leardini et al.’s midfoot ROMs. Besides the midfoot, Leardini et al.’s model (1999) and Jenkyn and Nicol’s model (2007) reported smaller ROMs in dorsi-/plantar flexion, larger ROMs in internal/external rotation and inversion/eversion for the hindfoot, forefoot and hallux. One possible explanation is that the Leardini’s model and Jenkyn and Nicol’s model described intersegmental foot motions using the joint coordinate system convention from Grood and Suntay (1983), which described joint motions by spins and rotations about
segment-fixed axes and/or around a floating axis created by two segment-fixed axes of the adjacent body segments. In comparison, the current model and the original DuPont foot model used the Cardan angle sequence, which described joint motions by sequential rotations of a segment about body-fixed axes of its proximal segment. The choice of methods may have affected how joint motions were decomposed into the three anatomical planes. In addition, Leardini et al.’s model and Jenkyn and Nicol’s model used clustered markers instead of single, spherical markers to track movements. Clustered markers are heavier and bigger in size, and thus are more likely to wobble and shift during movements and generate greater soft tissue artifacts, leading to an underestimation of dorsi-/plantar flexion.

Rotation angle curves of the current model were also compared with Jenkyn and Nicol’s model (2007, the original foot model (Seo et al., 2014) and Leardini et al.’s model (2007). Although foot segments were defined differently, hindfoot, forefoot and hallux all showed similar movement patterns during the gait cycle.

Segment definitions and rotation angle curves of Jenkyn and Nicol’s model (2007) are shown in Figure 2-10 and Figure 2-11 respectively. Hindfoot dorsi-/plantar flexion and inversion/eversion relative to the tibia showed similar trends with the current model, except that the hindfoot was neutral in the swing phase, while in the current model the hindfoot was slightly inverted. Jenkyn and Nicol’s model (2007) measured midfoot motions the other way around. Figure 2-11 middle A and middle B show hindfoot motions relative to the midfoot. As a result, patterns of the curves were expected to be the opposite or flipped about the x axis compared to the current model. Although the midfoot in the current model stabilized at the neutral position with less of a specific trend in the frontal and transverse planes, the eversion peak and internal rotation peak of midfoot at around TO corresponded with supination (inversion) and slight external rotation of hindfoot in Jenkyn and Nicol’s model (2007). The forefoot supination/pronation (inversion/eversion) demonstrated a similar trend between models. The MLA height/length ratio in both models decreased during the stance phase and rises approaching TO.
Figure 2-11: Intersegmental rotation curves of a gait cycle adapted from “A multi-segment kinematic model of the foot with a novel definition of forefoot motion for use in clinical gait analysis during walking,” by Jenkyn, T. R., & Nicol, A. C., 2007, Journal of Biomechanics, 40(14), 3271-3278. Copyright 2007 by © Elsevier. From the left column to the right, left A: hindfoot dorsi-/plantar flexion relative to tibia, left B: hindfoot inversion/eversion relative to tibia, middle A: hindfoot supination/pronation relative to midfoot, middle B: hindfoot internal/external rotation relative to midfoot, right A: forefoot supination/pronation relative to midfoot, right B: MLA height/length ratio normalized to MLA length.

For the original foot model (Seo et al., 2014) (Figure 2-12), the hindfoot and the hallux showed highly similar trends in all three motions. The MLA height/length ratio also showed a similar pattern (Figure 2-12, third row right, Arch Index). The forefoot motions in the three-segment DuPont model were more in accordance with the forefoot segment in the current four-segment model instead of the midfoot, but it dorsiflexed faster starting at HS and plateaued during midstance. In comparison, the forefoot in the current model showed a gradual increase of dorsiflexion and reached a higher peak at the beginning of pre-swing. This is likely due to the creation of midfoot in the current model. As shown in Figure 2-8, the midfoot quickly dorsiflexed in the early stance phase then gradually plantar flexed until right before TO. The forefoot dorsi-/plantar flexion profile of the original DuPont model was likely showing an add-up effect of the forefoot and midfoot dorsi-/plantar flexion in the current model.
Segment definitions and rotation angle curves of Leardini et al.’s model (1999) are shown in Figure 2-13 and Figure 2-14 respectively. Leardini et al.’s model provided intersegmental rotation angle curves of the stance phase rather than a whole gait cycle. Therefore, only the proportion of 0-61% in the current model were used for comparison. Leardini et al.’s model presented pronation (eversion) as positive and supination (inversion) as negative, so curves of the frontal-plane motions were expected to be the opposite to the current model. Most intersegmental motions were highly similar in trend between models. However, the hindfoot was found to internally rotate during the stance phase in Leardini et al.’s model (1999) but externally rotate in the current model. The forefoot in Leardini et al.’s model (1999) first inverted until midstance then everted until the end of stance phase, while in the current model, it started inverted at HS, became neutral in midstance, then inverted to its peak at TO.
Figure 2-14: Intersegmental rotation angle curves of the stance phase of a gait cycle adapted from “An anatomically based protocol for the description of foot segment kinematics during gait,” by Leardini, A., Benedetti, M. G., Catani, F., Simoncini, L., & Giannini, S., 1999, Clinical biomechanics (Bristol, Avon), 14(8), 528–536. Copyright 1999 by © Elsevier.

2.7.2 Multi-segment foot kinetics

Bruening et al.’s study (2012) and Saraswat et al.’s study (2014) are the only studies that measure multi-segment foot kinetics for three-segment foot models in six degrees of
freedom (DOFs) and calculate internal joint moments using inverse dynamics. Figure 2-15, 2-16, 2-17 show joint moment and power curves from the two studies, respectively.

When compared with the current model, joint moments in the sagittal (dorsi-/plantar flexion) and the transverse planes (internal/external rotation or abduction/adduction) for all foot segments as well as ankle and midfoot joint powers were comparable in terms of direction and magnitude. The ankle inversion/eversion moment was also similar in three models. Differences in patterns are found in the midfoot and hallux. Saraswat et al.’s model (2014) reported a midfoot inversion moment in pre-swing (Figure 2-17, second row left), while in the current model, the midfoot (Chopart joint) and the forefoot (Lisfranc joint) both generated an inversion moment in the early stance phase and an eversion moment in pre-swing, similar to the mid-tarsal inversion/eversion pattern shown in Bruening et al.’s model (2012) (Figure 2-15, center). With regards to the hallux, the current model showed comparable plantar flexion moment and 1st MP power peak in pre-swing to both models. However, large dorsiflexion moment, abduction moment and
energy absorption (power valley) at the 1st MP joint that were absent in Bruening et al.’s (2012) and Saraswat et al.’s model (2014) were generated in the early stance phase.

It is speculated that the methods adopted to measure GRFs may have led to differences in hallux kinetic profiles. Bruening et al.’s study (2012) used two adjacent force plates to partition the foot at the mid-tarsal joint and 1st MP joint and measure GRFs applied to adjacent foot segments separately. Saraswat et al.’s study (2014) combined GRF data with plantar pressure data to identify when and for how long a foot segment was loaded and used the ratio of segmental vertical forces to partition shear forces accordingly. Both methods deliberately disregarded the inertial effects of foot segments during walking since the whole foot has a relatively small inertial effects compared to the whole body. Another method “CPcross” which quantified GRFs only when the center of pressure crossed anterior to the joint is also based on the same assumption (Bruening & Takahashi, 2018; Stefanyshyn & Nigg, 1997). In contrast, the current model uses one force plate without foot partitioning and takes the segmental inertial effects throughout a gait cycle into account. If the foot was partitioned using “CPcross” in the current model, GRFs sequentially crossed the proximal joint of the hindfoot, midfoot, forefoot and hallux at approximately 0%, 12%, 23%, and 41% of a gait cycle, respectively. Figure 2-18 below shows the new multi-segment foot kinetic profiles of the current model using the “CPcross” method. The hallux generated a peak plantar flexion moment of 0.14Nm/kg, a peak adduction moment of 0.03Nm/kg, and the minimal power was -0.59W/kg. The new hallux (1st MP) moment and power curves, in particular, are highly similar in patterns and magnitude compared to Bruening et al.’s (2012) and Saraswat et al.’s (2014) models.
Figure 2-18: Joint moment and power curves of each foot segment in a gait cycle using the “CPcross” method.

2.8 Summary

A novel multi-segment foot model that enables kinematic and kinetic analysis using a motion capture system was proposed in Chapter 2. This foot model divides the foot into four functional segments: the hindfoot, midfoot, forefoot, and hallux. For the kinematic foot model, an X-Z-Y-Cardan angle rotation convention was used to determine intersegmental dorsi/plantar flexion, inversion/eversion, and internal/external rotation. MLA height/length ratio was also measured. For the kinetic foot model, proximal joint moments and joint powers were measured and calculated by inverse dynamics. Masses and inertial parameters of foot segments were reasonably assumed based on anthropometrics. Cortex 7.0 (Motion Analysis Corporation, Santa Rosa, CA, USA) and Microsoft Excel were used to process data. Ten healthy, young adults were tested using this foot model. Test-retest and within-subject CMCs were greater than 0.7 for twenty-
four out of twenty-seven outcome measures, showing strong repeatability of this foot model. This model was also compared with other multi-segment kinematic and kinetic foot models for validation.
Chapter 3

Pre-/Post-HTO patients with medial OA, knee varus

Four patients with medial knee OA and knee varus participated in this study and were successfully tracked using the modified DuPont foot model. Patients’ demographics are presented in Table 3-1. Three patients (patient 1, 2, 3, labeled as post-HTO 1, 2, 3 respectively in figures) received HTO so that their knee varus was corrected by the time of testing. One patient (patient 4, labeled as pre-HTO in figures) had not yet received an HTO by the time of testing and was tested with varus knee alignment. Data from the ten healthy, young adults discussed in Chapter 2 served as the control group. Given the small sample size of the patient group, individual curves of the patients are presented with the control’s average curves with standard deviations in the figures.

<table>
<thead>
<tr>
<th>Patient</th>
<th>Age</th>
<th>Sex</th>
<th>Height (cm)</th>
<th>Weight (kg)</th>
<th>BMI (kg/m²)</th>
<th>Affected limb</th>
<th>Pre-op varus angle (degree)</th>
<th>Pre-op KL score</th>
<th>Post-op varus angle (degree)</th>
<th>Post-op KL score</th>
<th>Time post-HTO (month)</th>
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<td>L</td>
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<td>2</td>
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<td>122.1</td>
<td>31.9</td>
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<td>3.5</td>
<td>3</td>
<td>108</td>
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<td>Male</td>
<td>183.0</td>
<td>119.0</td>
<td>35.5</td>
<td>L</td>
<td>8</td>
<td>3</td>
<td>4.2</td>
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</tr>
<tr>
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<td>Male</td>
<td>180.0</td>
<td>107.6</td>
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<td>R</td>
<td>6.1</td>
<td>2</td>
<td>NA</td>
<td>3</td>
<td>60</td>
</tr>
</tbody>
</table>

3.1 Kinematic analysis

The average walking speed of the four patients was 1.21±0.14 m/s and the average stride length was 1.38±0.14 m. As can be seen in Figure 3-1, all four patients demonstrated similar hindfoot and forefoot dorsi-/plantar flexion patterns in the gait cycle. Hallux dorsi-/plantar flexion of patients had a similar range of motion (ROM), but the dorsiflexion peak appeared later in the gait cycle. All post-HTO patients showed a lower MLA height/length ratio throughout the gait cycle.
3.1.1 Post-HTO patients

Besides the characteristics in common, the three post-HTO patients each demonstrated an individual movement pattern during walking (Figure 3-1).

As shown in Figure 1, Patient 1’s (red curve in Figure 3-1) hindfoot showed a greater internal rotation starting at toe off (TO) and lacked an inversion peak at TO (around 61%). The midfoot, although had a smaller ROM in the sagittal plane, showed an inversion peak at TO (around 61%) that was not present in the control group. The forefoot was more externally rotated throughout the gait cycle and had a larger ROM in the frontal plane with a higher inversion peak right after TO (around 64%). The hallux of
patient 1 was in a more abducted position (stabilized around 20° of abduction) and showed an opposite movement pattern compared to the control group and other patients.

For Patient 2 (yellow curve in Figure 3-1), the hindfoot and midfoot in the transverse and frontal planes demonstrated similar movement patterns to the control group. However, hindfoot was internally rotated (stabilized around 3° of internal rotation) and everted (stabilized around 12° of eversion) throughout the gait cycle. The midfoot was internally rotated (stabilized around 4° of internal rotation) and inverted (stabilized around 10° of inversion) throughout the gait cycle, and it was in dorsiflexion during the stance phase. The forefoot was more plantar flexed with a smaller ROM in the sagittal and more everted in the frontal plane (stabilized around 3° of eversion) throughout the gait cycle. The forefoot also lacked an internal rotation peak before TO. The hallux abduction/adduction was similar to the control group.

Patient 3 (light blue curve in Figure 3-1) demonstrated similar movement patterns to the control group during the stance phase but had greater ROMs for all foot segments in all three planes during the swing phase. The midfoot in the frontal plane continuously everted during the stance phase. The hallux had an obviously higher abduction peak at TO.

### 3.1.2 Pre-HTO patient

For the pre-HTO patient (dark blue curve in Figure 3-1), the hindfoot had a greater internal rotation after TO. The midfoot had a smaller ROM in the sagittal plane without a planter flexion peak in pre-swing and was continuously evert ing during the stance phase. The forefoot was more everted throughout the gait cycle and lacked an internal rotation peak in pre-swing. The hallux had a larger ROM in abduction/adduction and a higher abduction peak at TO. The MLA height/length ratio was rising instead of dropping during the stance phase.

### 3.2 Kinetic analysis

Plantar/dorsiflexion moments in the sagittal plane for all foot segments were similar to the control group (Figure 3-2). Most deviations of joint moments were found in the
transverse and frontal planes. Since the dorsi-/plantar flexion moment was significantly greater than the inversion/eversion moment and the internal/external rotation moment for all foot segments, it contributed most to the power generation of the proximal joint. Proximal joint powers for all foot segments reached their peaks in the beginning of preswing, while the 1st MP joint power had another peak in the early phase of a gait cycle.

![Figure 3-2: Joint moment curves and joint power curves of each foot segment in a gait cycle for patients and the control group.](image)

### 3.2.1 Post-HTO patients

Patient 1 (red curve in Figure 3-2) had a smaller hindfoot external rotation moment, a greater hindfoot inversion moment, a greater midfoot internal rotation moment, a greater midfoot inversion moment and a greater forefoot inversion moment compared to the control group. This patient did not have a peak hallux abduction moment in early midstance. The ankle power and the Chopart joint power were smaller than the control group. The 1st MP joint had a lower valley than the control.
Patient 2 (yellow curve in Figure 3-2) showed a greater hindfoot internal rotation moment, a smaller midfoot external rotation moment, a greater midfoot inversion moment, a smaller forefoot external rotation moment, a greater forefoot inversion moment and smaller hallux abduction moment. The Chopart joint power was greater and appeared later in the gait cycle.

Patient 3 (light blue curve in Figure 3-2) had greater plantar flexion moments for all foot segments, a greater hindfoot eversion moment, a greater midfoot external rotation moment, a greater midfoot eversion moment, a greater forefoot external rotation moment and the highest hallux abduction moment. The Lisfranc joint power was greater than the control group.

3.2.2 Pre-HTO patient

The pre-HTO patient (dark blue curve in Figure 3-2) had a greater hindfoot internal rotation moment, a greater hindfoot eversion moment, a greater midfoot external rotation moment, a greater midfoot inversion moment and a greater forefoot inversion moment. The ankle power and the Lisfranc power were greater than the control, while the Chopart joint power was lower. This patient also had a significantly lower valley of 1st MP joint power in early midstance. All joint powers appeared later in the gait cycle than the control group.

3.3 Discussion

Pre-/post-HTO patients with medial OA and knee varus were tested using the multi-segment kinematic and kinetic foot model proposed in Chapter 2. The average curves for kinematic and kinetic measures were not calculated for the patient group because no generalizable conclusion could be made based on such a small sample size.

Although pre-/post-HTO patients with medial OA and knee varus showed variability and individuality in their inter-segmental foot kinematics and kinetics, some features were shared. All patients showed greater hindfoot internal rotation after toe-off (TO) and the three post-HTO patients walked flatfooted with lower MLA. The increasing MLA height/length ratio of the pre-HTO patient can be explained by the midfoot movements.
during the stance phase. This patient’s midfoot was more internally rotated than the control group and was continuously everting during the stance phase, meaning that the NV (navicular) marker moved medially from the vector between BC (bottom calcaneus) and 1MTH (first metatarsal head), which defined the MLA length. As a result, the height of MLA, calculated as the perpendicular distance between NV and the MLA length, may have increased during the stance phase, and thus leading to an increase in MLA height/length ratio. In fact, this pre-HTO patient still walked with a collapsed MLA. Additionally, all patients walked with increased hallux abduction (abducted hallux and/or increased abduction/adduction ROM). The increased hallux abduction was also seen in flatfooted individuals due to the increased activation of abductor hallucis (Gray & Basmajian, 1968).

Besides the common features, each patient showed abnormal foot/ankle motions at various locations. Post-HTO patient 1 was suspected to have foot deformations because abnormal movement patterns were found in the hindfoot and midfoot (lack of hindfoot inversion at TO, abrupt midfoot inversion at TO), and the forefoot demonstrated greater flexibility in inversion/eversion. Post-HTO patient 2 demonstrated an obvious malalignment of foot segments during walking because this patient’s foot segments stabilized far away from the neutral position determined by static standing (everted/internally rotated hindfoot, inverted/internally rotated midfoot, and everted forefoot). The everted, internally rotated hindfoot and internally rotated midfoot were also found in flatfooted individuals and medial knee OA patients with knee varus (Levinger et al., 2010; Levinger et al., 2012; Saraswat et al., 2014). Post-HTO patient 3 showed close-to-normal movement patterns during the stance phase but had significantly greater ROMs for all foot segments in all three planes during the swing phase. Patient 3’s gait may explain this phenomenon. From observation, the patient’s foot showed greater flexibility with visible sequential contact of foot segments with the ground during walking. It is worth noting that the three post-HTO patients still held a small varus angle (2°-4°) with knee OA progression in their affected limbs nine to ten years after the surgery. This could explain why post-HTO patients did not demonstrate close-to-normal foot motions in this study. Interestingly, the pre-HTO patient who had the varus knee alignment preserved did not show the everted hindfoot or increased hindfoot eversion as
those in previous studies (Barrios et al., 2009; Levinger et al., 2012; Arnold et al., 2014). Instead, compensating movements happened mostly in the midfoot and forefoot. Both foot segments showed eversion and the forefoot was externally rotated in pre-swing. The pre-HTO patient also had a more rigid midfoot in the sagittal plane without any reduction in ROM in the frontal plane as well as normal hallux dorsi-/plantar flexion, not in accordance with Arnold et al.’s study (2014) on medial OA patients with knee varus which has found increased foot rigidity at the midfoot in the frontal plane and at the hallux in the sagittal plane. The increased foot rigidity was also not seen in post-HTO patients.

With regards to multi-segment foot kinetics, patients showed higher peak plantar flexion moments for all foot segments. Most variations of joint moments happened in the hindfoot and midfoot in the frontal and transverse planes (inversion/eversion, internal/external rotation). Patients either showed significantly higher moment peaks than the control group or joint moments in the opposite direction. For example, the pre-HTO patient’s hindfoot (ankle) was under an internal rotation moment and an eversion moment, and the midfoot (Chopart joint) was under a higher external rotation moment and an inversion moment. The combination of these joint moments indicated a tendency of twisting between the hindfoot and the midfoot. Higher peak powers occurred at the ankle, the Lisfranc joint and the 1st MP joint and the peaks appeared later in the gait cycle than the control group, meaning that these joints were under higher and faster loadings during walking. Higher joint moments (higher forces) and/or higher velocity (e.g., larger transverse-plane ROM of the hindfoot in a normalized gait cycle) of these foot segments all contributed to the higher energy generation at the joint.

More details of foot/ankle motions could be revealed when kinematic and kinetic measures are combined for analysis. When the joint moment is in the opposite direction of the joint movement, the muscles and ligaments acting across the joint are eccentrically contracting and being stretched, respectively. Taking the pre-HTO patient as an example, the midfoot was everting throughout the stance phase under an inversion moment, meaning that muscles and ligaments acting on the midfoot at the Chopart joint (e.g., tibialis posterior, dorsal talonavicular ligament, bifurcate ligaments) were eccentrically
contracting and being stretched. This resulted in energy absorption and a smaller Chopart joint power of the pre-HTO patient as shown in Figure 3-2.

Based on the current data, all patients with medial knee OA walked with flat feet and other abnormal foot/ankle movements and loadings were found in different foot segments in all three planes, regardless of their varus knee alignment being corrected or not. The pre-HTO patient who had the varus knee alignment preserved did show greater forefoot eversion, but hindfoot eversion was not the compensatory strategy adopted as previous studies suggest. Given that only four patients participated in the study, and they were not followed from pre-HTO to post-HTO, there was not enough data to support any firm conclusion or detect any change occurred before and after the surgical correction of knee varus. The preliminary findings of HTO patients support the need for a validated multi-segment foot model to quantify pathology in clinical populations with lower limb alignment issues or musculoskeletal conditions in general.
Chapter 4

4.1 Summary

The primary objective of this study was to develop a clinically friendly multi-segment foot model that enables both kinematic and kinetic analysis. Inspired by the definition of a separate midfoot segment in Jenkyn and Nicol’s model (2007) and using the three-segment DuPont foot model as the base, the current model divides the foot into four segments: hindfoot (calcaneus), midfoot (cuneiforms, navicular, cuboid), forefoot (five metatarsals), and hallux. Intersegmental movements are measured relative to the proximal foot segment and described in all three anatomical planes (dorsi-/plantar flexion in the sagittal plane, inversion/eversion in the frontal plane, internal/external rotation in the transverse plane). The intersegmental rotation angles are calculated using the X-Z-Y Cardan angle sequence. The innovative part is the addition of multi-segment foot kinetic analysis. The whole foot’s mass is partitioned in proportion to the foot segment volume ratio to enable the calculation of joint moments and joint powers between foot segments by inverse dynamics. Using an optical motion capture system with a force plate in the laboratory, the intersegmental rotation angle, the joint moment, and the joint power were successfully measured and calculated for the following motions during a gait cycle:

- Hindfoot dorsi/plantarflexion
- Hindfoot inversion/eversion
- Hindfoot internal/external rotation
- Midfoot inversion/eversion with respect to the hindfoot
- Midfoot internal/external rotation with respect to the hindfoot
- Midfoot dorsi/plantarflexion with respect to the hindfoot
- Forefoot inversion/eversion with respect to the midfoot
- Forefoot internal/external rotation with respect to the midfoot
- Forefoot dorsi/plantarflexion with respect to the midfoot
- Hallux dorsi/plantarflexion with respect to the forefoot
- Hallux ab/adduction with respect to the forefoot
- Rise and fall of the medial longitudinal arch (MLA)

The second objective was to test the reliability and validity of this novel multi-segment foot model. As hypothesized, test-retest CMCs and within-subject CMCs for twenty-four
out of twenty-seven outcome measures were greater than 0.7 (0.3 ≤ R < 0.7 for rotation angle curves of hindfoot inversion/eversion, midfoot internal/external rotation, midfoot inversion/eversion), showing overall strong reliability and repeatability. Kinematic measures of this foot model were compared with Jenkyn and Nicol’s model (2007), the original DuPont model and Leardini et al.’s model (1999). Kinetic measures were compared with data from three-segment foot models used in Bruening et al.’s study (2012) and Saraswat et al.’s study (2014). Outcome measures are validated by data in previous studies. Multi-segment foot kinetic data with a separate midfoot segment are novel data presented to the field.

The third objective of this study was to apply this multi-segment foot model to pre-/post- HTO patients with medial knee OA and knee varus and compare their foot/ankle motions with healthy controls. Four patients participated in the study, among which three patients received HTO and one patient did not receive HTO. It was hypothesized that pre-HTO patients will show greater forefoot and hindfoot eversion and lower MLA, while post-HTO patients will show improved or close-to-normal foot/ankle movements. The pre-HTO patient did show greater forefoot eversion but no obvious eversion in the hindfoot. Opposed to the hypothesis, however, post-HTO patients did not show close-to-normal gait but walked flatfooted with increased hallux abduction. Other abnormal motions and loadings were detected in different foot segments in all three planes regardless of the presence of varus knee alignment or not, suggesting individual compensating strategies. Although no firm conclusion can be drawn from the small patient group, the current pilot study suggests that the multi-segment kinematic and kinetic foot model is sensitive enough to detect abnormal foot/ankle motions in clinical populations.

This multi-segment foot model is considered more clinically friendly and advantageous over the other foot models. First, this model only requires eleven small, single markers on the foot to define four functionally important foot segments including a separate midfoot. It is simple and reliable to apply to clinical populations and does not require any gait-altering testing protocol that may interfere with the collection of real data. Second, the model supports multi-segment foot kinetic analysis, which can provide more information on loadings within the foot. Third, Cortex is a user-friendly motion capture and
processing software. It supports model construction for repetitive use on new subjects, automatically identifies gait events, and calculates intersegmental rotation angles as well as joint moments and joint powers when combining GRF data with kinematic measures.

4.2 Limitations

The multi-segment kinematic and kinetic foot model proposed in this study has a few limitations.

First, masses and inertia parameters for the foot segments constructed were reasonably assumed based on anthropometric data from De Leva’s study (1996) and Drillis et al.’s study (1964) rather than experimentally determined. Masses of foot segments were partitioned in proportion to the segmental volume ratio. The center of mass (CM) for each foot segment was arbitrarily set as 0.5 of the segment length and the radius of gyrations (RG) for all segments was set as the same as the whole foot. In fact, the hindfoot is more compacted and consists mostly of bones (e.g., calcaneus and talus) while the midfoot/forefoot contains more soft tissues with more spread-out structures, leading to variations in the inertia parameters. Therefore, the multi-segment foot kinetic model holds potential systematic errors. However, such a modeling assumption can be justified as GRFs applied to the foot are much larger in comparison to the small masses of foot segments, making the impacts of foot inertial parameters on internal joint moment calculations much smaller. In addition, soft tissue artifacts associated with the use of skin markers are inevitable despite that the foot is bony with fewer fact deposits compared to other parts of the body. We used single, spherical markers instead of larger clustered markers to minimize relative movements between markers and the skin, but relative movements between markers became a new limitation.

Second, we used only one force plate without foot partitioning when measuring GRFs in this study. As a result, GRFs could be oversimplified when more than one segment is in contact with the ground (e.g., during midstance). Subarea GRFs, shear forces in particular, applied to different foot segments cannot be detected due to limited spatial and temporal resolutions of the force plate. Moreover, the direction of joint moments could be influenced by the relative position between the overall center of pressure on the force
plate and the CMs of segments. However, the current model quantified the inertial effects of foot segments that have been neglected in previous multi-segment foot kinetic analysis.

Finally, although this multi-segment foot model is considered more clinically friendly than the other foot models, its implementation still requires a gait laboratory with motion capture equipment.

4.3 Implications

As mentioned in the limitation section 4.2, masses and inertia parameters of foot segments were reasonably assumed based on anthropometric data. Future studies should be done to experimentally determine masses and inertia parameters including CMs and RGs in cadavers or using gamma-ray imaging as Zatsiorsky and Seluyanov (1983) has done on human volunteers.

Kyung et al.’s study (2021) (n=24) used the original three-segment DuPont model and found that post-HTO patients showed close-to-normal kinematics of all foot segment in all three planes, especially a reduction in compensatory midfoot eversion (forefoot eversion relative to hindfoot) compared to their pre-HTO gait. However, in the current study, abnormal motions (e.g., lower MLA, larger ROM in hindfoot internal/external rotation around TO, variations in inversion/eversion moment and internal/external rotation moment on hindfoot and midfoot) were still present in post-HTO patients. One explanation to the difference of results is that Kyung et al.’s study (2021) tested patients 12 months post-HTO while we tested patients nine to ten years post-HTO. Post-HTO patients in the current study all had knee OA progression (indicated by increased KL scores) and regained a small varus knee angle (2-4°), potentially leading to abnormal foot/ankle motions during walking. However, it is unknown if more subtle, abnormal foot/ankle motions that were once undetected are now revealed due to the separate midfoot segment and the addition of multi-segment foot kinetic analysis. As a result, more research needs to be done on patients with medial OA, knee varus with a larger sample size and compare with age, sex, BMI matched control groups. Longitudinal studies that follow up patients from pre-HTO to post-HTO using the multi-segment foot
model developed in this study would provide more information on which parts of the foot demonstrate compensating movement patterns for knee varus and how HTO changes the foot/ankle compensation.

Moreover, knee adduction moment is identified as a characteristic surrogate measure of knee osteoarthritis that reflects knee alignment and knee loading distribution and associates with pain (Birmingham et al., 2007; Cho et al., 2015; Yamagata et al., 2021). There are also plenty of studies investigating the effects of foot progression angles and lateral wedge insoles on reducing knee adduction moment in healthy individuals and/or patients with OA (Cho et al., 2015; Tokunaga et al., 2016; Ulrich et al., 2021; Wang et al., 2021). However, motions and loadings within the foot are often not directly measured in these scenarios. Future work should be done using this multi-segment foot model to explore if there is any association between foot measures and knee adduction moment and how the relationships modulate under different conditions in both healthy individuals and patient with knee OA, for example, from pre-HTO to post-HTO as knee alignment changes, with different foot progression angles, wearing lateral wedge insoles or not, etc.

Additionally, since inverse dynamics only calculates net joint moments and the summed effects of internal forces on joints, muscle activation would remain unknown. Mills et al.’s study (2013) shown that knee OA patients have increased, prolonged co-contraction and activity of muscles surrounding the knee during level walking. Hunt and Smith’s study (2004) has also reported different muscle activity patterns in normal and flatfooted individuals during walking. EMG can be a powerful addition to gait analysis for the detection of potential muscle co-contractions in OA patients with knee varus.

### 4.4 Conclusion

In conclusion, this thesis study proposed a clinically useful multi-segment foot model that allows kinematic and kinetic gait analysis. This foot model divides the foot into four segments: the hindfoot, midfoot, forefoot, hallux. Intervertebral rotation angles (dorsi-/plantar flexion, inversion/eversion, and internal/external rotation), joint moments, joint powers and MLA height/length ratio were measured in an optical motion capture laboratory setting. This foot model showed strong test-retest and within-subject reliability
(R>0.7) for most joint motions (24/27) in healthy adults and was sensitive enough to detect abnormal foot motions including lower MLA, increased hallux abduction in pre-/post-HTO patients with medial knee OA and knee varus. This novel model can be a clinically useful tool for research and assessment on clinical populations as it models all four functionally important foot segments with an easy-to-apply marker set. It is also warranted because it measures both intersegmental foot kinematics and kinetics that are unable to be revealed by other multi-segment foot models. The necessity of these foot measures is further confirmed when patients’ data were compared with the healthy control group and showing that motion deviations can potentially happen in all four foot segments.
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Appendices

Appendix A: Linear interpolation of gait data in Excel (normalize a gait cycle to 100% in terms of time) using the FORECAST function.

Appendix B: MLA height/length ratio calculation in Excel.

Appendix C: Permissions to use materials from previously published articles

Table 1-1: Adjusted mass percentages, CMs and RGs for body segments.

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Figure 1-1: Anatomical movements (right).

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Figure 0-1: Arches of the foot. Red: LLA. Green: MLA. Blue: transverse arch.

May 6, 2022
Simpin Zhu

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Figure 1-7: The mechanical axis (LBA) and HKA in knee varus and in knee valgus. https://s100.copyright.com//CustomerAdmin/PLF.jsp?ref=76b7c46c-32e5-4d51-a9a9-3bb0aa0fd2ba


Figure 2-11: Intersegmental rotation curves of a gait cycle of Jenkyn and Nicol’s model (2007). From the left column to the right, left A: hindfoot dorsi-/plantar flexion relative to tibia, left B: hindfoot inversion/eversion relative to tibia, middle A: hindfoot supination/pronation relative to midfoot, middle B: hindfoot internal/external rotation relative to midfoot, right A: forefoot supination/pronation relative to midfoot, right B: MLA height/length ratio normalized to MLA length. https://s100.copyright.com//CustomerAdmin/PLF.jsp?ref=91b4c1e7-300a-4451-b0d3-c41c55c36a0e

Figure 2-13: Segment definitions of Leardini et al.’s multi-segment foot model (1999). https://marketplace.copyright.com/rs-ui-web/mp/license/528ac04f-005d-4394-ad95-e593f67f98e5/8d502788-3a1a-4b9a-848c-9a719851111c
Figure 2-14: Intersegmental rotation angle curves of the stance phase of a gait cycle of Leardini et al.’s model (1999).
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Figure 2-15: Joint moments during the stance phase of a gait cycle from a three-segment foot model.
Figure 2-16: Joint powers during the stance phase of a gait cycle from a three-segment foot model.
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Figure 2-17: Joint moments and powers in a gait cycle from a three-segment kinetic foot model (mSHCG foot model).
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