Validation of the multi-segment foot model with bi-planar fluoroscopy

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

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VALIDATION OF THE MULTI-SEGMENT FOOT MODEL WITH BI-PLANAR FLUOROSCOPY

(Thesis format: Integrated Article)

by

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Department of Kinesiology
Graduate Program in Biomechanics

Submitted in partial fulfillment
of the requirements for the degree of
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1 ABSTRACT

A multi-segment foot model (MSFM) is a useful tool for measuring foot joint kinematics although soft-tissue artefact is often present. Quantifying this error is needed to evaluate the accuracy of this model. This study validated the MSFM against bi-planar radiostereometric analysis (RSA) fluoroscopy. Heel-strike, mid-stance, and toe-off events during the stance phase were compared between motion capture and fluoroscopy. Rise/drop of the medial longitudinal arch showed a significant difference ($p < 0.05$) during toe-off, but no significant difference during heel-strike or mid-stance. Hindfoot supination/pronation and internal/external rotation, and forefoot supination/pronation motions showed no significant difference between the two techniques. The lack of significant difference will allow the MSFM to be used as a sufficiently accurate technique for measuring foot joint motions.

**Keywords:** multi-segment foot model, soft-tissue artefact, bi-planar fluoroscopy, RSA, validation
This work would not have been completed without the help of other people below, whose help is greatly acknowledged by the author.

**Chapter 2:** Megan Balsdon designed the study and collected the data.

**Chapter 3:** Lisa Oikawa was a big help by collecting data.
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I can’t forget to thank Megan Balsdon for teaching me all there is to know about how to use the fluoroscopes, calibrate them, process the data, making bone models, matching, and everything about the fluoroscopy part of this project.

A big thanks to Andrew Dragunas for lending me his video capture equipment so that I would be able to collect the video feed coming from the fluoroscopes.

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Thanks to my little brother with great artistic skills, much better than mine, for drawing me a nice foot for the Introduction chapter.
Thanks to my parents for their support and calmness towards graduate school. I have to say, rarely do I receive hand written letters in the mail, and usually they only come from my dad, so here is one that I received from him at some point during my two years in London. He said it hung in his kitchen during his graduate studies, so then it hung on my fridge during mine.
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8 LIST OF ABBREVIATIONS, SYMBOLS, AND NOMENCLATURE

° - degree(s)
%
- percent
2D – two dimensional
3D – three dimensional
3D pose – 3D position and orientation
CAER – calcaneal eminentia retrotrochlearis
CALT – calcaneal lateral tuberosity
CAMT – calcaneal medial tuberosity
Cavus – (pes cavus) participants with high arches
CCD – charge couple device
CMOS – complementary metal oxide semiconductor
CT – computed tomography
d – distance in mm from the fluoroscope source to the image intensifier
DH – distal head of the hallux
fps – frames per second
G_DS_ST – generic different size, same type foot model
G_SS_DT1 – generic same size, different type foot 1 model
G_SS_DT2 – generic same size, different type foot 2 model
G_SS_ST – generic same size, same type foot model
Hz – hertz
ISB – International Society of Biomechanics
JCS – joint coordinate system
LED – light emitting diode
LPH – lateral head of interphalangeal
LLM – lateral malleolus
LMM – medial malleolus
mm – millimeter(s)
mA – milliampere(s)
MCI – first cuneiform
MCU – cuboid
MIB – base of fifth metatarsal
MIH – head of first metatarsal
min - minute
MLA – medial longitudinal arch
MNT – navicular tuberosity
MPH – medial head of interphalangeal
MRI – magnetic resonance imaging
MSFM – multi-segment foot model
ms – millisecond
MVB – base of fifth metatarsal
MVH – head of fifth metatarsal
planus – (pes planus) participants with flat feet or low arches
RSA – radiostereometric analysis
s – second(s)
SS – subject specific model
STA – soft tissue artefact
STC – Standardization and Terminology Committee
STH – talar head
Tiff – tagged image file format
TKA – total knee arthroplasty
μR – micro roentgen
μSv – micro Sievert
WOQIL – Wolf Orthopaedic Quantitative Imaging Laboratory
x-direction – x direction within a coordinate system
y-direction – y direction within a coordinate system
z-direction – z direction within a coordinate system
1 CHAPTER 1 – INTRODUCTION

1.1 Foot Anatomy

1.1.1 Bones of the Foot

The foot has the task of giving the body a stable, and efficient interface between the body and the ground for locomotion. During the gait cycle, the foot has to go from a rigid lever to allow for push off, to a flexible structure that will allow the foot to adapt to the ground by absorbing and transmitting forces while keeping whole body stability (Nordin & Frankel, 2012). The bones, ligaments, tendons, and fascia form joints in the foot that allow for its vast mobility. The human foot has 26 bones plus 2 sesamoid bones for a total of 28 bones (Abrahams, 2007). As seen in Figure 1-1, the bones of the hindfoot and midfoot are the talus, calcaneus, navicular, cuboid, and the three cuneiforms. The forefoot contains the 5 metatarsal bones and the phalanges. All 5 digits are formed by a distal and proximal phalange (Nordin & Frankel, 2012).
There are 6 important joints in the foot that allow for movement to occur. The ankle joint is formed of the talus and distal parts of the fibula and tibia; it has three dimensional motion and 6 degrees of freedom. The subtalar joint is formed of the calcaneus and the talus bone. This joint allows for translations of motion between the tibia and the foot. It is a hinge joint with an oblique axis, which allows for inversion-eversion and abduction-adduction motions of the hindfoot. The transverse tarsal joint is made of two joints, the talonavicular joint and the calcaneocuboid joint. The talonavicular joint is formed of the talar head and the posterior surface of the navicular. The calcaneonavicular joint is a saddle joint with not very much motion compared to the talonavicular joint. The transverse tarsal joint moves as a whole and contributes to pronation and supination of the foot. The distal intertarsal joints are between the navicular and cuneiform bones, between the cuboid and lateral cuneiform, and between the three cuneiform bones. These joints have a few degrees of motion and contribute to pronation-supination of the foot.
The tarsometatarsal joints are located between the tarsal bones and the metatarsal bones. The intermetatarsal joints are between the metatarsals themselves. Their mobility varies depending on which toe is concerned; the hallux has the most mobility, followed by the 3rd, 4th, and 5th toes. The 2nd toe has limited mobility since it is wedged in between the cuneiforms and 1st metatarsal base. The metatarsophalangeal joints are the joints between the metatarsal bones and the phalanges. This joint has its primary motion in the sagittal plane. Finally, the interphalangeal joints are hinge joints between the phalanges. Their motion is mostly flexion (Oatis, 2009).

1.2 Multi-Segment Foot Model

Many multi-segment foot models (MSFM) have been developed to quantify three-dimensional (3D) motions of the joints of the foot. Several models track three foot segments, the hindfoot (calcaneus and talus), forefoot (metatarsals), and hallux (phalanges) (Bruening, Cooney, & Buczek, 2012; Carson, Harrington, Thompson, O'Connor, & Theologis, 2001). Other models incorporate the tibia and fibula as well as the hindfoot and forefoot (Kidder, Abuzzahab, Harris, & Johnson, 1996; Leardinin, Benedetti, Catani, Simoncini, & Giannini, 1999; Rattanaprasert, Smith, Sullivan, & Gilleard, 1999).

Jenkyn and Nicol (2007) developed the MSFM used in this thesis. The model is used for tracking four segments of the foot. The first segment is the hindfoot and is formed by the calcaneus. The second segment is the midfoot and is formed by the tarsal bones including the three cuneiforms, navicular, and cuboid. The third segment is the medial forefoot consisting of the first and second metatarsals. Finally, the fourth segment is the lateral forefoot consisting of the third, fourth, and fifth metatarsal bones. Figure 1-2
represents the different segments of the MSFM as well as the three bony landmarks per segment, which form the segment fixed axis systems (Jenkyn & Nicol, 2007).

Figure 1-2: Rigid segments of the multi-segment foot model defined as the hindfoot in dashed grey, midfoot in stripped grey, medial forefoot in solid grey, and lateral forefoot in tethered grey. The three bony landmarks per segment, which form the segment-fixed axis systems, are explained in Table 1-1. (Jenkyn & Nicol, 2007)

There are six foot joint motions defined by Jenkyn and Nicol’s (2007) MSFM, 1) ankle joint, 2) subtalar joint, 3) hindfoot segment motion with respect to the midfoot in the frontal plane, 4) hindfoot segment motion with respect to the midfoot in the transverse plane, 5) forefoot segment motion, and 6) the height-to-length ratio of the medial longitudinal arch (MLA). These motions are visually depicted in Figure 1-3.
The ankle joint motion, also called talocrural motion, is defined as the rotation of the talus with respect to the lower leg segment about the vector 2-axis of the ankle joint coordinate system (JCS). A positive rotation about the vector 2-axis is representative of dorsiflexion as represented in part A of Figure 1-3. The subtalar joint, also called talocalcaneonavicular joint, motion was defined as midfoot segment rotation with respect to the talus about the vector 2-axis of the Subtalar-JCS. A positive rotation about the vector 2-axis is representative of inversion and a negative rotation is eversion of the midfoot segment as represented in part B of Figure 1-3 (Jenkyn & Nicol, 2007). These joint motions were defined initially by the Standardization and Terminology Committee (STC) of the International Society of Biomechanics (ISB). The STC defined the ankle joint as the articulation between the talus and the tibia/fibula. The subtalar joint was defined as the articulation between the talus and the calcaneus. From those joint motions, they defined a JCS for the ankle and subtalar joints (Wu et al., 2002). Hindfoot motion is presented in part D of Figure 1-3; the movement of the hindfoot with respect to the midfoot segment is defined as supination/pronation about the midfoot vector 3-axis and internal/external rotation about the midfoot vector 1-axis. These motions are described in the method of Grood and Suntay (1983) (Grood & Suntay, 1983). Moving on to the forefoot, which is presented in part C of Figure 1-3, the compound twisting of the lateral and medial forefoot segments with respect to the midfoot segment defined the fifth joint motion of the MSFM. This angle is created between the vector 2-axis of the midfoot and the vector joining the heads of the first and fifth metatarsals projected onto the midfoot vector 1- and 2-axis. An increasing angle represented supination of the forefoot (Jenkyn & Nicol, 2007).
Figure 1-3: Joint motions of the MSFM. A) Ankle joint motion defined as the rotation of the talus with respect to the lower leg segment. B) Subtalar joint motion defined by the midfoot segment rotation with respect to the talus bone. C) Compound twisting of the medial and lateral forefoot segments with respect to the midfoot segment. D) Hindfoot segment motion with respect to the midfoot. E) Shape of the medial longitudinal arch described as the height-to-length ratio. (Jenkyn & Nicol, 2007)
1.3 Kinematic Measurement Techniques

There are several techniques that have been developed to measure kinematics of gait and human movement. Kinematics is the study of motion of the limbs and joints of the body irrespective of forces. Movement is described in terms of displacement, velocity, and acceleration. Displacement is the distance travelled by an object between two locations as, for example, the displacement of the knee during walking, which goes from 10° at heel strike to 70° of flexion at toe off, thus creating 60° of angular displacement. The change in position, or displacement over time is called velocity. Change in linear or angular velocity over time is acceleration. Most of gait analysis is based on displacement information. Many factors can affect walking and running patterns such as walking speed, age, height, weight, strength and flexibility, and aerobic condition (Oatis, 2009). Kinematic analysis techniques range from goniometers, film cameras, stereophotogrammetry, medical imaging, and fluoroscopy.

1.3.1 Goniometry

A simple and basic way to measure joint kinematics is using a goniometer. Goniometry allows one to measure the range of motion of a joint. There are several different types of goniometers as described by Goodwin et al. (1992). Universal goniometers are easy to use, but restricted mostly to simple joint movements or static joint positions. Fluid goniometers are made of a circular clear tube filled with liquid. As the device is rotated the fluid moves relative to the graduated disk and makes an angle equal to the angular displacement of the base. This type of goniometer works independently of the center of rotation. Another type of commonly used goniometer is an electrogoniometer. This type
contains a strain gauge steel strip placed between two plastic sections. The angular displacement of the joint is displayed digitally on the display unit. The beginning of the movement is set to zero and the end of the motion will be displayed as the angle of the movement (Goodwin, Clark, Deakes, Burdon, & Lawrence, 1992).

1.3.2 Cinefilm

Another technique for kinematic analysis, which has been widely used in research, requires the use of cameras, cinefilm, and high-speed cameras. High-speed cameras allow for assessment of activities with velocities and accelerations greater than walking. These types of cameras do not require any wires and cables attached to the subject, thus their range of motion is greatly improved and movement is not obstructed. The downside with this type of motion capture is that each frame needs to be digitized separately, which requires a great amount of time and effort (Schneck & Bronzino, 2002).

1.3.3 Stereophotogrammetry

Stereophotogrammetric systems such as ELITE and VICON are commonly used for kinematic analysis. These systems use two or more cameras placed in different locations covering a specific capture volume. The subject wears reflective markers placed on specific body landmarks. Each marker has to be seen by two cameras in order for its location to be collected by the system (Leardinin et al., 1999). Some cameras have LED rings around the camera lens. These LEDs act like a strobe light and reflect off the markers. Infrared lights have become commonly used today for optical motion capture cameras (Roesler, 2011). The infrared light bounces off the markers covered in retro-reflective tape and returns to the camera. This type of marker, covered with retro-
reflective tape, is called a passive marker. Another marker system commonly used in kinematic analysis requires active markers. Active markers are small LED makers that are placed on the subjects’ bony landmarks. They produce light that is captured by the camera’s lens. There are advantages to both types of motion capture systems. Active markers allow for the location of bony landmarks to be known immediately because the markers are each fired sequentially and therefore the system can immediately determine the location of each marker. The main disadvantage of active markers is that they require a system of cables to power the markers. These cables could be intrusive for movement patterns. As for passive markers, an advantage is that they simply go on the body with double-sided tape and are not as intrusive to movement. The disadvantage lies in the post-processing phase, as the researcher is required to identify the markers after testing, although algorithms have been developed to make this process faster and automatic (Schneck & Bronzino, 2002).

### 1.3.4 Markers

Markers are placed on specific bony landmarks either as single units or as a cluster of connected markers. The 3D coordinates of the markers in the laboratory frame of reference are the output of data acquisition using the video camera based systems. Each body segment requires three markers or reference points in order for a body-fixed coordinate system to be created and to allow for determination of the six-degree of freedom motions of that body segment. Vector cross products, from unit vectors connecting specific markers, produce perpendicular vectors to the marker plane. Using the newly created cross product vector, a segment-fixed coordinate system is created. Body-fixed coordinate systems of specific body segments, such as the thigh and the
shank, allow for the absolute orientation of body segments, or the relative angle between body segments to be analyzed (Schneck & Bronzino, 2002).

1.3.5 Soft-tissue artefact

Soft-tissue artifact (STA) is defined as the relative movement between the skin markers and the underlying bone (Dumas, Camomilla, Bonci, Cheze, & Cappozzo, 2014). Depending on the placement of markers, different factors will contribute to STA. When markers are placed closer to joints, inertial effects, deformation, and sliding contribute to STA. Further away from joints, muscular contraction is the main contributor to STA. Muscular contraction has a frequency content similar to that of bone movement therefore it is very difficult to distinguish between the two by using any sort of filtering technique (Leardini, Chiari, Croce, & Cappozzo, 2005).

Many studies measure and evaluate STA. Reinschmidt et al. (1997) determined the STA for the tibiofemoral and tibiocalcaneal joint motions during walking using a set of external skin markers. Intracortical Hofmann bone pins were inserted surgically into the femoral condyle, lateral tibial condyle, and the posterolateral aspect of the calcaneus of the right leg. Marker triad clusters were attached to the femur, tibia, and calcaneus bone pins. Single markers were placed on the shoe, thigh, and tibia, with six on each segment. It was concluded that most of the error for knee rotations came from STA at the thigh. Skin markers are the better option when determining flexion/extension at the tibiofemoral joint since the error was lower. The STA error was nearly as large as the magnitude of the real joint motion when trying to determine abduction/adduction and internal/external rotation at the knee (Reinschmidt et al., 1997). The same researchers looked at STA during running trials and found that for flexion/extension of the knee there was good
agreement between the skin and bone based patterns. On the other hand, the errors observed for abduction/adduction and internal/external rotation were large, 70% and 64% respectively relative to the full range of motion. It was concluded that joint motion was overestimated with the use of skin markers. STA errors at the shank were approximately 5° across all subjects and all rotations, whereas errors at the thigh reached values higher than 10° for internal/external rotation. Errors due to skin movement were higher during running trials than walking, as would be expected (Reinschmidt, van den Bogert, Nigg, Lundberg, & Murphy, 1997).

More closely related to this thesis, Westblad et al. (2002), looked at ankle complex motion during the stance phase of walking. Three markers were attached to the shank, heel, and forefoot. These were accompanied with Hoffman pins that were inserted into the tibia, fibula, talus, and calcaneus. Single markers were attached to each pin for the walking trials. Their results showed that the mean maximal difference was less than 5° between skin- and bone-based joint rotations. Moreover, the smallest absolute difference was found for plantar/dorsiflexion movement (Westblad, Hashimoto, Winson, Lundberg, & Arndt, 2002).

The type of marker used also affects the magnitude of STA. Skin-mounted markers create larger STA than markers mounted on rigid plates. Cappozzo et al. (1996), tested patients being treated for femur and tibia fractures. Unilateral external fixation devices were fixed to the bones, thus permitting a new set of axes, fixator technical frame, to be created, which would be a rigid body alongside the relevant bone. Additional skin markers were placed on the skin’s surface on anatomical landmarks; greater trochanter, lateral femoral epicondyle, head of femur and fibula, lateral malleolus for the tibia and
fibula. Clusters of three markers were placed on the pelvis and non-instrumented segments of the lower limb. Results showed that STA could be of magnitudes ranging from a couple mm up to 40mm. Skin mounted markers placed above anatomical landmarks showed displacements that were proportional to the angular displacement of the closest joint. Movement of the greater trochanter marker was affected by motion of the hip joint for example and the motion of the knee joint mostly affected movement of the head of the fibula marker. Therefore, this marker placement location is not optimal. Markers placed on the shank and thigh showed smaller displacements, indicating that this would be a better marker placement location. Greatest artefact values were seen during flexion/extension movements, from 6-20° at the femur and 4-10° in the tibia. Also, different clusters yielded different artefact results (Cappozzo, Catani, Leardini, Benedetti, & Croce, 1996).

Several conclusions can be drawn in regards to STA. Errors caused by STA are larger than errors coming from stereophotogrammetry. STA presents systematic and random errors which are reproducible within, but not among subjects. STA is task dependent, but tends to be greater in the thigh compared to other lower limb segments (Leardini et al., 2005).

When markers are formed as clusters, their movement over the underlying bone is explained by the sum of four different components. These components are translation of the cluster, rotation of the cluster about the origin of the reference frame (representing the pose of a deformable marker cluster), the change in size of the cluster, and the change in cluster shape, also called deformation. All these transformations may be independent of each other. Work done by Grimpampi et al. (2014), attempted to describe STA and its
effect on position/orientation, size, and shape of marker clusters. They defined STA of a single marker as the local displacement from a reference position fixed in the reference frame of the analyzed bone. STA at the cluster level was defined as a rigid displacement or change in position and orientation, a scaling or a change in size, and a deformation of the cluster. Steel pins were inserted into the iliac crest, proximal third of the right femoral diaphysis, and anteromedial aspect of the tibia. Each pin had a cluster of four markers placed on it. Twelve single markers were placed on the anteromedial, anterior and anterolateral aspects of the right thigh. Maximal hip and knee flexion were produced to determine STA. All the parameters describing STA saw pronounced variability across specimens and across clusters. It was found that STA’s were specimen-specific and cluster-specific. The subject with the greatest thigh mass exhibited the largest STA at the single marker and cluster level (Grimpampi, Camomilla, Cereatti, de Leva, & Cappozzo, 2013).

Therefore, the location and type of marker, as well as the body composition of the subject will have an effect on the type and amount of STA observed during kinematic analysis.

1.4 Medical Imaging

In order to overcome problems related to STA, medical imaging techniques such as magnetic resonance imaging (MRI) and computed tomography (CT) scans may be used to produce subject-specific kinematic bone models. These techniques benefit the study of kinematics because they are non-invasive when compared to radiostereometric analysis (RSA) techniques that use bone embedded tantalum beads. From the scans, subject-specific kinematic models are made with a joint coordinate system that is based on the subject’s own bony structures. A study by Scheys et al. (2011), examined the difference
in gait kinematic values using generic bone models and subject-specific MRI bone models. Full leg MRI images were taken of subjects in supine position. From the MRI images bony landmarks were identified manually. Results showed that generic bone models were substantially different from subject-specific bone models. Generic models systematically introduced significantly increased hip flexion, external hip rotation and knee flexion over the gait cycle. When MRI images were compared to kinematic analysis using VICON and reflective markers, smaller differences were found; only hip flexion was significantly increased as opposed to all three motions when using generic bone models (Scheys, Desloovere, Spaepen, Suetens, & Jonkers, 2011).

1.4.1 X-Ray and Fluoroscopy

X-rays are produced by applying a large electrical potential difference between an electron source and a target. Electrons leaving the x-ray source convert their kinetic energy to electromagnetic energy as they decelerate and interact with a target material. An external power source provides high voltage to accelerate the electrons. For diagnostic purposes, the x-ray source is placed on one side of the patient and the detector is place of the other side of the patient. During x-ray exposure, some x-rays are differentially attenuated by the anatomical structures of the patients, these are incident x-rays. Small portions of x-rays pass through the patient and are recorded on the detector, thereby creating a radiographic image (Bushberg, Seibert, Leidholdt Jr., & Boone, 2012). Fluoroscopy allows for real-time x-ray viewing of patients with high temporal resolution. Real-time imaging produces ‘videos’ with 30 frames per second (fps), which coincides with older analog television frame rates in the USA. Being able to collect video data is what separates traditional radiography from fluoroscopy. A fluoroscopic unit is formed
of different parts. Motorized collimators adjust to the field of view or the source-to-image distance. It also has a detector like other radiographic systems, but it is in the form of an image intensifier. As fluoroscopy allows for prolonged real-time image capture, extremely decreased doses of radiation have to be used. Doses may be one thousandths of that used for traditional radiography. Typical fluoroscopic detector dose ranges from 1µR to 5µR per image. Thus, the image intensifiers used are very sensitive low-noise detectors in order to detect low radiation signals. There are four components to the image intensifier: a) a vacuum housing to allow for unimpeded electron flow, b) an input layer that transforms the incident x-rays into light, c) an electron optics system that takes the electrons emitted by the input layer and transfers them to the output layer, and d) an output phosphor that converts the output electrons into a light image. Coupled with the output layer of the image intensifier, a light-sensitive camera, such as an analog vidicon, a solid-state charge couple device (CCD) or complementary metal oxide semiconductor (CMOS) system is needed in order to relay the output image to a video monitor for viewing purposes (Bushberg et al., 2012).

Continuous fluoroscopic imaging is possible by producing a continuous x-ray beam, which uses 0.5 to 6mA. Each fluoroscopic image is displayed on a camera for 33ms; hence any fast motion will be blurred. Pulsed fluoroscopy can counter the blurring of the image during continuous fluoroscopy. Pulsed fluoroscopy uses x-ray pulses that can be between 3 and 10ms in length, allowing for viewing of faster movements (Bushberg et al., 2012).
1.4.2 Computed Tomography

Computed tomography has been available for clinical use since the 1970’s. Over the past 50 years the rotation speed of the scanner has increased from a 4.5 min scan to a sub half-second rotation. Today, a CT scan can collect over 200 images per second. A scanner is composed of a CT gantry, which is the rotating part of the scanner, and a patient table that is controlled with precise motors to be positioned in the appropriate position for the scan. Laser lights also help for the proper positioning of the patient inside the bore. The scanner’s field of view is a circle in the x-y direction, but when extended in the z-direction, it becomes a cylindrical field of view. Scans are produced by having the x-ray tube rotate around the patient. Rays from the x-ray source create a fan beam projection onto detector arrays. Most detector arrays in clinical CT are arranged in an arc relative to the x-ray tube (Bushberg et al., 2012).

1.5 Radiostereometric Analysis

1.5.1 Traditional Radiostereometric Analysis

Radiostereometric analysis (RSA) comes from two words, photogrammetry and stereo. Photogrammetry means to obtain a picture that comes from light and stereo means that an object has the property of being solid, or having three dimensions (Selvik, 1989). Thus radiostereometric analysis takes measurements from 2D pictures and reconstructs three-dimensional objects. In 1898, a London radiologist, Davidson, was the first to attempt to localize bodies with the use of x-rays. He used a x-ray tube that could be moved along a horizontal scale. Below it, an x-ray plate was placed with two wires extended on top, at 90° angles with each other. Two images are taken using the x-ray tube; the negative is
developed and then brought to another machine called the localizer. In the localizer, there are two silk threads that go from the foci, during the exposures, to the images of the radiopaque object that was studied. The point in space where the two silk threads intersect represents the position of the object (Selvik, 1989). Today, RSA is a computerized system that allows for the precise location of landmarks in the human body to be known. Instead of using a localizer, modern RSA uses a cage with fiducial and control points to calculate 3D coordinate systems (Bottner, Nestor, Azzis, Sculco, & Bostrom, 2005). Since the body doesn’t have well-defined radiopaque landmarks, other markers need to be used, which are often surgically implanted into the body. The most common type of marker is a tantalum bead. The beads have a high inertness to body tissues and have a high absorption of x-rays, which makes them the ideal choice for RSA (Selvik, 1990).

There are many applications today for RSA. The first use of RSA occurred in 1973, when Aronson tested three children with delayed growth by implanting tantalum beads in the growth zone of their fibulas. Since then, multiple joints and areas of the body have been studied using RSA. Namely, RSA of the craniovertebral joints, shoulder joints, hand, spine, pelvis, hip, knee, lower extremities, ankle/foot complex, and growth disorders has been investigated (Selvik, 1990).

1.5.2 Markerless Radiostereometric Analysis

Classic RSA requires the use of tantalum markers to be inserted into bones of study or implants. This procedure is rather invasive and allows for the study of only a certain injured population. Thereby the migration of implants is one of the main study areas of RSA. The standard tantalum bead used has a diameter of 0.5, 0.8, or 1.0 mm (Valstar et
Implants like metal-backed cups for hip arthroplasty and femoral components in knee arthroplasty often hide the attached markers (Valstar, de Jong, Vrooman, Rozing, & Reiber, 2001). Tantalum beads may also compromise the strength and integrity of the implant itself. Thus markerless RSA or model-based RSA techniques have been developed to overcome the downsides of classic RSA (Hurschler, Seehaus, Emmerich, Kaptein, & Windhagen, 2009). Computer-aided design data or reverse engineering is used with this novel technique. Through these techniques, geometric surface models of prosthetics or bones can be produced. These virtual models can then be matched to the real contour of the prosthetic or a bone from a stereographic image. Hurschler et al. (2009) studied the migration of a TKA tibial component, where a manually implanted prosthetic was compared with a prosthetic implanted with the aid of a kinematic navigation system. They also compared model based and marker-based RSA techniques. Using reverse engineering, a computer model of the knee prosthetic was produced. This model was matched to stereographic images of the prosthetic. Their results showed that there was high similarity in the results between model-based and marker-based RSA. The difference in the means between model-based and marker-based ranged from -0.08mm to 0.08mm for in-plane translation, -0.14 to 0.14mm for out-of-plane translation, from -0.80 to 0.74° for out-of-plane rotation, and from -0.21 to 0.22° for in-plane rotation (Hurschler et al., 2009).

Valstar et al. (2001) tested the accuracy of a model-based RSA technique using phantom knee prosthetics. Three components of the prosthetics were analyzed, the femoral and tibial component of an Interax total knee prosthetic and the femoral component of a Profix total knee prosthetic. They used a Plexiglas cylinder, with 12 tantalum markers
embedded into its surface, to which they attached the prosthetic components to the base as a phantom. The phantom was positioned in seven different poses for each RSA radiograph. The location of the phantom and the knee implant was analyzed. The contour of the implant and the phantom were compared and the position of the implant was determined by minimizing the difference between the detected contour and the calculated model contour. The Interax component showed large standard deviation for rotations. The Profix femoral component showed smaller dimensional differences between the model and actual prosthetic. Moreover, the micromotion results were more accurate as the micromotion parameters; especially rotations were closer to zero than the observed parameters for the Interax component. This method of model-based RSA needs to have improved sensitivity to dimensional tolerances in order to get better accuracy as so to replace marker-based RSA (Valstar et al., 2001).

As an alternative to making computer-aided models of bones or prosthetics, CT scans may be used for creating bone models. A 2D-3D image registration method is used to find the 3D pose of the CT volume. Once this is done, each 2D radiograph can be matched to the 3D CT for further kinematic analysis. This method was used by Bruin et al. (2008) to determine scapular positioning with the intention of validating the procedure against conventional RSA. Image-based RSA was compared with traditional RSA using a cadaver specimen and a sawbone structure. The results showed that image-based RSA had high accuracy with migration of below 0.083mm for translations and below 0.021° for rotations. The maximum standard deviations were smaller than 0.30mm for translations and 0.33° for rotations (de Bruin et al., 2008).
1.5.3 X-Ray Fluoroscopy

Going a step ahead of RSA is x-ray fluoroscopy. This method allows for *in vivo* joint kinematics to be studied during dynamic weight-bearing activities. Fluoroscopy has several other benefits; one being that it exposes the patient to less radiation than traditional RSA. A 20 second protocol will expose the subject to 80 µSv of radiation (Ackland, Keynejad, & Pandy, 2011). Most C-arm fluoroscopy units sample at 25 fps, which is adequate for studying walking or dynamic motions of different joints. Some devices are able to capture at frame rates up to 250 fps, which allows for high-speed analysis for motions such as running, jumping, or cycling (You, Siy, Anderst, & Tashman, 2001). X-Ray fluoroscopy has been used to measure kinematics of the glenohumeral joint (Fox, Kedgley, Lalone, Johnson, & Athwal, 2011), hip joint (Ioppolo et al., 2007; Tsai et al., 2013), femur (Baka et al., 2012; Hurvitz & Joskowicz, 2008), knee (Acker et al., 2011; Banks & Hodge, 1996; Ioppolo et al., 2007; Li, Van de Velde, Samuel K., & Bingham, 2008; Tersi, Barré, Fantozzi, & Stagni, 2013), ankle and foot (Martin et al., 2012). Most x-ray fluoroscopy analysis is done by single-plane or dual-plane. Single-plane fluoroscopy allows for determination of motions with six degrees-of-freedom (three for translation and three for rotation), but shows rather large out-of-plane motion errors. Accuracy is increased when using dual-plane fluoroscopy; nonetheless, single-plane is a useful and a valid way to measure joint kinematics (Ackland et al., 2011). Acker et al. (2011) used single-plane fluoroscopy and determined its accuracy by comparing it to optical motion tracking. Knee joint kinematics showed absolute mean differences between both methods of 2.1°, 0.3°, and 1.1° in extension, abduction, and internal rotation respectively, and 1.3, 0.9, and 1.9mm in anterior, distal, and lateral
translations respectively (Acker et al., 2011). Similarly, Banks et al. (1996) measured the accuracy of single-plane fluoroscopy on knee replacement kinematics and found that knee rotations could be measured to the accuracy of 1° and knee translations could be measured with an accuracy of about 0.5mm. Their method for measuring accuracy was different, relative poses of implant components against the radiograph were measured and accuracy was determined as the estimate pose relative to a modeled pose (Banks & Hodge, 1996).

Tersi et al. (2013) compared the accuracy of single-plane and bi-plane fluoroscopy. They performed their analysis on dynamic movements of the tibiofemoral joint. Using the same movements for both fluoroscopic techniques, they validated them against dynamic fluoroscopic marker-based RSA, which is considered the gold standard. A sawbone model of the knee joint was made with four tantalum beads embedded in it. Five repetitions of 10s were performed for three motions, absolute pose kinematics for each bone segments were calculated for single-plane, bi-planar 3D fluoroscopy, and RSA. Their results showed that for single-plane fluoroscopy, when calculating in-plane pose parameters, un-biased and low dispersion pose estimates could be obtained. The errors for in plane pose parameters were of the same magnitude for single-plane and dual-plane fluoroscopy. Magnitude of translational errors was less than 0.5mm for single-plane and 0.3mm for dual-plane. Whereas, rotation error was two fold for single-plane compared to dual-plane and only one magnitude greater for translation errors (Tersi et al., 2013).
1.6 Fluoroscopic Calibration

When placing two fluoroscopic devices in a laboratory setting, their specific location will have to be known for data analysis. Laboratory coordinate systems have to be determined in order to track the movement of specific anatomical landmarks and joints of interest. Calibration frames have been developed in order to achieve this. Calibration frames have two sets of planes, control and fiducial. The fiducial plane is used to transform the image coordinate system to the laboratory coordinate system. The control plane is used to determine the focal point from which the x-rays originate (Kedgley & Jenkyn, 2009). Most often bi-planar fluoroscopy is used with the fluoroscopes being placed at 90° angle relative to each other. With this arrangement, calibration boxes have been developed in the shape of cubes (Valstar et al., 2005).

This thesis does not place the fluoroscopes at 90°. A non-traditional orientation was chosen in order to get the best view of a joint or anatomical structure. Placing C-arm fluoroscopes at 90° relative to each other is very restricting for the study of movement of joints. Kedgley and Jenkyn (2009) assessed the accuracy of RSA when imaging devices were placed in a non-traditional orientation. They used both an orthogonal calibration frame (where the fiducial and control planes were oriented 90° relative to each other) and a calibration frame where fiducial and control planes were oriented at an angle greater than 90°. A calibration frame was constructed from an acrylic sheet with each fiducial and control planes embedded with 45 beads in a 9-bead by 5-bead matrix. Control and fiducial planes could be set to be at 90°, 105°, 120°, and 135° relative to each other. The use of an angled calibration frame did not improve the accuracy of the overall calibration of the system. When the fluoroscopes were placed at angles equal to or smaller than
135°, the use of the calibration frame at 90° showed equivalent or better accuracy than when the fluoroscopes were positioned at right angles. Greatest accuracy values were in the range of 90.0±24.0μm for calibration frame and fluoroscopes placed at 105° and lowest accuracy was found with a 135° position having magnitudes of 227.2±120.9μm. Accuracy for a 90° calibration frame and fluoroscope placement fell between these two values. Thus, RSA imaging can be performed with the devices being placed at relative angle to each other of other than 90° with proper accuracy (Kedgley & Jenkyn, 2009). Figure 1-4 is an example of how the fluoroscopes and calibration frame is placed in this study.
1.6.1 Pincushion Distortion

Fluoroscopic analysis may cause extensive spatial distortion of radiographic images (Wearing et al., 2005). Types of distortions that may occur are pincushion distortion, shading, veiling glare, characteristic curve, and de bias. Pincushion distortion is the most significant cause of spatial non-linearity. It is caused by the combination of a curved design of the image intensifier and limitations of electron focusing, which result in a non-
uniform magnification of the peripheral aspect of the image (Boone, Seibert, Barrett, & Blood, 1991).

Distortion is most often corrected with a grid of beads or wires that is placed in front of the image intensifier to quantify the amount of distortion that is present. There are several ways to correct for distortion. Local distortion correction algorithms use three or four points that surround a small area of an image and correct for that area. Global distortion correction algorithms use the coordinates of as many grid points as can be seen in the image and calculate a distortion vector at each point. Positions of the beads in the image are related to the known positions of the beads according to a polynomial. Global distortion correction was found to be superior to local distortion correction techniques as it is not only considerably faster, but it also has less digitization error than local distortion as it is removed by using a least-squares fit method (Gronenschild, 1997). After each testing procedure during this thesis, pincushion distortion had to be assessed. The technique used was the one previously developed in the lab by Kedgley et al. (2012). A grid made of a 9.5mm thick Delrin sheet with 131 2-diameter stainless steel beads spaced apart by 15mm was used (Figure 1-5 and 1-7). The positions of the beads were determined using a coordinate measuring machine. After a testing session, the grid was attached to each image intensifier and an image with the position of the beads was collected. The position of each bead was located manually using a custom-written algorithm in MATLAB (Figure 1-6). A range of polynomials was used for distortion correction, from first degree in each direction (second order polynomial) to third degree in each direction (sixth order polynomial). Kedgley et al. (2012) found that a fourth order polynomial was preferred for their investigation, hence second degree in each
direction polynomial fit. They also found that image distortion is most important for 2D analysis. The use of a calibration frame for 3D analysis tempers the effects of distortion, which lead to accuracies in the RSA reconstruction with uncorrected points that were much better than anticipated. The error in RSA reconstruction of uncorrected points was found to be 192±68µm (Kedgley, Fox, & Jenkyn, 2012).

Figure 1-5: Distortion grid attached to fluoroscope B. Known locations of stainless steel beads embedded in the plastic allow correction for image distortion.
Figure 1-6: Sample image of distortion grid taken by fluoroscope B.

Figure 1-7: Close up view of the distortion grid with the numbered beads used for MATLAB algorithms.
1.6.2 Experimental Setup Recreation

Through the use of MATLAB algorithms (The MathWorks; Natick, MA, USA) the x-ray source positions, the orientation and location of the image plane with respect to the x-ray source were determined. This was done by determining and optimizing three Euler angle rotations and the distance ‘d’ from the source to the image plane. Once the fluoroscopic parameters are determined, the experimental set-up can be recreated in solid modeling software Rhinoceros (Rhinoceros, Robert McNeel & Associates, Seattle, WA, USA). The virtual set-up allows for the import of bone models that will be matched to the fluoroscopic images. The set-up was done following the instructions from Appendix E and F from Anne-Marie Allen’s thesis (2009). The first step orients the fluoroscopic coordinate system in the correct orientation. A point for the x-ray source is recreated using the x-ray source coordinates that were found by running the MATLAB algorithms. A vector of length ‘d’ is created from the last rotation of the fluoroscope coordinate system. The vector is linked to the x-ray source and an image plane orthogonal to the vector is created and is coincident with the other end of the ‘d’ vector. The image plane is formed according to the known size of the fluoroscopic images (540x720 pixels). The fluoroscopic calibration images are imported into the image plane as are the 2D distortion-corrected fiducial and control points. These points have to be aligned with the 3D calibration frame points for the final image plane correction. Each fluoroscope calibration is done separately and then one is imported into the other in order to have a virtual recreation of the laboratory set-up.
3D bone models have to be imported into the recreated lab set-up to be matched to the fluoroscopic images in this thesis. The bone-models used for the second chapter of this thesis are subject-specific bone models, which were created from CT scans of the tested subjects. The bone models used in the third chapter of this study were taken from a bank of ‘generic’ CT scans and matched to the subject’s foot by size.

The CT scans are converted into bone models in an open source image processing and DICOM viewing software OsiriX (Pixmeo, Geneva, Switzerland). Bone segmentation step-by-step instructions are presented in Appendix 4. Each bone of interest is segmented individually in order to be imported into Rhinoceros for matching. The 3D Volume Rendering window allows for segmentation and isolation of each individual bone using specific tools. It is during this part of bone model creation that the bony landmarks are placed on the bones (Figure 1-8). The bony landmarks used in the third chapter of this thesis are reported in Table 1.
Table 1-1: Name and position of landmarks placed on the bone models

<table>
<thead>
<tr>
<th>Trial #</th>
<th>Segment</th>
<th>Tracked landmarks</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Hindfoot</td>
<td>CAER: eminentia retrotrochlearis (greatest lateral elevation)</td>
</tr>
<tr>
<td>2</td>
<td></td>
<td>CALT: lateral tuberosity (lateral to Achilles tendon attachment)</td>
</tr>
<tr>
<td>3</td>
<td></td>
<td>CAMT: medial tuberosity (medial to Achilles tendon attachment)</td>
</tr>
<tr>
<td>4</td>
<td></td>
<td>STH: talar head (most dorsal points at joint with navicular)</td>
</tr>
<tr>
<td>5</td>
<td>Midfoot</td>
<td>MCI: first cuneiform (distal dorsal crest)</td>
</tr>
<tr>
<td>6</td>
<td></td>
<td>MNT: navicular tuberosity (most medial point)</td>
</tr>
<tr>
<td>7</td>
<td></td>
<td>MCU: cuboid (lateral dorsal edge at joint with calcaneus)</td>
</tr>
<tr>
<td>8</td>
<td>Medial Forefoot</td>
<td>MIH: first metatarsal head (most dorsal point)</td>
</tr>
<tr>
<td>9</td>
<td></td>
<td>MIB: first metatarsal base (most dorsal point)</td>
</tr>
<tr>
<td>10</td>
<td>Lateral Forefoot</td>
<td>MVH: fifth metatarsal head (most dorsal point)</td>
</tr>
<tr>
<td>11</td>
<td></td>
<td>MVB: fifth metatarsal base (most dorsal point)</td>
</tr>
<tr>
<td>12</td>
<td>Ankle JCS</td>
<td>LMM: medial malleolus (most medial point) defined on the lower leg segment</td>
</tr>
<tr>
<td>13</td>
<td></td>
<td>LLM: lateral malleolus (most lateral point) defined on the lower leg segment</td>
</tr>
<tr>
<td>14</td>
<td>Hallux</td>
<td>DH: most distal point of the hallux (and the foot)</td>
</tr>
<tr>
<td>15</td>
<td></td>
<td>LPH: lateral head of the interphalangeal</td>
</tr>
<tr>
<td>16</td>
<td></td>
<td>MPH: medial head of the interphalangeal</td>
</tr>
</tbody>
</table>
1.6.4 Matching

The goal of matching the bone models to the fluoroscopic images is to recreate their position and orientation in the field of capture of the fluoroscopes. Bone models created in OsiriX are imported into Rhinoceros where they can be rotated and translated in three dimensions (Figure 1-9). The anatomical landmarks of interest are first identified as dots from the black mesh that was exported from OsiriX. The bones are imported as set of grouped models; this allows for an initial visualization of the orientation of the bones. Then the bones are ungrouped so that each individual bone can be matched to its respective fluoroscopic bone image. The bones are first moved by 1° or 1mm to get an initial match. Then the image is enlarged for fine-tuning of the matching, where bones

Figure 1-8: Bone model generation in OsiriX. The red dots correspond to the bony landmarks that will be used for the study of joint movements with the multi-segment foot model (OsiriX, Pixmeo, Geneva, Switzerland).
are moved by increments as small as 0.01mm and 0.01°, as described by Allen (2009). Specific bone landmarks, such as the outline of the lateral calcaneus or the metatarsal shafts, are used to align the bones in the proper orientation. The bones may be moved by 1mm, 0.5mm, and 0.05mm increments in order to properly align their silhouettes. Once the bones are aligned, the anatomical landmarks are exported into a spreadsheet using a custom RhinoScript created by Allen (2009) (Rhinoceros, Robert McNeel & Associates, Seattle, WA, USA). The bony landmark coordinates were used for the calculation of different joint motions using the MSFM code from Shultz (2009). New bone models and fluoroscopic images are imported for each new condition for matching.
1.7 Rationale

The rationale of this study was based on previous research that has been accomplished in our laboratory. A MSFM was developed by Jenkyn and Nicol (2007) for use during running and investigating gait kinematics in barefoot and shod motion. This model is designed to measure gait kinematics using an optical motion-capture system with stereophotogrammetric cameras. This method has many benefits, such as it is effective at studying motions of the foot, it is non-invasive, safe for the patient, and post-processing is not as extensive as other measurement techniques. On the down side, optical motion capture gives rise to STA since this technique requires marker clusters be placed on the surface of the skin overlying the bones. Another source of error comes from determining the location of anatomical landmarks through palpation.
Previously developed in the Wolf Orthopaedic Quantitative Imaging Laboratory (WOQIL) is a markerless fluoroscopic RSA system, which was validated by Anne-Marie Allen (2009) and is now used for measuring in-vivo kinematics. This measurement technique is considered the gold standard for measuring foot-joint kinematics. It allows for the capture of static positions and dynamic motions of the foot. The sources of error present with optical motion-capture are eliminated as the anatomical bony landmarks may be identified on the bones themselves when producing subject-specific or generic bone models. The downfall of this method on the other hand is that it requires the subject to have increased exposure to radiation, both when getting a CT scan for the production of bone models or during testing using the fluoroscopic RSA system. Moreover, this technique is expensive, especially when subject-specific bone models are required, as every subject needs to get a CT scan of the studied bones. Post processing requires a lot more time than optical motion analysis; thereby a lesser number of subjects can be tested given the same amount of time. Thus, evaluating the MSFM using the optical motion capture system against the fluoroscopic RSA system will allow us to determine if the optical motion capture system is accurate enough for the study of foot-joint kinematics.

1.8 Objectives and Hypothesis

The main objective of this study was to validate the MSFM when used with optical motion capture against the fluoroscopic RSA system. These following objectives allowed for the reach of the main objective:

1. Compare medial longitudinal arch (MLA) angle when using subject-specific bone models and generic bone models for matching when using the fluoroscopic RSA
system with the intention of validating generic bone models for the use of RSA fluoroscopy.

2. Compare the motions of the MSFM used by Shultz (2009) when using optical motion capture and fluoroscopic RSA.

It was hypothesized that:

1. The use of generic bone models gives accurate results for fluoroscopic RSA. Generic bone models will estimate the MLA angle found using subject-specific bone models by less than 5°.

2. The MSFM using optical motion capture can track foot segment motions with errors of less than 1° when compared to bi-planar RSA fluoroscopy.

1.9 Thesis Overview

Chapter 2 validates the use of generic bone models. This chapter compares the use of subject-specific bone models with the use of generic bone models for the calculation of the MLA angle. Chapter 3 validates the use of optical motion capture when using the MSFM to determine the action of joints in the foot for a period of the gait cycle. It looks at the joint motions produced by the MSFM and compares the results when they are calculated using optical motion capture and when using bi-planar fluoroscopic RSA. Chapter 4 summarizes the conclusions drawn from this thesis and discusses its significance for future research.
Chapter 2 – Validation of Generic Bone Models for the Use With Bi-Planar RSA Fluoroscopy to Evaluate the Medial Longitudinal Arch

2.1 Introduction

Model-based fluoroscopic RSA is becoming a more common tool for studying the biomechanics of joints and movement. The complete 3D pose (position plus orientation) of each bone of interest can be measured using fluoroscopic RSA with sub-millimeter and sub-degree accuracy (Ackland et al., 2011).

Marker-based fluoroscopic RSA was initially used for the study of movement of orthopaedic implants. Implants with tantalum beads embedded within them were placed into the patient during surgery. Disadvantages of this method are the possibility of beads being hidden by parts of the implant and the reduction of strength of the implant due to the inserted beads. A model-based pose-estimation method, where the contour of a model is aligned with the contour of the actual prosthetic, has been found to be interchangeable with marker-based methods and is currently the preferred technique (Hurschler et al., 2009).

The model-based technique requires 3D models of the bones of interest to be created before data is collected for matching with 2D fluoroscopic images. These bone models are usually created from CT scans (de Bruin et al., 2008; Dennis, Mahfouz, Komistek, & Hoff, 2005; Fox et al., 2011; Hurvitz & Joskowicz, 2008; Torry et al., 2011) or MR scans (Arnold, Blemker, & Delp, 2001; Baka et al., 2012; Li et al., 2008; Scheys et al., 2006) of the patient’s bony anatomy. Subject-specific bone-models produced from CT or MR scans prior to testing allows for accurate matching with the patients’ fluoroscopic images.
Moreover, model-based techniques allow for testing of patients with unusual or pathological anatomies or who have previously undergone fractures (Hurvitz & Joskowicz, 2008). However, this technique is very labour intensive and requires manual segmentation of the bones needed for analysis (Scheys et al., 2006). It also requires the patient be subjected to an extra dose of x-ray radiation (in addition to the fluoroscopic data collection). There are risks associated with exposure to x-ray radiation, and the dosage to the patient should be kept to a minimum. A previous study from our laboratory found that decreasing CT dosage by 98% only negligibly reduced the accuracy of our fluoroscopic RSA measurements (Fox et al., 2011). Although, eliminating the required CT scans altogether would be an even greater benefit to the patient by making the testing procedure safer and less time intensive. In the absence of bone models from subject-specific CT scans, generic bones models of the bones of interest can be used. These can come from a library of CTs from other patients, or from bone mimicking objects such as sawbones. The elimination of CT and MR scans would reduce hospital and research costs. The use of generic bone models will allow for a faster post-processing time.

The purpose of this study was to investigate the use of generic bone models versus subject-specific bone models on the accuracy of our fluoroscopic RSA technique in its use to measure the behaviour of the MLA of the left foot.

### 2.2 Methods

Five participants, one male and four female, were selected for this study. Participants had an average age of 32 ± 1 years. Subjects had no prior diagnosed foot problems or injuries. They read and signed a consent form prior to testing. The subjects were required to walk along a custom built wooden platform while fluoroscopic images of
their left foot were taken. Prior to testing subject anthropometric data was collected (Table 2-1) and each subject had a CT scan of his or her feet done. Their foot type was determined by a pedorthist from the Fowler Sports Medicine Clinic. The subjects were placed into a normal, planus, or cavus foot type category. From the CT scans, bone models using OsiriX (Pixmeo, Geneva, Switzerland) were created for the calcaneus, navicular, and first metatarsal.

<table>
<thead>
<tr>
<th>Subject</th>
<th>Gender</th>
<th>Age</th>
<th>Left Foot Length (cm)</th>
<th>Foot Type</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>F</td>
<td>25</td>
<td>23.7</td>
<td>Normal</td>
</tr>
<tr>
<td>2</td>
<td>F</td>
<td>20</td>
<td>24.1</td>
<td>Normal</td>
</tr>
<tr>
<td>3</td>
<td>M</td>
<td>23</td>
<td>26.4</td>
<td>Planus</td>
</tr>
<tr>
<td>4</td>
<td>F</td>
<td>18</td>
<td>26.0</td>
<td>Planus</td>
</tr>
<tr>
<td>5</td>
<td>F</td>
<td>50</td>
<td>23.8</td>
<td>Cavus</td>
</tr>
</tbody>
</table>

2.2.1 Calibration

The fluoroscopes (SIREMOBIL Compact (L); Siemens Medical Solutions USA Inc., Malvern, PA, USA) were placed in such a way that each device would have the same view of the foot, but from a different angle. They were placed such that one image intensifier would capture a sagittal, lateral view of the foot and one image would capture an oblique anterior/posterior view of the foot. Once the fluoroscopes were in their required positions, they were calibrated by imaging a calibration box designed by Kedgley (2009c) with beads embedded in known locations. The box was placed such that each fluoroscope viewed a fiducial and control plane. Following testing, the
calibration box was imaged again with the addition of an image distortion grid to correct for pincushion distortion. Custom written algorithms in MATLAB (The MathWorks, Natick, MA, USA) were used to finalize the calibration process and recreate the laboratory set up for data processing.

2.2.2 Data Collection

Prior to testing, subjects wore a lead skirt, vest, and thyroid shield as protection against radiation. Subjects were required to stand still on the platform with their left foot in the field of view of the fluoroscopes while the laboratory x-ray technician took a still x-ray image of their foot (Figure 2-1). This process was repeated until one good image was taken in which the edge of the back of the calcaneus and part of the shaft of the first metatarsal was seen. Subjects performed the trials barefoot.
Figure 2-1: Laboratory set-up with a model foot in position to have fluoroscopic images taken of the left foot.
2.2.3 Data processing

From the static standing trials, one frame was exported using Adobe Premiere Pro (Adobe Systems Inc., San Jose, CA, USA) into a .tif format. After recreation of the experimental setup in a solid modelling program Rhinoceros (Robert McNeel & Associates, Seattle, WA, USA), the select .tif images and the bones models were imported for matching. The bone models were aligned with both radiographic images. Bone model contours were consistently aligned the same way, for all matching conditions, with the bone contours on the radiographic image. Each image was matched to the subject-specific bone model (SS), and four generic bone models. The generic bone...
models were selected from a bank of CT scans that were collected previously in the laboratory. The generic bone models were chosen by matching as close as possible the subject’s left foot length to the generic bone model’s left foot length as well as the type of the foot. The generic bone models were of: 1) same foot length and type (G_SS_ST), 2) same foot length and different types (G_SS_DT1 and G_SS_DT2), and 3) different foot length and same foot type (G_DS_ST). Bony landmarks were exported using a custom written script in Rhinoceros. A custom written MATLAB code was run on the landmarks to determine the MLA angle of the foot. MLA angle was calculated by forming the angle between the medial process of the calcaneus, navicular tuberosity, and distal head of the first metatarsal (Figure 2-2). MLA angle was compared between subject-specific bone models and generic bones models.

![Figure 2-3: Landmarks used to calculate the medial longitudinal arch angle value. Angle theta is formed by the medial process of the calcaneus, navicular tuberosity, and distal head of the first metatarsal (Balsdon, 2011).](image)
2.3 Results

Subjects were matched to their subject-specific bone models produced from a CT scan and to four generic bone models. The size and type of bone models is presented in Table 2-2. Two subjects had a normal foot type, two subjects had a planus foot type and one subject had a cavus foot type.
Table 2-2: Matching conditions for all 5 subjects. Each subject was matched with 1) a subject-specific bone model (SS), a generic bone model that was of the 2) same size and same type foot (G_SS_ST), 3) same size and 1st different type foot (G_SS_DT1), 4) same size and 2nd different type foot (G_SS_DT2), 5) different size and same-type foot (G_DS_ST).

<table>
<thead>
<tr>
<th>Subject #</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
<th>5</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Size (cm)</td>
<td>Type</td>
<td>Size (cm)</td>
<td>Type</td>
<td>Size (cm)</td>
</tr>
<tr>
<td>SS</td>
<td>23.7</td>
<td>normal</td>
<td>24.1</td>
<td>normal</td>
<td>26.4</td>
</tr>
<tr>
<td>G_SS_ST</td>
<td>24.1</td>
<td>normal</td>
<td>23.7</td>
<td>normal</td>
<td>26.0</td>
</tr>
<tr>
<td>G_SS_DT1</td>
<td>23.8</td>
<td>cavus</td>
<td>24.1</td>
<td>cavus</td>
<td>26.5</td>
</tr>
<tr>
<td>G_SS_DT2</td>
<td>23.9</td>
<td>planus</td>
<td>23.9</td>
<td>planus</td>
<td>26.7</td>
</tr>
<tr>
<td>G_DS_ST</td>
<td>26.7</td>
<td>normal</td>
<td>28.1</td>
<td>normal</td>
<td>29.7</td>
</tr>
</tbody>
</table>
The MLA angles for all subjects and conditions are presented in Table 2-3 and in a visual representation in Figure 2-3. The MLA angles ranged between $124.3^\circ$ and $141.2^\circ$ for the subject-specific bones models. The G_{SS\_ST} bone models were the closest to the SS bone models for subjects 1 and 5. The closest bone model to the SS bone model for subject 2 was the G_{DS\_ST} bone model. The G_{SS\_DT1} and G_{SS\_DT2} models showed the closest MLA angle to the SS bone model for subjects 3 and 4.

<table>
<thead>
<tr>
<th>MLA angle (degrees)</th>
<th>Subject 1</th>
<th>Subject 2</th>
<th>Subject 3</th>
<th>Subject 4</th>
<th>Subject 5</th>
</tr>
</thead>
<tbody>
<tr>
<td>SS</td>
<td>124.3</td>
<td>141.4</td>
<td>134.4</td>
<td>139.5</td>
<td>132.2</td>
</tr>
<tr>
<td>G_{SS_ST}</td>
<td>122.8</td>
<td>139.5</td>
<td>132.4</td>
<td>142.4</td>
<td>129.1</td>
</tr>
<tr>
<td>G_{SS_DT1}</td>
<td>130.2</td>
<td>131.6</td>
<td>135.8</td>
<td>145.0</td>
<td>124.2</td>
</tr>
<tr>
<td>G_{SS_DT2}</td>
<td>128.5</td>
<td>130.9</td>
<td>132.9</td>
<td>140.5</td>
<td>125.3</td>
</tr>
<tr>
<td>G_{DS_ST}</td>
<td>127.9</td>
<td>142.3</td>
<td>139.9</td>
<td>147.0</td>
<td>125.4</td>
</tr>
</tbody>
</table>

Table 2-4 displays the angle difference for each subject between the generic conditions and the subject-specific MLA angle value. The differences are presented as an overestimation (positive number) or underestimation (negative number). The mean differences and respective standard deviations were calculated using absolute difference values. The mean difference for G_{SS\_ST} was $2.3^\circ \pm 0.7^\circ$ thus, subjects 2, 3, and 4 all fall within $\pm 1$ SD of the mean angle difference. The mean difference for both G_{SS\_DT1} and G_{SS\_DT2} grouped together was $5.5^\circ \pm 3.5^\circ$. The mean difference for G_{DS\_ST} was $4.8^\circ \pm 2.7^\circ$. This table indicates that the G_{SS\_ST} model had the smallest angle difference from the SS condition.

No pattern of under- or overestimation of the MLA angle was seen over all five subjects. For subject 1, 3 out of the 4 generic bone models overestimated the MLA angle where as
for subject 2, 3 out of the 4 generic bone models underestimated the MLA angle. For subject 4, all the generic bone models overestimated the MLA angle and for subject 5 all the generic bone models underestimated the MLA angle. For subject 3, 2 generic bone models overestimated the MLA angle and 2 underestimated the MLA angle.

Table 2-4: Difference in degrees between each generic condition and the subject-specific condition for each subject. The difference is presented as an overestimation (positive number) or underestimation (negative number). The mean and standard deviation across all subjects for each condition is also indicated. The average and standard deviation was calculated based on absolute differences.

<table>
<thead>
<tr>
<th>Degree Difference (°)</th>
<th>Subject 1</th>
<th>Subject 2</th>
<th>Subject 3</th>
<th>Subject 4</th>
<th>Subject 5</th>
<th>Mean</th>
<th>SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>G_SS_ST</td>
<td>-1.5</td>
<td>-1.9</td>
<td>-2.0</td>
<td>2.8</td>
<td>-3.1</td>
<td>2.3</td>
<td>0.7</td>
</tr>
<tr>
<td>G_SS_DT1</td>
<td>5.9</td>
<td>-9.9</td>
<td>1.4</td>
<td>5.4</td>
<td>-8.1</td>
<td>6.1</td>
<td>3.2</td>
</tr>
<tr>
<td>G_SS_ST2</td>
<td>4.1</td>
<td>-10.5</td>
<td>-1.4</td>
<td>1.0</td>
<td>-6.9</td>
<td>4.8</td>
<td>4.0</td>
</tr>
<tr>
<td>G_DS_ST</td>
<td>3.6</td>
<td>0.8</td>
<td>5.5</td>
<td>7.5</td>
<td>-6.8</td>
<td>4.8</td>
<td>2.7</td>
</tr>
</tbody>
</table>

Figure 2-4 plots the difference in angle between the generic bone model conditions and the subject-specific bone model condition for all subjects. The G_SS_ST line is the smoothest and closest line to 0. This shows that over the five subjects, the G_SS_ST bone model had the least variability.
Figure 2-4: Medial longitudinal arch (MLA) angle in degrees for each subject across each condition.

Figure 2-5: Medial longitudinal arch (MLA) degree angle difference for each generic condition from the subject-specific condition for all subjects. Values above 0 indicate an overestimation of the MLA angle and values under 0 indicate an underestimation of the MLA angle.
2.4 Discussion

The purpose of this study was to investigate the use of generic bone models versus subject-specific bone models for the evaluation of the MLA angle during quiet standing. Subject-specific bone models were produced from the subject’s respective CT scans prior to testing. The generic bone models were produced from a ‘bank’ of CT scans. Each subject was matched to four generic bone models of:

1) Same size, same type foot
2) Same size, different type foot 1
3) Same size, different type foot 2
4) Different size, same type foot.

The MLA angle was evaluated from fluoroscopic images during quiet standing. The average MLA angle of all the conditions was $133.8° \pm 7.1°$, which is in accordance with values in the literature (Saltzman, Nawoczenski, & Talbot, 1995). The same size and same type foot generic bone model had a mean difference of $2.3° \pm 0.7°$; this was the smallest difference from the subject-specific bone model when evaluating the MLA angle (Table 2-4).

No trend among the subjects was observed with the generic bone models. Only two subjects demonstrated a consistent pattern. The generic bone models underestimated the MLA angle for subject 5 and overestimated the MLA arch for subject 4. The other subjects showed a lot of variability with some generic bone models overestimating the MLA angle and some underestimating it. This large variability in results can be attributed to the way the generic models were selected. The selection of the generic bone models was based solely on the length of the subject’s foot and their type of foot. Even if
the length and type of foot of the subject and generic model was the same, that didn’t mean that the shape of the bones were similar to each other. Considering that this type of data processing is qualitative, bone models should also be matched to the subject’s fluoroscopic images according to their shape. Table 2-4 refers to the difference in MLA angle between the generic and subject-specific bone models. For subjects 2,3, and 4, the best generic bone models were the DS_ST, SS_DT1 & SS_DT2, and SS_DT2 respectively. Although the size of the foot and type of foot was different, the bony contours and landmarks of interest matched the fluoroscopic image the best.

Previously, some studies have looked at the accuracy of statistical-shape models (SSM) compared to generic CT or MR images for the study of kinematics. Baka et al. (2012) looked at drop-landing sequences at the distal femur and found that SSMs, when compared to a segmented subject-specific bone surface, had a tracking accuracy or 1-1.5mm. They found the greatest errors occurring at the rotation of the femoral shaft. They concluded that eliminating CT or MR scans when using fluoroscopy was appropriate (Baka et al., 2012).

A limitation in this study was the small sample size. A greater pool of subjects and bank of generic bone models would allow for a more representative analysis. With a greater sample size, differences in joint angles within normal, planus, and cavus foot types can be analyzed to look for systematic patterns within these groups. Further research in this direction will allow for better use of fluoroscopy in clinical settings. Although, no specific trend was observed and there was a lot of variability within subjects, the small MLA angle difference between the G_SS_ST model and SS model proves to be promising for future fluoroscopic analysis using generic bone models. This outcome will
allow elimination of a prior CT scan, which will decrease the amount of radiation patients are exposed to. Moreover, eliminating the making of a subject-specific bone model will speed up the testing protocol as currently segmentation needs to be done manually, for each individual bone, which is very labour intensive.

2.5 Conclusion

This study looked at the effectiveness of using generic bone models as opposed to subject-specific bone models in the evaluation of the MLA angle during quiet standing. Large variability was found between the different generic bone models when predicting the MLA angle. Overall, the best generic bone model was the model that was of the same size and same type foot as the subject. This has great relevance in a clinical setting as generic bone models would allow for elimination of prior CT scans before testing, which would prove to be safer for patients and would yield faster testing procedures. Thus, creating a bank of generic bone models produced from CT scans of people with different size and different type feet would prove to be beneficial in a clinical or laboratory setting.
Chapter 3 – Validation of the Multi-Segment Foot Model against Bi-Planar RSA Fluoroscopy

3.1 Introduction

During clinical gait analysis using optical motion capture the foot is usually assumed to be a single rigid segment. This approach is useful since it is simple and can measure the progression angle of the foot, the dorsi/plantar flexion of the ankle and the inversion/eversion of the subtalar joint (Carson et al., 2001). However, this approach is restrictive when clinically relevant kinematic information is needed about the motions of the joints within the foot, since these cannot be measured with simple methods. It is well known that the joints of the foot work together with the ankle and subtalar joints during gait, allowing for flexibility of the foot and the safe transfer of large biomechanical loads during walking. At the beginning of the stance phase of walking, the foot is flexible and compliant. Later in stance, the foot transitions into a rigid lever through which large propulsive forces can be applied to the ground. This dual function of the foot can be disrupted by injury. Therefore, measuring the motion of the joints of the foot during functional, weight-bearing activities, such as walking is clinically necessary (Jenkyn & Anas, 2009).

Measuring foot joint motions requires that the foot is tracked as several connected segments, rather than a single segment. Several MSFMs have been described in the literature to address this problem. Previous research has looked at motions of specific segments of the foot separately: such as the forefoot and hindfoot segments (Carson et al., 2001), medial and lateral forefoot segments (Kidder et al., 1996) and midfoot (tarsal)
segment (Leardinin et al., 1999). The MSFM used in this study was developed by Jenkyn and Nicol (2007). This model tracks the foot as four connected segments: the hindfoot (calcaneus), midfoot (cuneiforms I-III, navicular, cuboid), medial forefoot (metatarsal I and II) and lateral forefoot (metatarsal III-V) (Jenkyn & Nicol, 2007).

Optical tracking systems are the most common way of quantifying three-dimensional joint kinematics in patients and volunteers in a laboratory setting. Reflective markers are placed at various anatomical landmarks and their movement in three-dimensions is recorded by multiple digital cameras (Kedgley, Birmingham, & Jenkyn, 2009). While optical motion capture has the advantage of allowing the patient a great freedom of movement, a limitation is that the reflective markers must be attached to the skin or clothing. Therefore, there is always an amount of relative motion between the markers and the bones they are tracking. This is known as ‘soft-tissue artefact’ or STA. STA introduces error into each reflective marker trajectory (Shultz, Kedgley, & Jenkyn, 2011a). At present, the magnitude of the errors in each marker trajectory is unknown and their effect on the overall measurement foot segment kinematics is also unknown. This currently limits the clinical applicability and validity of MSFMs (Deschamps et al., 2012).

A more accurate method for tracking bone kinematics has become available with three-dimensional bi-planar fluoroscopic analysis (Garling et al., 2007). X-ray fluoroscopy allows for the direct and simultaneous visualization of the skin-mounted markers of the optical system and the bones themselves during dynamic, weight-bearing activity (Shultz et al., 2011a). Bi-planar fluoroscopy uses two fluoroscopes at different angles to simultaneously image bones of interest. Using the radiostereometric analysis (RSA)
method and virtual models of the bones of interest, created from patient computed tomography (CT) scans, the three-dimensional motion of each bone of interest is reconstructed. Previous work in our laboratory has demonstrated that bi-planar fluoroscopy can be applied to the bones of the foot during walking gait. X-ray fluoroscopy and the CT scans used to produce the virtual bone models require that the patient be exposed to x-ray radiation with its associated risks (Fox et al., 2011). Every effort is made to keep this exposure to an absolute minimum during the testing protocol. In the current study, the kinematics of the joints of the foot using the MSFM with optical motion capture will be validated against bi-planar RSA fluoroscopy. The purpose of this study was to validate the measurements of the motion of the joints of the foot by optical motion capture and a MSFM against the gold standard of bi-planar RSA fluoroscopy in normal feet. The errors in the optical motion capture trajectories, such as STA; will be quantified for walking. It was hypothesized that the MSFM using motion capture will track foot joint kinematics with an accuracy of less than 1° when compared to bi-planar RSA fluoroscopy.

3.2 Methods

3.2.1 Subjects

Five subjects with an average age of 26 ± 5.7 years were recruited for this study. Three were male and two were female. Subjects were not diagnosed with any prior foot problems. Prior to testing, subject anthropometric data was collected (Table 3-1). The institution’s Research Ethics Board for Health Science approved this study and all participants gave their informed, signed consent.
Table 3-1: Subject information data.

<table>
<thead>
<tr>
<th>Subject</th>
<th>Gender</th>
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<th>Height (cm)</th>
<th>Weight (kg)</th>
<th>Left foot length (cm)</th>
</tr>
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<td>M</td>
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<td>185</td>
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<td>28.2</td>
</tr>
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<td>F</td>
<td>31</td>
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<td>56</td>
<td>21.5</td>
</tr>
<tr>
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<td>M</td>
<td>22</td>
<td>178</td>
<td>64</td>
<td>26.0</td>
</tr>
</tbody>
</table>

3.2.2 3D-Kinematic Motion Capture Data Collection

A six-camera motion capture system (2 Eagle/4 Hawk HiRes cameras, Cortex 2.6 system, Motion Analysis Corp., Santa Rosa, CA, USA) was used for the 3D analysis of gait. Prior to each testing day, the motion capture system was calibrated using first an L-Type calibration unit and second, a calibration wand. Subjects were set up with 10 single passive reflective markers placed on the left knee, right knee, left shank, right shank, left heel, right heel, left medial malleolus, left lateral malleolus, right lateral malleolus, and right foot, as well as five clusters of three passive reflective markers placed on the surface of the left foot. The marker clusters were taped to the foot in order to reduce possible falling or moving of the clusters (Figure 3-1). Data was collected in Cortex 2.6 at a collection rate of 60Hz (Appendix 1). Prior to static and dynamic data collection, motion capture trials were collected for each bony landmark defined in Table 1-1 (p.30) in the Introduction. Bony landmarks were palpated at the surface of the skin and a wand with three markers was pressed into the landmark position.
Figure 3-1: Location of marker clusters used with the MSFM. Marker clusters are located on the interphalangeal joint of the hallux, mid-shaft of first and fifth metatarsals, dorsal to navicular tuberosity, and lateral to Achilles tendon.

3.2.3 Fluoroscopic Data Collection

The fluoroscopes (SIREMOBIL Compact (L); Siemens Medical Solutions USA Inc., Malvern, PA, USA) were placed in such a way that each device would have the same view of the foot, but from a different angle. They were placed in such a way that one image intensifier would capture a sagittal, lateral view of the foot and one image would capture an oblique anterior/posterior view of the foot. The image intensifier for the latter
position was placed under the platform so as to capture a ‘top view’ of the foot while the subject was walking. Once the fluoroscopes were in their required positions, they were calibrated by imaging a calibration box designed by Kedgley (2009c) with beads embedded in known locations. The box was placed such that each fluoroscope viewed a fiducial and control plane. Following testing, the calibration box was imaged again with the addition of an image distortion grid to correct for pincushion distortion. Detailed instructions on how to use the fluoroscopes and calibrate them are found in Appendix 2. Custom written algorithms in MATLAB (The MathWorks, Natick, MA, USA) were used to finalize the calibration process and recreate the laboratory set up for data processing. Prior to testing, subjects wore a lead skirt, vest, and thyroid shield in order to protect themselves from the x-rays. An x-ray technician assisted in data collection by powering the fluoroscopes as the subjects were performing the task of interest, by aligning the subjects’ foot in the correct position and orientation so that it would fit the field of view of the image intensifier, and by changing the settings of the fluoroscope to get a good quality image. The video feed from the fluoroscopes was collected to an SD card using a dual channel video and audio recording system (Ultimate Digital Video Platform – Pro, datatoys.com, Milwaukee, WI, USA). This video system collected the video feed from each fluoroscope simultaneously, thus the frames aligned. Although, a second method was used to verify the proper alignment of the video feeds. A metal wire was placed in the field of view of both fluoroscopes and very quickly pulled out at the beginning of each trial. The disappearing of the metal wire from the fluoroscopic field of view was used to align the video feeds. Fluoroscopic data was collected at 29.97fps.
3.2.4 Testing Protocol

Subjects underwent both a 3D gait analysis and bi-planar fluoroscopic analysis simultaneously (Appendix 3). The subjects were required to stand or walk along a custom built wooden platform while fluoroscopic images and 3D kinematic data of their left foot were collected. First a static standing trial was collected. Subjects were required to stand still on the platform with their left foot in the field of view of the fluoroscopes while the laboratory x-ray technician took a still x-ray image (Figure 3-2). This process was repeated until one good image was taken of the hindfoot and the forefoot. Following the static trials, the subjects had to perform four dynamic trials for both the hindfoot and the forefoot. Subjects performed the trials barefoot.
Figure 3-2: Laboratory testing set-up. Subject walking along the custom built wooden platform while placing her foot in the field of view of the fluoroscopes.

3.2.5 Data Analysis

Four foot segments were identified and used for analysis; the hindfoot (calcaneus), midfoot (tarsals, cuneiform I, II, and III, navicular, and cuboid), medial forefoot (1st metatarsal), and lateral forefoot (5th metatarsal). Six motions of the foot can be reported using the MSFM: ankle joint (talus with respect to the lower leg), subtalar joint (midfoot
with respect to the talus), hindfoot segment motion (with respect to midfoot) in the frontal and transverse planes, forefoot segment motion (with respect to midfoot) and the height-to-length ratio of the MLA. Due to the limiting 9” field of view of the fluoroscopes, four motions could be analyzed in this project. Ankle joint and subtalar joint motions with the MSFM require shank and knee markers for their calculation. As the field of view of the fluoroscope is only 9”, solely the foot is visible, thus the ankle and subtalar joint motion as defined by the MSFM cannot be calculated. Therefore, the hindfoot segment motion (with respect to the midfoot) in the frontal and transverse plane, forefoot segment motion (with respect to midfoot) and height-to-length of the MLA were calculated in this chapter.

3.2.6 Data Analysis – Step 1 – Motion capture data

Motion capture data was processed in Cortex 2.6 and the 3D coordinates of each marker were exported in the form of a .trc file. Each bony landmark was exported into Excel as a .trc file and contained the marker coordinates for the marker clusters and wand markers. The .trc files were read into a custom written MATLAB algorithm and joint angle motion values were exported into an Excel document. The MATLAB algorithm calculates each joint motion angle during a dynamic trial as the variation from static standing. Motion capture data was filtered using a 4th order Butterworth smoothing zero-lag filter with a cut off of 6 Hz to remove vibration artefacts from the marker clusters.

3.2.7 Data Analysis – Step 2 – Fluoroscopic data

The frames of interest for each trial were exported using Adobe Premiere Pro CC (Adobe Systems Inc., San Jose, California, USA). For each dynamic trial, the first visible frame
at heel strike and the last visible frame at toe-off were selected as well as the middle frame (mid-stance) between these two points. Visible frames were defined as the first or last frame where the bones of interest were seen clearly enough for the matching process. Bone ridges and specific landmarks had to be well defined and not blurry to be considered visible. These frames were used to match with bone models in Rhinoceros. For each time point, frames from both fluoroscopes were imported into Rhinoceros twice. The image was matched the first time to the calcaneus, lateral malleolus, navicular, and first and fifth metatarsals. The second time, the image was matched to the talus, medial malleolus, first cuneiform, and cuboid. This separation of bones was chosen so that each bone was matched to the fluoroscopic image without having any overlapping bones hiding the image contours. Specific bone contours and landmarks were matched to each image consistently the same way. The 3D coordinates of the bony landmarks from Table 1-1 in the Introduction were exported using a custom written RhinoScript. Figures 3-3, 3-4, and 3-5 show the location and name of the bony landmarks.
Figure 3-3: Lateral view of the bones of the foot with red dots indicating exported bony landmarks.

Figure 3-4: Medial view of the bones of the foot with red dots indicating exported bony landmarks.
Files from the exported bony landmarks in the form of .trc were created and read by a custom written MATLAB algorithm in order to calculate joint motion angles.

3.2.8 Data Analysis – Step 3 – Alignment of Motion Capture and Fluoroscopic Data

The next step required motion capture and fluoroscopic data be aligned properly for comparison of joint motions. This was done using three steps. The first step was to determine the moment of heel strike in motion capture data. Displacement of the heel marker in the forward y direction was plotted. The minimum value of the curve was assumed to be the moment of heel-strike. The second step was to look at the fluoroscopic output frames and approximate the frame of heel-strike. This part had to be
approximated since the collection frame rate of fluoroscopes is only 30 fps. Any fast movement becomes blurry as during heel-strike and toe-off if the subject is walking at a faster pace. The number of frames between the approximated heel-strike and first clear fluoroscopic image was counted and multiplied by two since the collection rate of motion capture was double that of fluoroscopy. This number was added to the heel strike frame number determined from the y-motion graph of the heel marker to estimate the frame that corresponded to the first matched fluoroscopic image. The third step was to use the output values and graphs from each method and make sure that either the values aligned or the shape of the curves aligned. Once the fluoroscopic data was aligned with the motion capture data the curves were normalized to 100 points to offset the difference in speed of each subject.

3.2.9 Data Analysis – Step 4 – Statistics

The first, middle, and last 5% of the frames for the stance phase of each motion capture trial was averaged and used as a comparison to the first, middle, and last stance phase value from the fluoroscopic data. These values were averaged for each joint motion and an independent samples T-test was run using SPSS (IBM Corporation, Armonk, New York, USA) on the heel-strike, mid-stance, and toe-off values between motion capture and fluoroscopic data to test for statistical difference. Statistical significance was set at \( p < 0.05 \). Levene’s test for equality of variances was computed first via SPSS first. When a variable did not show a significance level greater than 0.05, the values corresponding to ‘equal variances not assumed’ were selected for the T-test. The SPSS output data sheet is presented in Appendix 5.
3.3 Results

Table 3-2 indicates the range of change in degrees for hindfoot supination/pronation motion, hindfoot internal/external rotation motion, and forefoot motion as well as the maximum and minimum degree value averaged over all subjects. The MLA height-to-length ratio is shown as a range deviating from normalized quiet standing at 0, with its respective maximum and minimum values. The greatest angle motion was seen in the forefoot motion. The range was 56.51° and 10.18° for motion capture and fluoroscopy respectively. The large range for forefoot motion with motion capture was due to one subject that had results that were much higher than the others. Motion capture values overestimate the range when compared to fluoroscopy.

Table 3-2: Average of joint motion values over all trials. The range, maximum, and minimum values are given in degrees, except for the MLA, for motion capture and fluoroscopy data. MLA height-to-length ratio is given as a range deviating from 0.

<table>
<thead>
<tr>
<th>Joint Motion (°)</th>
<th>Range</th>
<th>Maximum-minimum</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Motion capture</td>
<td>Fluoroscopy</td>
</tr>
<tr>
<td>Hindfoot</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Supination/pronation</td>
<td>15.80</td>
<td>8.05</td>
</tr>
<tr>
<td></td>
<td>(2.43, -13.37)</td>
<td>(3.86, -4.19)</td>
</tr>
<tr>
<td>Internal/external rotation</td>
<td>13.16</td>
<td>8.70</td>
</tr>
<tr>
<td></td>
<td>(5.93, -7.23)</td>
<td>(4.57, -4.13)</td>
</tr>
<tr>
<td>Forefoot</td>
<td>56.51</td>
<td>10.18</td>
</tr>
<tr>
<td></td>
<td>(34.56, -21.95)</td>
<td>(12.07, 1.89)</td>
</tr>
<tr>
<td>MLA height-to-length ratio</td>
<td>0.12</td>
<td>0.02</td>
</tr>
<tr>
<td></td>
<td>(0.09, -0.03)</td>
<td>(0.00, -0.02)</td>
</tr>
</tbody>
</table>
The next set of graphs from Figure 3-6 to 3-13 show joint motion curves. Each curve is scaled to 100% of weight-baring stance phase with 0 at heel-strike and 100 at toe-off. Each joint motion has two figures associated with it. The first figure plots all the individual trials as thin lines and a dotted thick line indicates the average over all 10 trials for motion capture and fluoroscopy. The second figure plots the average motion for each joint as a thick line and the thin line indicates one standard deviation above and below the mean. Figures 3-6 and 3-7 display supination/pronation motion of the hindfoot, a positive increase in angle indicates supination motion. At heel-strike, both techniques showed a pronated position of the hindfoot, when approaching mid-stance, the position of the hindfoot approached 0 and as the subject approached toe-off, the hindfoot showed an increase in supination. Figures 3-8 and 3-9 show internal/external rotation motion of the hindfoot. A positive increase in angle is associated with internal rotation. At heel-strike, the hindfoot was in internal rotation, which kept on decreasing towards a neutral standing position during mid-stance. As the heel started lifting off the ground and approaching toe-off, the hindfoot showed a trend towards increased external rotation. Figures 3-10 and 3-11 display supination/pronation angle of the forefoot. An increase in angle is associated with supination of the forefoot. At heel-strike, the forefoot was in a supinated position compared to neutral standing. As the stance phase progressed, the angle approached 0 and as the heel lifted and got closer to toe-off, the forefoot showed a pattern towards pronation. Finally Figures 3-12 and 3-13 show the MLA height-to-length ratio. An increase in the ratio indicates a rise in the MLA. At heel-strike, the MLA showed a slight drop compared to quiet standing, meaning the arch was flattening out. As the stance phase progressed, the arch approached its shape during quiet standing and
continued to rise as it approached toe-off. The data from motion capture emphasizes this movement more than fluoroscopic data. The shapes of the curves indicate more variability among the motion capture data collection technique. Although, the shapes are very similar for both collection methods and the fluoroscopic data falls within the standard deviation curves of the motion capture plots.

Figure 3-6: Hindfoot supination/pronation motion normalized to 0 at quiet standing. Dashed lines are all the trials calculated using motion-capture and solid lines are trials calculated using fluoroscopy. Thick dashed and solid lines represent the average of all trials for motion capture and fluoroscopy, respectively. Trials were normalized to percentage of stance phase visible using fluoroscopy, 0 representing heel strike and 100, toe-off.
Figure 3-7: Hindfoot supination/pronation motion normalized to 0 at quiet standing. The thick dashed line represents the average of all trials and the thin dashed lines represent ±1SD for motion capture. The thick solid line represents the average of all trials and the thin solid lines represent ±1SD for fluoroscopy.

Figure 3-8: Hindfoot internal/external rotation motion normalized to 0 at quiet standing. Dashed lines are all the trials calculated using motion-capture and solid lines are trials calculated using fluoroscopy. Thick dashed and solid lines represent the average of all trials for motion capture and fluoroscopy, respectively. Trials were normalized to percentage of stance phase visible using fluoroscopy, 0 representing heel strike and 100, toe-off.
Figure 3-9: Hindfoot internal/external rotation motion normalized to 0 at quiet standing. The thick dashed line represents the average of all trials and the thin dashed lines represent ±1SD for motion capture. The thick solid line represents the average of all trials and the thin solid lines represent ±1SD for fluoroscopy.

Figure 3-10: Forefoot angle motion normalized to 0 at quiet standing. Dashed lines are all the trials calculated using motion-capture and solid lines are trials calculated using fluoroscopy. Thick dashed and solid lines represent the average of all trials for motion capture and fluoroscopy, respectively. Trials were normalized to percentage of stance phase visible using fluoroscopy, 0 representing heel strike and 100, toe-off.
Figure 3-11: Forefoot angle motion normalized to 0 at quiet standing. The thick dashed line represents the average of all trials and the thin dashed lines represent ±1SD for motion capture. The thick solid line represents the average of all trials and the thin solid lines represent ±1SD for fluoroscopy.

Figure 3-12: MLA arch ratio normalized to 0 at quiet standing. Dashed lines are all the trials calculated using motion capture and solid lines are trials calculated using fluoroscopy. Thick dashed and solid lines represent the average of all trials for motion capture and fluoroscopy, respectively. Trials were normalized to percentage of stance phase visible using fluoroscopy, 0 representing heel strike and 100, toe-off.
Table 3-3 presents the average angle value for heel-strike, mid-stance, and toe-off for motion capture and fluoroscopy for each foot joint motion examined. Angle values during mid-stance were the most similar when comparing motion capture and fluoroscopy. A significance asterisk (*) indicates a significant different between motion capture and fluoroscopy values. The level of significance was set to $p < 0.05$. From this table, the only significant difference between motion capture and fluoroscopy was found at the level of the MLA height-to-length ratio for toe-off.
Table 3.3: Comparison of motion capture and fluoroscopy joint angle during heel strike, mid stance, and toe off. (*) indicates a significant difference between motion capture and fluoroscopy. Significance level was set at $p < 0.05$.

<table>
<thead>
<tr>
<th></th>
<th>Heel Strike</th>
<th>Mid Stance</th>
<th>Toe Off</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Motion Capture</td>
<td>Fluoroscopy</td>
<td>Motion Capture</td>
</tr>
<tr>
<td>Hindfoot</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Supination/pronation (*)</td>
<td>-8.10</td>
<td>-3.66</td>
<td>-4.37</td>
</tr>
<tr>
<td>Internal/external rotation (*)</td>
<td>2.48</td>
<td>3.82</td>
<td>-0.23</td>
</tr>
<tr>
<td>Forefoot (*)</td>
<td>19.41</td>
<td>11.73</td>
<td>1.45</td>
</tr>
<tr>
<td>MLA height-to-length ratio</td>
<td>-0.01</td>
<td>-0.01</td>
<td>0.02</td>
</tr>
</tbody>
</table>
The following four figures, Figure 3-14 to Figure 3-17 display the difference for each trial between motion capture and fluoroscopy. The order of foot joint motions is hindfoot supination/pronation (Figure 3-14), hindfoot internal/external rotation (Figure 3-15), forefoot supination/pronation (Figure 3-16), and MLA height-to-length ratio (Figure 3-17). Forefoot supination/pronation graph shows the greatest spread of difference values between both measurement techniques. Hindfoot motion has a much smaller difference range, with hindfoot supination/pronation motions showing a greater range than hindfoot internal/external rotation.
Figure 3-14: Hindfoot Supination/Pronation degree difference between fluoroscopy and motion capture for both trials from all five subjects. Black cross indicates the average angle difference between both measurement techniques.
Figure 3-15: Hindfoot internal/external rotation degree difference between fluoroscopy and motion capture for both trials from all five subjects. Black cross indicates the average angle difference between both measurement techniques.
Figure 3-16: Forefoot supination/pronation degree difference between fluoroscopy and motion capture for both trials from all five subjects. Black cross indicates the average angle difference between both measurement techniques.
Figure 3-17: MLA height-to-length ratio value difference between fluoroscopy and motion capture for both trials from all five subjects. Black cross indicates the average angle difference between both measurement techniques.

Table 3-4 displays the absolute difference in angle values between motion capture and fluoroscopy for all foot joint motions. For supination/pronation of the hindfoot, the smallest difference between both motion capture techniques was seen at heel-strike. For internal/external rotation of the hindfoot, the smallest difference was seen at mid-stance. For forefoot motion, the smallest difference was also seen at mid-stance. For the MLA height-to-length ratio at heel-strike and mid-stance, there was no significant difference seen between the value obtained from motion capture and fluoroscopy.
Table 3-4: Absolute difference in foot joint motions between motion capture and fluoroscopy.

<table>
<thead>
<tr>
<th></th>
<th>Heel Strike</th>
<th>Mid Stance</th>
<th>Toe Off</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hindfoot</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Supination/pronation (°)</td>
<td>4.45</td>
<td>5.12</td>
<td>6.70</td>
</tr>
<tr>
<td>Internal/external rotation (°)</td>
<td>1.33</td>
<td>0.45</td>
<td>0.95</td>
</tr>
<tr>
<td>Forefoot (°)</td>
<td>7.68</td>
<td>1.81</td>
<td>13.63</td>
</tr>
<tr>
<td>MLA height-to-length ratio</td>
<td>0.00</td>
<td>0.02</td>
<td>0.05</td>
</tr>
</tbody>
</table>

The following 4 figures present the average joint motion at heel-strike, mid-stance, and toe-off for both motion capture and fluoroscopy. An error bar of one standard error above and below the mean was plotted. A significant difference between the testing methods is indicated by an asterisk (*) above the bar graph.
Figure 3-18: Average supination/pronation motion of the hindfoot at heel-strike, mid-stance, and toe-off for motion capture and fluoroscopy in stripped and solid grey respectively. One standard error above and below the mean is indicated by error bars.

Figure 3-19: Average internal/external rotation of the hindfoot at heel strike, mid stance, and toe off for motion capture and fluoroscopy in stripped and solid grey respectively. One standard error above and below the mean is indicated by error bars.
Figure 3-20: Forefoot angle motion at heel strike, mid stance, and toe off for motion capture and fluoroscopy in stripped and solid grey respectively. One standard error above and below the mean is indicated by error bars.

Figure 3-21: Average MLA height-to-length ratio at heel strike, mid stance, and toe off for motion capture and fluoroscopy in stripped and solid grey respectively. Error bars indicate one standard error above and below the mean. A significant difference between motion capture and fluoroscopy is indicated by an asterisk (*) with $p < 0.05$. 
3.4 Discussion

The purpose of this study was to validate the MSFM developed by Jenkyn and Nicol (2007) used with motion capture against the gold standard of bi-planar fluoroscopy. Quantifying errors in motion capture analysis hadn’t previously been done for this multi-segment foot model while using another 3D motion capture system. Four of the six motions of the model were analyzed because the field of view of the fluoroscopes does not allow for calculation of the ankle and subtalar joints in the manner the MSFM does. Thus, the hindfoot supination/pronation and internal/external rotation motion, forefoot supination/pronation, and MLA height-to-length ratio were examined. It was hypothesized that motion capture would show differences of less than 1° when compared to fluoroscopy. The hypothesis was supported for hindfoot internal/external rotation during mid-stance and toe-off. Hindfoot and forefoot supination/pronation motions showed greater differences than 1° although there was no significant difference found between motion capture and fluoroscopy. The only significant difference found between motion capture and fluoroscopy was at the level of the MLA height-to-length ratio during toe-off.

The results gathered with motion capture were comparable to those found in literature with previous use of this MSFM. Hindfoot motion curves show the same pattern when comparing results from Jenkyn and Nicol (2007). The section of the gait cycle prior to toe-off for supination/pronation of the hindfoot showed an increasing pattern towards greater supination of the hindfoot (Figure 3-22). Internal/external rotation of the hindfoot pattern was also the same, showing a decreasing slope, thus towards greater external rotation as the foot approaches toe-off (Figure 3-23). The MLA height-to-length ratio
curves both show and increasing slope prior to toe off, indicating a rising arch. Although, results from Jenkyn and Nicol (2007) displayed a greater amplitude of motion (Figure 3-24). On the other hand, the pattern of the curve from this research for the motion of the forefoot did not correspond with the results from Jenkyn and Nicol (2007). This study showed a curve with a decreasing pattern, which is equivalent to increased pronation of the forefoot during the stance phase of walking. Moreover, the results from this study show a much greater amplitude of movement that Jenkyn and Nicol (2007) (Figure 3-25).

Figure 3-22: Comparison of hindfoot supination/pronation motion between this study and Jenkyn and Nicol (2007) study. Solid black line indicates the results from Jenkyn and Nicol (2007) and dashed black line indicates the results from this study.
Figure 3-23: Comparison of hindfoot internal/external rotation motion between this study and Jenkyn and Nicol (2007) study. Solid black line indicates the results from Jenkyn and Nicol (2007) and dashed black line indicates the results from this study.

Figure 3-24: Comparison of MLA height-to-length ratio between this study and Jenkyn and Nicol (2007) study. Solid black line indicates the results from Jenkyn and Nicol (2007) and dashed black line indicates the results from this study.
As for fluoroscopic data, a study conducted by Arndt et al. (2007) calculated bone joint kinematics within the foot using marker-based fluoroscopic RSA. Individual joint motions were reported in their study. The mean frontal plane range of motion for the calcaneus with respect to the talus was 8.9°. This joint motion is a good approximation of the supination/pronation movement of the hindfoot tested in this study, which had a mean range of motion of 8.05° (Arndt et al., 2007). As the MSFM used in this study had not been used with fluoroscopic RSA in research before, comparison of joint motions with literature values did not prove to be possible.

Over all the four joint motions, the MLA height-to-length ratio values showed the smallest absolute difference between motion-capture and fluoroscopy. The differences found were 0.00, 0.02, and 0.05 respectively for heel-strike, mid-stance, and toe-off.
These differences show that motion capture, when looking at the deformation of the MLA, is a good technique and can be used in a clinical setting for valid results. The MLA is an important foot motion and is of interest to clinicians as it is related to injury patterns. High arched individuals often display a more rigid arch, which leads to a reduced shock absorption capacity. Understanding the mobility of the MLA during walking or running is an essential component in further understanding and preventing injuries. Williams et al. (2014) found that high arched individuals with more mobile arches displayed smaller initial loading forces as well as lower forces at the second vertical ground reaction force peak during the loading phase. This second peak in loading force may be attenuated by the compliance of the arch (Williams, Tierney, & Butler, 2014). As for flat-footed individuals, problems with shock absorption, pressure distribution, and weight transfer have been observed. Ground reaction forces and repetitive loading may induce injuries to the lower extremities of the body. Chang et al. (2012) found that individuals with flat feet had increased ground reaction peaks during two-feet drop landings when compared to normal foot individuals. They also found that flat-footed individuals had great muscular activation of the vastus lateralis and tibialis anterior muscles compared to individuals with normal feet. This altered change in muscle activation over time may lead to different lower extremity injuries (Chang, Kwon, Kim, Ahn, & Park, 2012). Thus, quantifying the MLA height-to-length ratio to categorize the patient’s foot type, as well as quantifying a patient’s arch mobility, would allow for knowledge about what type of injuries and increased loading or muscle activation patterns the patient could be subjected to during their activities of daily living.
Hindfoot internal/external rotation was the next joint motion, which overall, displayed the smallest difference between motion capture and fluoroscopy. Differences were 1.33°, 0.45°, and 0.95° for heel-strike, mid-stance, and toe-off respectively. The greatest difference in angle might be seen during the heel-strike phase because the calcaneus contacts the ground first and the muscular and tendinous structures are the ones absorbing all the force and contact load. As the foot contacts the ground, the stabilizing muscles surrounding the calcaneus and ankle complex are the first ones activated. The increased muscle activation at this joint may cause a shift of the skin, which produces a change in the placement of the markers with respect to the underlying bone. The increased movement of the markers would explain the greater angle difference between motion-capture and fluoroscopy data. The calcaneal marker was placed on the lateral side of the hindfoot, close to the synovial sheaths of the peroneus longus and brevis muscles. These tendons are displaced during the stance phase in order to depress the first metatarsal head and elevate the fifth metatarsal base (Nordin & Frankel, 2012). Moreover, this joint motion showed the least variability among subjects. Figure 3-8 illustrates that the joint motion curves show the same overall pattern for motion capture and fluoroscopy.

The hindfoot supination/pronation motion of the foot had slightly greater angle differences when comparing motion capture and fluoroscopy. The differences were 4.45°, 5.12°, and 6.70° for heel-strike, mid-stance, and toe-off respectively. This motion also showed greater variability with a range of 15.80° for motion-capture and 8.05° for fluoroscopy. In this case, the greatest difference was displayed during toe-off. Consistent with published literature, this frontal plane movement of the hindfoot displayed greater STA error than hindfoot motion in the transverse plane. Several studies
have found that frontal plane markers and bone markers had poor agreement. Reinschmidt et al. (1997) found average errors relative to the range of motion of 70% for abduction/adduction motion of the knee during running (Reinschmidt et al., 1997). Similarly, Nester et al. (2007) found significant differences for tibial kinematics (tibial/calcaneal motion) between bone markers, skin markers, and plate-mounted markers of magnitudes of 3.6° for frontal plane motion. This was the greatest difference compared to sagittal and transverse plane motion, which were 2.6° and 2.3° respectively (Nester et al., 2007).

The joint motion that showed the greatest variability and difference between motion capture and fluoroscopy was supination/pronation angle of the forefoot. The differences seen between motion capture and fluoroscopy were 7.68°, 1.81°, and 13.63° for heel-strike, mid-stance, and toe-off respectively. The greatest difference was seen during toe-off since that is the moment where there is increased muscle activation in the lower leg. The supination/pronation angle was created by the projection of a vector, from the first and fifth metatarsal heads, onto the midfoot frontal plane axis. An increase in angle represented supination of the forefoot. To create the projected vector, the bony landmarks of the metatarsals are calculated by taking into account the location of the markers from the first and fifth metatarsal clusters. As the foot is approaching the toe-off phase, the muscles of the foot as well as the leg muscles will contract, thus causing the internal structures of the foot to change shape, which causes the skin to move on the surface of the foot. A muscle that produces this movement of the skin is the tibialis anterior muscle, which is responsible for assuring the clearing of the foot during toe-off. It concentrically contracts, which moves the tendon attached at the first cuneiform, where
the medial forefoot marker is placed (Perry & Burnfield, 2010). Another muscle that might affect skin movement during toe-off is the adductor hallux, as it acts as a forefoot stabilizer during this moment of stance (Nordin & Frankel, 2012). The variability between individuals for this joint motion was the greatest. The range was 56.51° for motion capture as opposed to 10.18° for fluoroscopy. This is a very big difference between the two types of motion analysis. One difficulty with this multi-segment foot model is the number of markers that are placed on the foot. Five clusters of three markers are placed on the foot with three of the clusters located on the medial side. This placement of markers causes problems when the subject has small feet, since that causes the markers to be in close proximity to each other. When the markers are closer together, there are more missing markers during the trials. Missing markers have to be created as virtual markers, thus their position in space is estimated and may not be their real position. As the placement of the clusters on the medial side is rather close, during dynamic movement of walking or running, the markers from one cluster often cross over with the markers of another cluster. Correcting this cross over of markers proves to be rather difficult at times during post-processing. Often, this crossing-over causes the marker trajectories to be altered; seen as a ‘flickering’ or uneven movement of the marker. Missing markers must also be created virtually, which estimates their actual position in space.

This specific marker set had previously been examined for translational errors from the markers compared to the underlying bone by Shultz et al. (2011). The hindfoot and navicular cluster were examined during heel-strike, mid-stance, and toe-off. Their results showed an increase in translational errors for toe-off compared to the other two phases of
stance. The hindfoot cluster had an origin displacement of 5.9mm, 6.5mm, and 12.1mm for heel-strike, mid-stance, and toe-off respectively. The navicular cluster displayed even greater origin displacement with errors of 7.6mm, 7.6mm, and 16.4mm for heel-strike, mid-stance, and toe-off respectively. These results are consistent with the ones found in this study. Both forefoot and hindfoot supination/pronation motion displayed the greatest difference in angle compared to fluoroscopy for the toe-off phase. The authors believe that this increase in error at toe-off is due to the increased muscular activation patterns needed to lift the foot off the ground and into the swing phase (Shultz, Kedgley, & Jenkyn, 2011b).

Several other studies have quantified skin motion error when using single markers, or marker clusters on the foot. Birch and Deschamps (2011) attempted to quantify skin motion artefact on one subject by placing skin markers on the lateral and medial malleoli as well as the head of the talus. They compared the location of the markers to radiographic images and found movement of 0.61mm to 22.18mm displacement of the markers. Angular movement errors were smaller, 1° in the sagittal and transverse planes and 5° in the frontal plane. The errors were also smaller for the markers placed on the malleoli as opposed to the talar head (Birch & Deschamps, 2011).

Another study by Nester et al. (2007) compared walking trials and looked at kinematic outcomes using bone pins, skin mounted markers, and 3-marker clusters. Although they performed each condition as a different trial, they found that there was no difference between the results from skin markers and 3-marker clusters, but found that skin mounted markers underestimated the range of motion in the frontal and sagittal planes. Moreover, the maximum differences between the bone and skin markers were the greatest. They
found no systematic or consistent differences between the skin markers or 3-marker cluster and bone pins. There was no clear pattern of over or underestimation (Nester et al., 2007). This finding is consistent with what was found in this project. When looking at Figures 3-14 to 3-17, there is no consistent pattern observed for the different moments of stance phase. Only the MLA height-to-length ratio showed a consistent overestimation of movement compared to fluoroscopy. Similarly, following the trend found by Nester et al. (2007) as well as this study, Westblad et al. (2002) found no consistent pattern of over- or underestimation of skin marker relative to bone anchored markers at the ankle-joint complex when examining the motion of the heel with respect to the lower leg looking at inversion/eversion, abduction/adduction, and plantar/dorsiflexion (Westblad et al., 2002).

This thesis study proved to partially support the starting hypothesis that the MSFM would be able to estimate joint motions while used with motion-capture when compared to bi-planar fluoroscopy. Although not all joint motions had a difference of less than 1°, no significant differences were found between the two techniques for three of the foot joint motions, only the MLA height-to-length ratio toe-off showed a significant difference. Although, there were several limitations to this study that have to be considered. The small sample size is one of the major restrictions. Another improvement that would benefit this project would be to find a way to collect motion capture and fluoroscopy data simultaneously within the same program as opposed to having two different types of hardware and software collecting data. Future studies should also look at a population with different type feet such as flat foot, or high-arched as these populations often show greater signs of risk of injury. Following this thought, validating the multi-segment foot
model during shod walking and running should be done as well. Shoes may restrict the movement or placement of marker clusters because the structures of certain shoes may be obstructive. Moreover, a shoe might also restrict the movement of the bones. During running, marker motion is increased and greater muscle activation is occurring because of increased loads during landing and stance phase.

3.5 Conclusion

In conclusion, this study attempted to validate a MSFM that uses retroreflective markers and optical motion capture against bi-planar RSA fluoroscopy. The inversion/eversion and supination/pronation of the hindfoot with respect with the midfoot and forefoot supination/pronation with respect to the midfoot motions were analyzed and were not significantly different. The MLA height-to-length ratio displayed only a significant difference for the toe-off phase. There was a trend to a greater difference between motion-capture and fluoroscopy during the heel-strike and toe-off phases. This is suspected to be because of the greater muscular contraction for stabilization and push-off needed during these phases, thus increasing the motion of the surface mounted markers. None the less, the lack of significant differences would allow for the use of the MSFM in clinical settings with appropriate accuracy, thus decrease radiation exposure to the patient as well as decreasing post-processing time.
Chapter 4 - General Discussion and Conclusions

4.1 Summary

The objective of this study was to validate the MSFM created by Jenkyn and Nicol (2007) against the bi-planar model-based fluoroscopy system, which was validated in the Wolf Orthopaedic Quantitative Imaging Laboratory (WOQIL) at Western University by Anne-Marie Allen (2009). The model separates the foot into four sections; the hindfoot (calcaneus), midfoot (tarsals: cuneiforms I, II, & III, navicular, cuboid), medial forefoot (first metatarsal), lateral forefoot (fifth metatarsal). This separation of the forefoot into medial and lateral parts is a novel definition specific to this model. The model requires five retroreflective 3-marker marker clusters be placed on the foot. The medial foot clusters are placed on the first proximal phalangeal, mid-shaft of the first metatarsal, and navicular. The lateral foot clusters are placed on the mid-shaft of the fifth metatarsal and lateral to the Achilles tendon on the calcaneus. Single markers are placed on the medial and lateral malleoli of the left foot. The model allows for the calculation of six joint motions. The first motion is ankle joint motion defined as rotation of the talus with respect to the lower leg. This motion defines dorsi/plantarflexion of the ankle joint. Second joint motion is subtalar joint motion. It is defined as midfoot rotation with respect to the talus as inversion/eversion of the midfoot. The third and fourth motions are the hindfoot with respect to the midfoot defined as supination/pronation and internal/external rotation. The fifth motion is movement of the forefoot with respect to the midfoot, defined as a compound twisting of both the lateral and medial segments of the forefoot describing supination/pronation motion. Finally, the last motion defined by
The MSFM is the change in height-to-length ratio of the MLA (Jenkyn & Nicol, 2007). Being able to study motions of the foot in this manner is a great strength compared to how the foot is analyzed during traditional clinical gait analysis (usually the foot is described as a rigid segment). Although, a downfall of this technique is the use of markers placed on the surface of the skin. The markers can be displaced during gait as the structures of the foot deform with muscle contractions. This deformation is known as soft-tissue artefact and is a primary source of error when conducting optical motion capture.

Therefore, this MSFM was validated against bi-planar RSA fluoroscopy, which is considered the gold standard. The bi-planar fluoroscopy system in the laboratory consists of two C-arm fluoroscopes positioned at about 120° with respect to each other. A custom built wooden platform is placed above one image intensifier and along side the other to allow for subjects to have their left foot imaged while walking. Calibration of the fluoroscopic system using a cube calibration box and a pincushion distortion grid was developed by Kedgley (2009). Custom MATLAB scripts are used for calibrating the system and image digitization. The remainder of the calibration process and laboratory set-up recreation in Rhinoceros was developed by Allen (2009) in our laboratory. Once the laboratory set-up is recreated, bone models are produced in a DICOM viewer OsiriX. Bones used by the model are segmented individually and bony landmarks of interest are identified on the bone itself. 3D coordinates of the bony landmarks are exported using a custom written RhinoScript (Allen, 2009). A custom written MATLAB script modified by Shultz (2006) was used to calculate joint motions from landmark coordinates.
The first part of the project required the validation of generic bone models, which was the bulk of Chapter 2, in order to proceed with their use in the validation of the MSFM. Markerless bi-planar fluoroscopy requires some type of bone model to be used for matching with fluoroscopic images. Using generic bone models decreases the amount of radiation a subject is exposed to and decreases post-processing times. Thus, Chapter 2 validated the use of generic bone models as opposed to subject-specific bone models for evaluation of the MLA arch with bi-planar fluoroscopy. From previous research studies, our laboratory created a ‘bank’ of bone models from subjects’ CT scans. From that bank of bone models, 5 subjects were selected to which 4 different bone models were matched according to foot type and size. The generic bone models were: 1) same size foot and same type foot (G_SS_ST), 2) same size foot and different types (G_SS_DT1, G_SS_DT2), and 3) different size and same type foot (G_DS_ST). Subjects were required to stand on a wooden platform while their left foot was imaged. MLA angle was calculated using the medial process of the calcaneus, navicular tuberosity, and distal head of the first metatarsal. There was no clear over- or underestimation trend observed from the generic bone models with respect to the subject-specific bone models. There was a lot of variability observed among all conditions, but overall, the smallest average difference from the subject-specific bone model was the G_SS_ST model with an average difference of $2.3^\circ \pm 0.7^\circ$. This was a small enough difference to use generic bone models instead of subject-specific bone models. For future research, this technique will decrease the amount of radiation a subject is exposed to and there will be no more need for individual segmentation of bones for each subject. The objectives for this study were achieved and the hypothesis was supported. Generic bone models of the same size
and same type foot were able to estimate the MLA angle by less than 5° when compared to subject-specific bone models.

The positive results from the generic bone model validation study allowed us to eliminate CT scans of the subjects used for the study in Chapter 3. This made it an easier procedure for the subjects as they only had to come to one testing session and they did not need to be exposed to any extra radiation. Eliminating CT scans also made post-processing much faster as the bones from the ‘bank’ were already segmented.

The second part, Chapter 3, validated the MSFM with bi-planar fluoroscopy. Four joint motion of the MSFM were analyzed since the field of view of the fluoroscopes restrict the view of the tibia, fibula, and knee needed for the calculations of the ankle and subtalar joint motions. Thus, the pronation/supination and internal/external rotation of the hindfoot, supination/pronation of the forefoot, and MLA height-to-length ratio were examined. Five subjects walked on the custom built platform while their left foot was being imaged by the fluoroscopes and kinematic data via Cortex was collected. Overall, graphs for motion capture and fluoroscopy followed the same trend. Angle values for heel-strike, mid-stance, and toe-off were tested for significance with p set to < 0.05. Hindfoot and forefoot motions were not significantly between motion capture and fluoroscopy data. MLA height-to-length ratio was significantly different for toe-off phase and not significantly different for heel-strike and mid-stance. For most foot motions, heel-strike and toe-off showed the most difference between the two techniques. This may occur since muscle activation is increased and there is greater movement of foot structures during these phases. Moreover, frontal plane motions such as supination/pronation of the hindfoot and forefoot showed greater variability between
motion capture and fluoroscopy compared to hindfoot internal/external rotation, which is a motion in the transverse plane. The hypothesis for this study was partially supported. The hypothesis initially stated that the MSFM would present a difference of less than $1^\circ$ when compared to fluoroscopy for the joint motions. This was only supported by hindfoot internal/external rotation motion during mid-stance and toe-off. All other joint motions presented a difference of greater than $1^\circ$, although there was no statistically significant differences between the two methods, which indicates positive results for the use of motion capture during gait.

For both projects, the selected accuracy of joint motion for the hypothesis, $5^\circ$ and $1^\circ$ for chapters one and two respectively, was deemed good enough. Podiatric decisions are often not made under $10^\circ$, as explains podiatrist Colin Dombrosky from SoleScience (SoleScience Inc., London, Ontario, Canada). Moreover, from literature, $5^\circ$ is cited as the clinical sufficient value (Nester et al., 2007). Nester et al. (2007), compared skin and plate mounted markers with bone fixed markers, which would be considered the gold standard. They found that the average maximum difference between two of the three protocols during stance was greater than $3^\circ$ in $100\%$ of the data and greater than $5^\circ$ in $73\%$ of the data. Thus, overall, results of foot joint motions using the MSFM when compared to fluoroscopy that are under $5^\circ$ would be considered accurate enough to make the model a viable tool in a clinical setting (Nester et al., 2007).

4.2 Strengths

This study assessed the accuracy of the MSFM in a research setting. As no significant difference was found between motion capture data and fluoroscopy for all foot joint motions but one, this model is seen as fit to be used in future research or clinical settings.
Fluoroscopic data was collected using a markerless RSA system, thus the technique was less invasive than a marker-based fluoroscopic technique. Tantalum beads were not required and generic bone models were used instead. This leads to another strength brought forth by the research from Chapter 2, validating generic bone models for RSA fluoroscopy allowed for them to be used for the study in Chapter 3. Generic bone models allowed for the elimination of CT scans of subjects prior to testing. This reduced the amount of radiation subjects were exposed to. Individual bones didn’t have to be segmented for each subject, which sped up post-processing time. Moreover, model-based RSA fluoroscopy allowed for direct visualization of the bones of interest, thus eliminating any STA that could introduce error into kinematic analysis. Landmarks used to calculate foot joint motions with the MSFM were placed directly on the bone.

4.3 Limitations

This study also had several limitations. The small sample size for both Chapter 2 and 3 projects was one of the biggest limitations. A greater sample size in Chapter 2 would of allowed to have several subjects within each foot type group. This would allow for averages to be calculated and statistics of comparison of means to be done. A separation of different foot types would also allow for better conclusions to be drawn in regards to whether generic bone models estimate foot angles differently for different type feet. A greater sample size in Chapter 3 would make averages more meaningful. When studying foot motions, there is a lot of variability between subjects, but even within one subject. From one step to the next, when looking at the same subject, joints can show different motion patterns.
An additional limitation to this study was the post-processing time. Bones have to be segmented manually and individually for each subject in OsiriX. The fluoroscopic laboratory set-up recreation is done by manually digitizing the calibration box frames and pincushion distortion grid frames. Matching of the bone models to the fluoroscopic images is also done manually for each bone. This increased time of data processing prevents the testing of more subjects within a smaller time frame.

The small fluoroscopic field of view due to the size of the image intensifier didn’t allow the calculation of two joint motions as defined by the MSFM. The field of view only allows for the foot to be imaged and doesn’t allow for the lower leg to be included in the analysis. The restricted size made it difficult to find appropriate subjects for comparison between motion capture and fluoroscopy. Smaller feet are better for fluoroscopic analysis since the whole foot can be imaged at once and the edge of both the calcaneus and first metatarsal can be seen. Unfortunately, smaller feet are a detriment when it comes to using motion capture with retroreflective markers. The smaller the feet are, the closer the markers are to each other, which causes more markers to be obscured. Bigger feet are harder to match with bone models when using fluoroscopy since the whole foot often doesn’t fit into the field of view, although they are much easier to image using motion capture.

4.4 Recommendations and Future Directions

Certain recommendations can be made for future studies in this area:

- Increasing the sample size would allow for more significant and meaningful trends to be observed. It would decrease the variability that is present in foot
motions between subjects and would decrease the standard deviations between groups.

- A technique to collect motion capture and fluoroscopy data within the same software would help to align the motions from both techniques with more accuracy. One collection system would also facilitate data processing, but especially data collection. Having a single system would allow one person to collect data and would reduce the chance of having one of the two systems not respond during a trial.

- A fluoroscope with a bigger image intensifier would allow for a greater capture volume. It would also permit testing subjects with bigger feet, which would be beneficial when using motion capture with the 3-marker marker clusters, as they are further apart on a bigger foot.

- Having a more automated data-processing technique will allow for post-processing to be sped up and more user friendly. Thus, more subjects could be tested in a smaller amount of time.

Future research should focus on determining how valid the MSFM is in a subject population with other foot types. It would be important to see if it is a valid tool to use on people who have planus and cavus feet. Moreover, testing the model to use with shoes would be an appropriate next step. It would allow for the effect of different shoes and running shoes be tested on foot joint motions.
4.5 Significance

In conclusion, this project first validated generic bone models for the use of bi-planar RSA fluoroscopy. Four generic bone models were used to estimate the angle of the MLA and compared to the angle found by using subject-specific bone models created from the subject’s CT scan of the foot. Overall, the generic bone model of the same size foot and same type foot exhibited the lowest difference from the subject-specific bone model. This finding has clinical significance because it would allow for the elimination of a CT scan prior to testing. This would be safer for the subjects/patients and would be more cost effective as well. The second chapter validated a MSFM used with motion capture against bi-planar RSA fluoroscopy. Hindfoot supination/pronation and internal/external rotation motion were not significantly different between the two techniques. Forefoot supination/pronation motion had the same results; the two techniques were not significantly different. MLA height-to-length ratio during the heel-strike and mid-stance part of stance were not significantly different, but toe-off was significantly different between motion capture and fluoroscopic data. Greater differences between motion capture and fluoroscopy were seen during heel-strike and toe-off. There may be more skin motion during these moments since muscles are activated to a greater extent to either stabilize the foot at landing or clear it for the swing phase. Quantifying STA errors of the MSFM gave us knowledge about its accuracy when used in research and clinical settings. This model is a good tool to use with motion capture and will be of great benefit for clinicians as it is cost effective, less time consuming than fluoroscopy, and relatively simple to use.
5 References


6 Appendix

6.1 Appendix 1 – Cortex Instructions

1. START SYSTEM UP
   a. Turn on the computer
   b. Turn on toaster

2. CALIBRATE CAMERAS
   a. Load Setup – “MSFMValSetUp.cal”
   b. Click the calibration tab at the top of the screen
   c. Click on the bottom display of the screen
   d. Go to ‘Data Views’ and select ‘2-D display’
   e. Click the ‘All On’ button in the bottom left corner to display all six camera views
   f. Click ‘Connect to Cameras’
   g. A box will now pop up saying you have connected – make sure it says six cameras. Click “OK”
   h. Click ‘Run’ located along the bottom menu

i. Calibrate with the square
   i. Place the square in the correct position
   ii. Check all 6 camera views to make sure that each camera is seeing 4 markers
      1. If there are more than 4 markers in a camera view, block out the phantom markers by holding the middle mouse button and dragging the mouse to create a mask over the marker
      2. If there are less than 4 markers in a camera view, then the cameras may need to be reconnected or adjusted. Try clicking ‘Disconnect – Use Raw Files’ and repeating steps 2.f. to 2.h. to reconnect to the cameras
   iii. Click ‘Collect and Calibrate’ on the right hand side menu under the ‘Calibration with a Square’ section
   iv. Click ‘OK to Overwrite’ and then ‘Collect and Calibrate’ again
   v. Remove the square from the force plate

j. Calibrate with the wand
   i. To calibrate with the wand, enter the capture volume with the wand and ask someone to click ‘Collect and Calibrate’ located under the ‘Calibration with Wand’ section
   ii. Proceed with the calibration for the required three minutes

k. Completing the Calibration
   i. When the three minutes is complete, a box will pop up on the screen automatically. Click ‘OK’
ii. Once the calibration is complete another box with 3D residual and wand length information will be displayed. Click ‘Run Again’

iii. After the second analysis has finished, check to make sure the numbers displayed are ok.
   1. For the 3D residuals, the average should be around 0.5 (less than 1) and the standard deviation should be under 0.2
   2. For the wand length, the average should be around 500 and the standard deviation should be under 1.0
   3. If the numbers are okay, record them in the calibration log along with the date and time of calibration

iv. After the numbers have been recorded, click ‘Accept’

v. Click ‘OK’ to save as the system calibration. Click ‘OK’ again.

vi. Save the project.

1. Loading the system calibration to the ‘HHGaitStatic’ file
   i. Load the ‘.prj’ file for the same person you just calibrated for by going to File, Load Project.
   ii. Once the ‘HHGaitStatic’ file is open, go to File, Load Calibration. Select ‘.prj’.
   iii. Save the project.

m. Loading the system calibration to all subjects of the day
   i. Select the appropriate study file. Locate the subject’s file that you just calibrated.
   ii. Select the ‘.prj’ and the ‘.prj’ files.
   iii. Copy the files.
   iv. Open each of the remaining subject’s folders for today’s appointment and paste the ‘.prj’ and the ‘.prj’ files into the folders.

3. MOTION CAPTURE
   a. Static Trials
      i. Load the ‘HHGaitStatic’ project file for the correct patient by going to File, Load Project.
      ii. Click the ‘Motion Capture’ tab located along the top of the screen.
      iii. Click ‘Connect to Cameras’
      iv. A box will pop up saying you have connected – make sure it says 6 cameras. Click ‘OK’.
      v. Click on the bottom half of the screen so the black box outlines the bottom display.
      vi. Go to ‘Data Views’ and select ‘Analog Display’.
      viii. Check the ‘Auto Increment’ box
ix. Set the ‘Trial #’ to 1 and the ‘Duration’ to 3, 5 or 20 sec, depending on the trial.
x. Click ‘Run’ along the bottom menu.
xii. Once the patient is in position click ‘RECORD’
iii. Trial saves automatically.
iv. Check if trial saved.

b. **Dynamic Trials**
   i. Increase time required for capture (~60s)
   ii. Press ‘RECORD’ – make sure time is running.
   iii. Say OK to the technician.
   iv. Press ‘STOP’ when the subject has walked through.

4. **POST PROCESSING**
   a. **Static Trials**
      i. Load the ‘HHGaitStatic’ project file for the correct patient by going to File, Load Project.
      ii. Click the ‘Post Process’ tab located along the top of the screen.
      iii. Click on the bottom half of the screen so the black box outlines the bottom display.
      iv. Go to ‘Data Views’ and select ‘XYZ Graphs’.
      v. Load the desired trial by going to File, Load Tracks File.
      vi. Verify that the cameras see X# of markers. The number of markers seen is located in the bottom left corner of the top screen.

vii. **Identifying Markers**
   1. To process the trial, check to make sure the lower body markers have been identified. If they are showing up black and there are no connecting lines between them, then the markers need to be identified.
      a. If the markers need identifying for the entire trial, make sure the entire bottom display is highlighted blue. To highlight the entire display, click on the correct arrow key in the bottom right hand corner.
      b. If the markers need identifying for only a portion of the trial, move the red line to the beginning of the section.
         i. The red line can be moved by clicking directly on the bottom display at the desired location or it can be moved by left clicking on the red line and holding the mouse key down while moving the line to the desired position.
         ii. To move the red line frame-by-frame, use the arrow keys located at the bottom of the screen in the middle.
iii. To highlight the section, middle click on the red line and hold the mouse key down while moving the line to the end of the section.

2. Once the correct section is highlighted, click on ‘Quick ID’ located on the right hand side menu. As each marker is called up on the Quick ID menu, click on the corresponding marker.
   a. If a body marker is missing on the figure and you are trying to identify the markers, skip the missing marker on the ‘Quick ID’ menu by using the arrow keys.

viii. Deleting Unknown Markers
1. The upper body markers should still be black and unidentified after using the ‘Quick ID’ function. They can now be deleted.
   a. To delete them, first click the ‘All/None’ button on the right hand side menu for the colored lower body markers. Keep clicking ‘All/None’ button until the lower body markers are completely un-highlighted.
   b. Now click the ‘All/None’ button for the black unknown markers that are listed in the section below the colored markers. Click the ‘All/None’ button until the unknown markers (identified by U_#) are all highlighted.
   c. Make sure the entire bottom display is highlighted blue and then delete the unknown markers. To delete the markers, left click on the mouse and select ‘Cut Inside’ or press the hot key ‘z’. (To undo something, right click and select ‘Undo’)
   d. Check that the unknown markers are now identified by ‘U_#<Empty>’.

ix. Joining Markers
1. Re-highlight the colored lower body markers and make sure the bottom display is highlighted blue. Look at the top of the XYZ graph for any vertical lines or triangles.
2. Vertical lines indicate that the corresponding coloured marker is missing for the section the vertical lines are located at.
3. Join – Cubic Function
   a. If the missing section is small (less than 1 cm), right click and select ‘Join – cubic’ or press the hot key ‘x’.
   b. You can also join cubic for one marker at a time. To do this, select the marker you wish to join
making it the only marker highlighted and then follow the same procedure just described for joining a section.

4. **Join – Virtual Function**
   a. If the missing section is large (greater than 1 cm), highlight/select the marker with the missing section, right click and select ‘Join – Virtual’ or use the hot key ‘v’.
   b. A ‘Virtual Marker Join’ menu will appear. The name of the marker that is missing a section should automatically appear in the ‘Multiple Markers to Join’ box. If the name does not appear, click on the ‘Multiple Markers to Join’ box to highlight it with the blue outline and then click on the desired marker.
   c. The blue outline should automatically move to the next box, the ‘Origin Marker’ box. If it does not move automatically, click on the box to highlight it.
   d. Select an appropriate neighbouring marker to the marker requiring joining, so that the name appears in the ‘Origin Marker’ box. When selecting an appropriate marker, make sure there are no sections missing from the marker.
   e. Repeat the same procedure for the ‘Long Axis Marker’ and ‘Plane Marker’ boxes.

5. **Creating a Virtual Marker**
   a. If a marker is missing for the entire trial, a virtual marker must be created.
   b. To do this, load the previous track file.
   c. Go to ‘Tools’, then ‘Virtual Marker Definitions’.
   d. Click on the ‘Enter Name of Virtual Marker’ box and create a new name for the virtual marker ie. V_L.ASIS.
   e. Click on and select an appropriate marker for each of the ‘Origin Marker’, ‘Long Axis Marker (Y)’ and ‘Plane Marker (XY)’ boxes.
   f. Click on the ‘Snap to this Marker (optional)’ box and select the marker that you are creating a virtual marker for.
   g. Click ‘Calculate Virtual Markers’
   h. Close the ‘Virtual Marker Definitions’ box.
   i. Save the project in the same subject’s file as something new ie. HHGait2.prj.
   j. Reload the track with the missing marker.
   k. Open the ‘Virtual Marker Definitions’ box under the ‘Tools’ menu.
l. Click ‘Calculate Virtual Markers’.
m. Close the ‘Virtual Markers Definition’ box.
n. Click on the virtual marker you just created. Hold the control key down and click on the marker name that is missing. These two markers should be the only markers highlighted.
o. Exchange the markers by clicking ‘Exchange’ located on the menu in the upper right hand corner of by pressing the hot key ‘e’.
p. Save the track.
q. Load the original project ie. HHGait.prj.
r. Finish tracking, saving and exporting as normal with the same trial.
x. Saving and Exporting the Trial
   1. Once all the markers have been joined and smoothed and all the unknown markers have been deleted, the trial is ready to be saved and exported.
   2. Click File, Export....
6.2 Appendix 2 - Using the Fluoroscopes

1. **Switching on the fluoroscopes**
   a. Unlock with the key. Insert into the side of the tower and turn.
   b. Switch on the unit using the ‘circle’ button on the control panel of the tower.

2. **Moving the fluoroscopes**
   a. Release the brakes before moving the fluoroscopes.
   b. To release the brakes, step on the lever above one of the two wheels on either side.
   c. When the levers are tilted either forward or backward the brakes are on.
   d. The fluoroscopes may be moved along a straight line, as dictated by the alignment of the wheels, or pivoted, when the wheels are centrally aligned.
   e. The wheels of the fluoroscopes can be turned using the handle. Pull the handle up to turn the wheels and push it own in the centre position to pivot the fluoroscopes.
   f. The C-arms can be raised and lowered using the arrow keys at either side of the control panel on the fluoroscopy units.
   g. Should the tower not respond to the arrow keys, ensure that the ‘STOP’ button has not been activated.
   h. The ‘STOP’ button may be found on front of the base of the unit, at about knee-level. It may be released by turning it, allowing it to pop out.
   i. Each C-arm may be further manipulated using three rotational axes (colour-coded blue, yellow, and orange) and one translational axis (colour-coded green).
   j. To move about or along any axis, turn the colour-coded handle to the unlocked position. Ensure that all axes are locked prior to beginning any data collection.
   k. To aid in positioning the C-arms there is a laser that can be used. It is turned on via the control panel on the fluoroscopy unit.

3. **Capturing Images**
   a. Prior to taking fluoroscopy images, be sure to follow the most appropriate action to avoid exposure to radiation.
   b. Images are taken using either the hand trigger or the foot pedals.
   c. The foot pedal on the left captures static images while the one on the right captures fluoroscopy.
   d. The three orange lights on the tower light up when any of the triggers are pressed.
   e. The C-arms are equipped with automatic brightness control to maintain the brightness of the image at a constant level. To accomplish this, the x-ray exposure rate is automatically regulated and the kVp and mA levels are automatically changed. The mA and kVp curves that are followed are shown in the Siemens SIREMOBIL Compact-L mobile C-arms manual.
   In most cases this will be the appropriate setting to use; however, in some cases in which there are dramatic differences in the contrast of the images it may be more appropriate to turn this off and manually adjust the settings.
to emphasize the objects of interest. Please see the Siemens SIREMOBIL Compact-L mobile C-arms manual for more detailed instructions if this is the case.

4. **Manipulating the Image**
   a. There are several lines of text that are always present at the bottom left corner of the image when the fluoroscopes are first turned on.
   b. This text will cover part of the image and so it is desirable to turn it off using ‘show/hide text’ button on the control panel on the tower.
   c. The image orientation may be altered by flipping it either horizontally or vertically using the ‘flip image horizontally’ and ‘flip image vertically’ button the control panel of the fluoroscopy unit.
   d. The image can be rotated.
   e. The contrast of the image can be altered within four levels, using the image contrast adjustment button.

5. **Calibrating the fluoroscopes**
   a. Calibrate the fluoroscopes before and after data is collected.
   b. Re-calibrate the fluoroscopes if the get bumped.
   c. Place the calibration frame such that the two fiducial planes are adjacent to the two IIs.
   d. Note the orientation of the calibration frame. Record which fluoroscope ‘sees’ F/C1 and F/C2.
   e. Make sure each fluoroscope sees 6 beads of each fiducial and control plane
Figure C.1 Base of the fluoroscopy unit
Figure C.2. Control panel on the fluoroscopy unit
6.3 Appendix 3 – Collection Process

1) Open cameras
2) Turn on fluoroscopes
3) Press 'text' to take away the text at the bottom left of the screen
4) Press the 'reverse R' button on the fluoroscopes
5) Turn screens A & B on
6) **Calibrate motion analysis**
   a) Open cortex
   b) Load setup - Rebecca's marker set
   c) Connect to cameras - OK
   d) Hit RUN
   e) Auto mask
   f) Place the wand on the platform for the origin
   g) Hit: collect & calibrate twice
   h) Swing wand around
   i) Click OK
   j) U-Resolution
   k) V-Resolution
   l) F-Length
   m) 3D Resolution: less than 1
   n) Wand length: close to 500
   o) Save setup
7) **Motion Capture**
   a) Make sure the following are checked on the right side of the screen
      i) Raw video
      ii) Reference video
      iii) Tracked binary
   b) Name trial
   c) Auto-increment
   d) Duration - ex: 20 sec
   e) Hit RECORD
   f) Hit STOP
      i) Check if trials are present in the file & watch it to make sure it worked
   g) DR: Static - Eye: Dynamic
   h) For dynamic recording:
      i) Put time higher (~60s)
      ii) Press record - make sure time is running
      iii) Say OK to the technician
      iv) Press stop when patient has walked through
   i) If system crashes:
      i) Load setup
      ii) Add marker set under the object on the right
8) **Fluoroscope calibration**
   a) Place F1 planes closest to Image Intensifier (II)
   b) Have the red X from the lasers seen on both planes
c) Have at least 6 beads seen by both fluors

d) **Calibration trials**
   i) Record
   ii) Double check that it's in the folder
   iii) Call file name FluoroA-0001

e) Recalibrate the fluors after the patient if he kicks them. Recalibrate at the end.

f) Fluoro A sees: F2 C2

g) Fluoro B sees: F1 C1

9) **Correct for distortion**
   a) Put white board on fluoro
   b) Beads should be as parallel as possible to the top of the image
   c) Note: move fluors as little as possible

d) Fluoro A: trial 1

e) Fluoro B: trial 2

f) Press record

g) Types of files for motion analysis:
   i) .cap
   ii) .trb

10) **Digitize frames**
   a) Adobe Premiere Pro
      i) New project
      ii) DV – NTSC
         (1) Standard 48 kHz
      iii) Import
         (1) Select all video files
      iv) Drag to Timeline
      v) Drag cursor over 1st image to the 1st frame to export
      vi) File --> export frame --> click on file name
      vii) Settings --> tiff
      viii) Video --> Compressor: none
      ix) Static “export frame”
      x) Dynamic “export movie”
      xi) 29.97 EPS
      xii) 720 x 540
      xiii) Square Pixel Ratio 1.0
      xiv) Tiff files
      xv) Movie
         (1) Drag cursor to the beginning of the movie
         (2) Synch both fluors
         (3) Move the black cursor to line up with the red
         (4) Move the red cursor to the end of the movie
         (5) Export movie
6.4 Appendix 4 - Bone Segmentation Steps

OsiriX (Pixmeo, Geneva, Switzerland)

**Note:** Do not save any work throughout this unless this guide explicitly tells you to do so.

Hitting save will result in losing data that you may need in order to proceed, and may result in an error message in the process.

1) Open ‘Finder’ on the Desktop and in the Applications on the left menu bar, find OsiriX and double click to Open.

2) First, the CT scan files must be imported to the program and copied to the system before any manipulation can happen. Click on ‘Import’ at the top left and then select the series of CT files that you want to make into a 3D model.

3) Once all the files have copied to the Local Database (above), double click the subject or patient CT whose bones you would like to segment.

4) The following screen will pop up > Click “I agree”.

---

![OsiriX Interface](image.png)
5) Choose the file to the left that has the most images or preview the one that appears most suitable (see highlighted pink area below). In this case, the one with the most images was the one chosen based on the slice thickness and CT properties.

6) Go to the top pull down menu under 3D Viewer and choose 3D Volume Rendering.

The 3D Volume Rendering window looks like the one below, with the tools in the second menu from the top.
7) On the second menu from the top (circled in red above), immediately change the level of detail to FINE (as far as it will go to the left).

8) Click on the 3D presets menu to the left of that and choose the ‘Basic’ Group. Click on ‘Low Contrast’ and then click ‘Apply’. This will allow for easier segmentation of the bones as there will be less visible noise and soft tissue surrounding the bone.
9) Description of tool functions (top left of the window in step 6):

![Mouse button function](image)

a. The **Poison sign** (far right) will get rid of an entire bone at once. This tool may be very useful, however, some bones may appear to be separate but in reality, there is some connection somewhere to another bone. If that’s the case, this tool will remove two or more bones at one time.

b. The **scissors** will allow you to select an area in bright green and then hitting ‘**Enter**’ once made your selection (below) will keep what you’ve selected, whereas the ‘**Delete**’ button will remove what you’ve selected.
   i. It’s easier to scissor around the bone you want right off the start, and press enter, and then use the delete button to eliminate the other bones that are near or touching afterwards.
   ii. Note that the scissor function will cut everything in three dimensions from the plane you’ve chosen and protruding into the screen and bones behind the selection so be careful where you cut.

c. The green circle with the red dot allows you to place a red sphere on the bone, marking any necessary landmarks. These spheres will export as separate ‘mesh’ items, along with the single bone mesh.
   i. You can choose to put the red points on the landmarks before or after segmentation, depending on how easily identifiable they are without the
surrounding bone. For segmenting the navicular in this study, the spheres were positioned before segmentation to mark the most medial point of the tuberosity as well as the most dorsal aspect.

d. The green line segment is a measurement tool if you want to determine the length of any two objects in two dimensions

e. Greenish blue sphere – used to re-position the camera view, since you may be looking near the end of an extremity, the camera position may need to be changed to zoom in close on the right area.

f. Box tool – used to rotate the model in three dimensions. The combination of these last two tools will allow you to zoom in and out and get the correct angle to use the scissor tool.

g. Semi-circular arrow – rotates the object in the plane of view.

h. Magnifying glass – used to zoom in and out (as well as the right click button at all times)

i. Move function (four arrows) – left click will move the object within that plane of view. Used to reposition the object (similar to rotating the camera)

j. Window level (black square far left) will adjust the window level and width – general CT settings. The 3D present chosen has default values for these parameters; therefore, this is not used for the purposes of this segmentation.

Note: Hold mouse over function to see what each does if you aren’t sure. DO NOT hit the save button. This will create an error in the next step.

10) Only segment one bone at a time in the window – it is easier to crop a single bone without having to worry about what is behind it. Also, you want to export each bone separately to import into Rhinoceros.

11) Once the bone is segmented, the surface of the bone must be smoothed. Click on ‘3D Presents’ similar to step 7 and in that window change the group type to: Bone CT. Select option 9 “Soft”. 

Note: This setting has specific presents that show the best balance between colour and density of the bone for this thesis. If you click on ‘Info’, the 3D Present parameters will be shown – window length/width, the colour look up table (CLUT) and the filter used for the CT scan. These are the best surface properties for exporting the bone model to the best of the author’s knowledge.

12) At the top menu, select the 3D Viewer drop down menu again and select the 3D Surface Rendering option (below). This will create a mesh of the segmented bone by defining a surface around its known volume. The segmented bone model can only be exported from this 3D view.
13) Once selected, the menu at the top of the window will pop up automatically for input regarding the desired surface settings (see below).
a. The settings above will change depending on the patient, their bone density, as well as which bone you are working with.
b. Move the ‘Resolution’ cursor to two notches to the left of high to start, and move to HIGH if that appears better.
c. Initial settings should have ‘Smooth – iterations’ function to 1 (meaning less smoothing will occur at first).
d. Initially, the Pixel Value should be set to 100 (instead of 300 by default). This setting represents the ‘density’ of the bone, for example, 50 for one patient made the bone too built up with sharp edges, whereas 100 created holes in the bone. The higher the pixel value, the less dense the bone – this value will need to be manipulated depending on the subject.
e. You can also change the colour of the bone which may be a good idea to choose something that will work well in rhinoceros background.

**Important:** Once you set these values initially and they are too high (bone has holes and is not dense enough) then you cannot make it more dense by changing them in the ‘surface settings’ tab in the toolbar. You must close the window back to the 3D Rendering window and then start step 12 again. However, if the pixel value is started low, with a low ‘Smooth’ number as well, and the bone appears too dense, you can edit the surface settings by increasing the Iterations and Pixel Value gradually. I’ve found this to be the easiest way to get the bone looking the way you want. Start with low numbers and gradually increase them to the desired output.

14) The bone will now resemble the model below (example bone: first metatarsal of the left foot). From this point, the model can be exported as a ‘Wavefront’ or object file (.obj), which is found in the ‘Export 3D-SR icon’ on the Surface Rendering Menu to the right.

a. Select the folder you wish to save it in. The file can now be transferred to the PC of your choice so long as you have Rhinoceros on the machine for further analysis.
15) When closing any window, use the buttons on the top left, the red button. First, close the 3D Surface Rendering window, followed by the 3D reconstruction (volume) rendering. As long as you don’t close the subject CT file (the 2D view at the beginning, that bone will remain segmented.

16) To start a new segmentation for another bone for the same patient, the last window must be closed and then the subject re-opened to start again.

**Note:** the red spheres will not ever disappear automatically from where they were placed, even when closing the subject CT files. So you have to manually use the tool function, click on them and hit delete before adding them to the next bone.

17) To quit OsiriX, you have to go to the top left and click Quit OsiriX, closing the last window will not do that for you

Taking screen shots with a MacBook Pro

1. **Apple (Command) Key +Shift+3**
   Captures entire desktop to a file on the desktop as 'picture #' . This option lets you capture the whole screen.

2. **Apple (Command) Key +Shift+4**
   Allows you to use your mouse to select a specific part of your desktop for capture. This will turn your mouse pointer into a cross, please hold down the mouse button and drag to select the part of the screen you want. When you release the button the screenshot will "snap" that part of the screen. Press 'Esc' to release.

3. **Apple (Command) Key +Shift+4 then press Spacebar**
   Allows you to select which window to capture.
6.5 Appendix 5 – SPSS Output

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## Independent Samples Test

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### Independent Samples Test

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6.6 Appendix 6 – Ethics Approval Notice

Use of Human Participants - Ethics Approval Notice

Principal Investigator: Prof. Thomas Jenkyn
File Number: 103455
Review Level: Full Board
Approved Local Adult Participants: 50
Approved Local Minor Participants: 0
Protocol Title: Validation of a multi-segment foot model using bi-planar x-ray fluoroscopy
Department & Institution: Engineering/Mechanical & Materials Engineering Western University
Sponsor:
Ethics Approval Date: September 17, 2013
Ethics Expiry Date: May 31, 2014

Documents Reviewed & Approved & Documents Received for Information:

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This is to notify you that the University of Western Ontario Health Sciences Research Ethics Board (HSREB) which is organized and operates according to the Tri-Council Policy Statement: Ethical Conduct of Research Involving Humans and the Health Canada/ICH Good Clinical Practice Practices; Consolidated Guidelines; and the applicable laws and regulations of Ontario has reviewed and granted approval to the above referenced study on the approval date noted above. The membership of this HSREB also complies with the membership requirements for REB’s as defined in Division 5 of the Food and Drug Regulations.

The ethics approval for this study shall remain valid until the expiry date noted above assuming timely and acceptable responses to the HSREB’s periodic requests for surveillance and monitoring information. If you require an updated approval notice prior to that time you must request it using the University of Western Ontario Updated Approval Request form.

Member of the HSREB that are named as investigators in research studies, or declare a conflict of interest, do not participate in discussions related to, nor vote on, such studies when they are presented to the HSREB.

The Chair of the HSREB is Dr. Joseph Gilbert. The HSREB is registered with the U.S. Department of Health & Human Services under the IRB registration number IRB 00000940.

Signature

Ethics Officer to Contact for Further Information

Erika Bastik
(ekbastik@uwo.ca)
Grace Kelly
(grace.kelly@uwo.ca)
Vicki Tran
(vicki.tran@uwo.ca)

This is an official document. Please retain the original in your files.

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London, Ontario, Canada
2012-2014, M.Sc.

Related Work Experience:

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University of Western Ontario (Fall 2012)
Kin 3330F
Laboratory in Exercise Physiology

Teaching Assistant
University of Western Ontario (Winter 2013/2014)
Kin 3343B
Biomechanical Analysis of Discrete Skills

Poster Presentations:


7th World Congress of Biomechanics – July 6-11, 2014