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Quantitative analysis of the kinematics of the foot during gait with respect to barefoot running shoes using a Multi-Segment Foot Model

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Abstract
Kinematics of the foot during static and dynamic tasks are technically challenging to accurately measure, making it difficult to evaluate their contribution to running-related injuries. Motion capture can detect changes in running mechanics, such as with differing footwear. Habitual barefoot runners and/habitual minimalist shoe runners underwent a biomechanical evaluation of their foot movement during running using reflective markers and optical motion capture. A five-segment foot model was used to compare motions between the different parts of the foot while running barefoot and wearing Vibram Five Fingers™ (VFF) barefoot mimicking shoes. Supination/pronation in the forefoot was larger but not significant between habitual shod and habitual VFF runners. In contrast, the other foot motions (forefoot spreading/rising, hindfoot pronation/supination and hindfoot adduction/abduction) were not significantly different between the groups of runners. Therefore it could be possible that transitioning from a regular running shoe to a barefoot mimicking minimalist shoe would cause the foot to adopt a more supinated gait cycle.

Lay Summary
Movement of the foot due to it's complexity during still and moving tasks are challenging to measure. This makes it difficult to compare movements to running related injuries. This study took a look at Vibram Five Fingers and shod runners using a segmented foot model to see if differences were evident. The study used reflective markers and motion caption cameras to compare motion between the two groups of runners. There were no differences seen between the two groups of runners.

Keywords
Foot, biomechanics, multi-segment analysis, motion analysis, kinematics, shoe, motion capture.

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Glossary & Nomenclature

abduction (ABD)	movement towards the midline
adduction (ADD)	movement away from the midline
axis of rotation	straight line going through fixed points of a rotating rigid body
BF	barefoot
distal	further away from a point of attachment or origin
dorsiflexion (DF)	flexion of the ankle towards the body
eversion	abduction of the ankle
extension	increasing the angle of the joint
flexion	decreasing the angle of the joint
frontal plane	defined as separating the body into front and back
forefoot strike (FFS)	characterized by initial ground contact on the forefoot (first 1/3 of the foot) proceeded by transferring weight to the heel
gait	manner of walking or running
inversion (inv)	when the ankle supinates and the foot rolls towards the midline
kinematics	study of motion of the body without regard to the forces that are producing the motion
kinetics	study of forces that cause motion
lateral	away from the midline of the body
Matlab	software used to write code to read and analyse data
medial	situated towards the midline of the body or attachment or origin
medial longitudinal arch (MLA)	arch formed at the bottom of the foot
midfoot strike (MFS)	characterized by initial ground contact on the midfoot (middle 1/3 of the foot)
mm	millimetre
proximal	closer to the point of reference
plantar flexion	movement where the toes are pointed downwards towards the sole
posterior	towards the back of the body
pronation	combination movement of eversion, dorsiflexion and abduction
rearfoot strike (RFS)	characterized as initial ground contact on the rearfoot (rear 1/3 of the foot)
midsagittal plane	anatomical plane that runs through the midline and divides the body into left and right
superior	higher when referring to a point of reference
supination	combination movement of inversion, plantar flexion and adduction
tendon	tissue connecting muscle to bone
transverse plane	separating the body into an upper half and a lower half
valgus	a deformity in which an anatomical part is turned outward away from the midline of the body to an abnormal degree

varus	a deformity in which an anatomical part is turned inward toward the midline of the body to an abnormal degree
VFF	Vibram Five Fingers running shoes
WOBL	Wolf Orthopaedic Biomechanics Laboratory

Chapter 1: Background & Introduction

1.1 Statement of purpose

The purpose of this research was to quantify the relative motion between the forefoot and hindfoot in participants with healthy, non injured feet during running. Participants were divided into two groups. One group habitually ran in traditional running shoes (more than 10 kilometers per week), and the second group habitually ran in Vibram Five Fingers™ or barefoot (more than 10 kilometers per week).

This study used optical motion capture and a multi-segment foot model to individually track the segments of the hindfoot, midfoot, medial forefoot, lateral forefoot and hallux. The optical motion capture system used skin-mounted auto-reflective markers organized into rigid triad clusters. At least one cluster was attached to each foot segment. The multi-segment foot model and optical motion capture system have been previously validated in our laboratory (Jenkyn & Nicol, 2007). The multi-segment foot model measures the degrees of freedom motion between the five foot segments so that the function of the foot can be compared between subject types and footwear conditions.

This study had two hypotheses. The first hypothesis was that the kinematics of habitually shod runners are significantly different than habitual Vibram Five Fingers™ runners when running barefoot. The second hypothesis was that the kinematics of habitually shod runners are significantly different than habitual Vibram Five Fingers™ when running in Vibram Five Fingers™.

1.2 Introduction

For several centuries, shoes have been used to provide protection for the soles of the feet, traction between the feet and the ground, for motion control and stability, and for dispersion of ground reaction forces (Mcpoil, 2000). Runners typically strike the ground 600 times per kilometer (Milner, Ferber, Pollard, Hamill, & Davis, 2006; Pohl, Hamill, & Davis, 2009; van Gent et al., 2007). Each time the foot strikes the biomechanical loading is approximately 1.25 times body weight during walking and 2 to 3 times body weight during running (Lieberman et al., 2010; Waetjen, Parker, & Wilken, 2012). Thus, due to the repetitive nature of these foot strikes, any imbalance in foot and gait mechanics may result in a running related injury.

The high loads associated with running, and the associated risk of injury, catalyzed research studies looking into barefoot and shod running. The article by Lieberman et al. (2010) was one of the first published studies that compared the differences between habitually barefoot and shod runners. It sparked a lot of controversy in the running community. In a 2014 Runner's World interview, Dr Lieberman stated that changing your running mechanics affects more than just your footstrike; it can also change your joint moments, the cadence and tissue loading (Lovett, 2014). He goes on to say that there is a need for future studies examining different aspects of running.

Incidences of running related injuries and overuse injuries such as plantar fasciitis, stress fractures, tendonitis and shin splints have increased over time as the number of runners in Canada have also increased (van Gent et al., 2007) with the incidence of running related injuries range from 19-79% in any given year (van Gent et al., 2007). Despite advancements in running shoe technology and design (increases in cushioning, alterations to materials, changes flexibility and stability) rates of repetitive strain injuries vary little from year to year (Richards, Magin, & Callister, 2008; Van Mechelen, 1992) and therefore perhaps variations in shoe material are not the sole contributor to overuse injuries.

There are many different types of shoes that will attract the attention of various runners (Klettler, 2005). These include a variety of colours, styles, types, foams, gels, cushioning, air pockets, lacing, etc. Running shoe prices range from $50-$225 with features varying in the addition of motion control (midsole stabilization), cushioning systems (gels and air pockets), and differing lacing techniques (Klettler, 2005). Although there are a lot of features to choose from, mileage is another factor (Mcpoil, 2000). Mileage affects how shoes can function over time, during inclimate weather and over varying distances and thus, it is important to regularly change your shoes (Mcpoil, 2000; Taunton et al., 2002).

The heel of a stability running shoe is compromised of the outsole and the midsole. Modern stability running shoes commonly have the heel-toe offset, with the heel/hindfoot raised higher than the forefoot. This is thought to encourage a heel strike during running. Recent evidence suggests that wearing a stability shoe during running has led to the weakening of the intrinsic muscles of the foot and increasing the risk for running related injury (Sinclair, 2015). There have also been studies examining the effects of various surfaces on foot kinematics. A study by Gantz & Derrick, (2018), showed that when running on an irregular surface, the forefoot has a decreased amount of inversion. Another study determined that there was an increase in variability in footstrike patterns between hard and soft surfaces (Lieberman et al., 2015). The hard surface was described as a dirt field with compacted soil and a soft surface was the same dirt field that was disrupted 10 cm down to loosen up the terrain (Lieberman et al., 2015). Habitually barefoot runners were more likely to rearfoot strike on soft surfaces, and habitually shod runners were more likely to forefoot strike on soft surfaces (Lieberman et al., 2015). Barefoot runners were significantly more likely to use more than one foot strike type on varying terrain (72%), compared to shod runners (32%) (Lieberman et al., 2015). The difference in footstrike patterns between hard and soft surfaces were not significantly different between habitually shod and barefoot runners, however, there was a significant difference in the variability of foot strike on the soft surface between all the runners.

Running shoes that have a decreased heel-toe offset are thought to encourage more of a midfoot strike or even a forefoot strike (Mcpoil, 2000). Barefoot mimicking or minimalist running shoes were introduced in an effort to achieve a forefoot strike (Altman & Davis, 2012; Curran & Tozer, 2010; Nunns, House, Fallowfield, Allsopp, & Dixon, 2013; Warne & Gruber, 2017; Zhang, Paquette, & Zhang, 2013). Barefoot mimicking running shoes are a technology that entered the market in 2005. This barefoot running trend started with the introduction of barefoot mimicking running shoes such as the Nike Free (Nike, Inc., Beaverton, Oregon) and most notably, the Vibram Five Fingers™ (Vibram, Albizzate, Italy). A lawsuit was launched against Vibram in 2012 as they claimed that the shoes prevented common injuries and worked to strengthen the feet (Tucker, 2014). Vibram settled this claim and refunded their customers between $20-$50 per pair as their claims could not be supported by science (Tucker, 2014).

In contrast, true barefoot running would consist of running with nothing on the soles of the feet (Murphy, Curry, & Matzkin, 2013). The goal of a minimalistic shoe is to mimic barefoot running via little-to-no cushioning, no support on the upper mesh portion, no heel-toe offset on the sole, and little-to-no laces that constrain forefoot spreading (Bonacci et al., 2013; Willy & Davis, 2013) .

Runners are likely to run in barefoot if they have grown up barefoot running such as tribesmen in remote locations in the world, those who have transitioned from minimalist shoes into full barefoot running, and those who believe it is better for the feet. Many runners currently wear or have tried training in minimalist shoes. Therefore, in theory, the use of barefoot mimicking shoes such as the Vibram Five Fingers™ (VFF), should yield similar outcomes in foot strength and movement to an individual running truly barefoot.

1.3 Foot anatomy

1.3.1 Bones of the foot

The bones of the foot consist of the bones distal to the ankle (Nordin & Frankel, 2001). The foot consists of 28 bones (including sesamoids), 33 synovial joints, 34 muscles and more than 100 ligaments (Nordin & Frankel, 2001). The foot structure is made up of the phalanges (14), metatarsals (5), cuboid, cuneiforms (3), sesamoids (2), navicular, talus and calcaneus (Figure 1.1). The sesamoid bones of the foot are oval shaped bones (two on each foot) that lie under head of the first metatarsal and are embedded within the tendons (Martin & McFerran, 2014). They act to bear weight from the first metatarsal and also act as a pulley during the toe off phase of the gait cycle, creating leverage during walking and running (Martin & McFerran, 2014). This is important when considering barefoot running, as this mechanism is what propels the body forwards. There are 3 phalanges in each toe except for the hallux, which has two. The phalanges act to stabilize the foot (Nordin & Frankel, 2001). Of the metatarsal bones, the hallux is the shortest and thickest and acts to propel the body forward, thus bearing the most body weight during walking, running and barefoot running (Nordin & Frankel, 2001). The phalanges are connected to the midfoot via the metatarsals. The metatarsals form the forefoot and connect the tarsals to the toes (Nordin & Frankel, 2001). Five of seven tarsals are found in the midfoot via the cuneiforms (3), the navicular and the cuboid.

The hindfoot consists of the remaining two tarsal bones: the talus and the calcaneus. The calcaneus is the largest tarsal bone and forms the heel of the foot (Nordin & Frankel, 2001). The talus is located directly above the calcaneus and forms the pivot point of the ankle joint (Nordin & Frankel, 2001). It enables the movements of inversion and eversion of the ankle. The calcaneal tuberosity is located at the rear of the calcaneus and serves as the attachment of the Achilles tendon (Nordin & Frankel, 2001). This area is covered by a fat pad makes contact with the ground during weight bearing activity (Nordin & Frankel, 2001). Sometimes, the bones of the midfoot and hindfoot (7) are call the tarsus. Their irregular shapes allows them to interlock to allow for a highly stable weight bearing structure (Nordin & Frankel, 2001). Since this mechanism is highly weight bearing, the tarsus take on a lot of the body weight and ground reaction forces during shod running and barefoot running (Cavanagh, Williams, & Clarke, 1981).

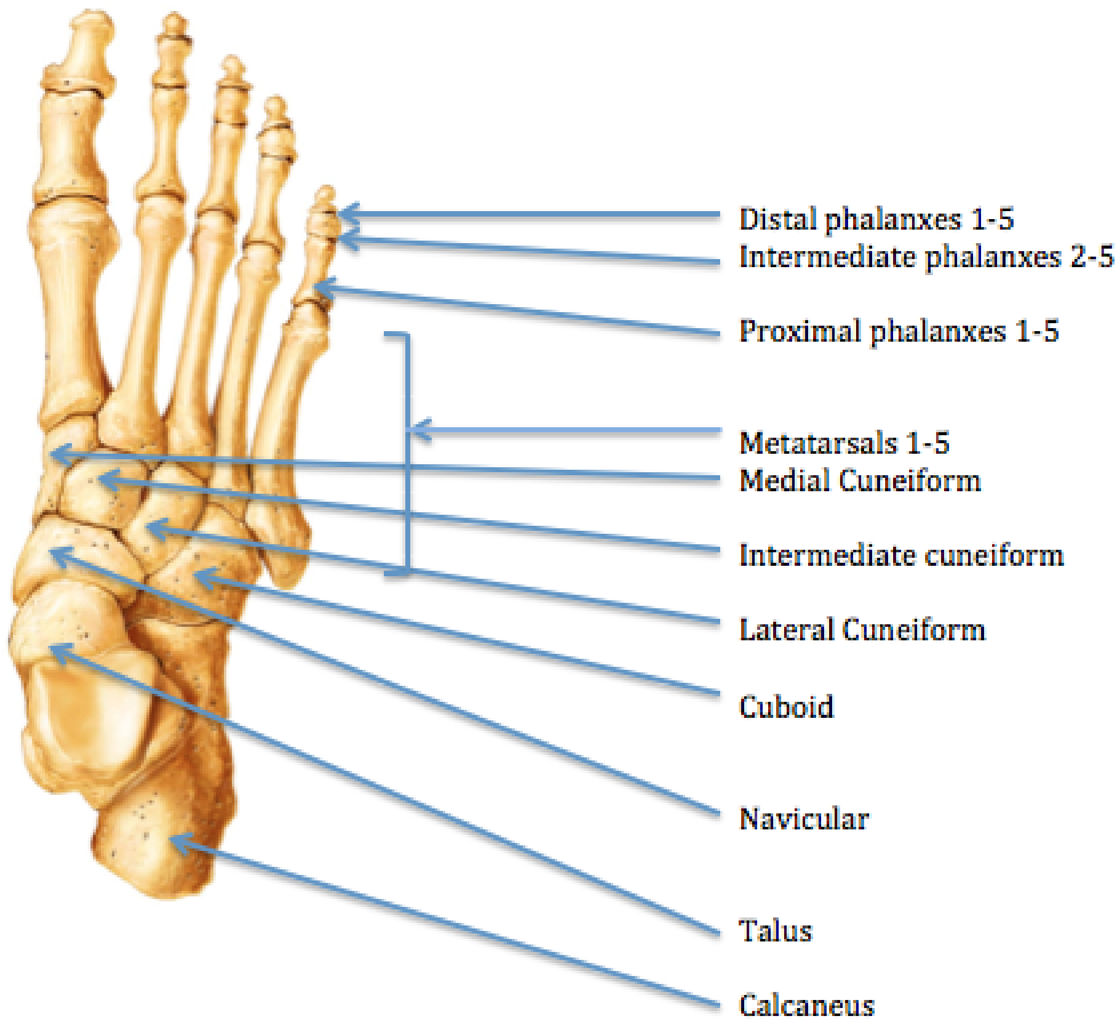


Figure 1.1: The 26 bones of the foot and ankle (shown in the transverse plane from above) (adapted from Nordin & Frankel, 2012).

The talus articulates superiorly with the tibia and fibula (talocrural joint) and inferiorly with the calcaneus and navicular (subtalar joint). The cuboid articulates with the navicular and lateral cuneiform medially, the calcaneus proximally and the base of the 4th and 5th metatarsal bases distally.

The articular surface of the navicular can be divided into three facets. Each facet articulates with the intermediate, medial and lateral cuneiform. The intermediate cuneiform articulates with the medial cuneiform medially, and the base of the 2nd metatarsal distally and lateral cuneiform laterally. The medial cuneiform articulates with the base of the 1st and 2nd metatarsals distally and the intermediate cuneiform laterally. The lateral cuneiform articulates with the 3rd metatarsal distally, the 2nd metatarsal medially and the 4th metatarsal laterally.

Interphalangeal joints are formed between the proximal and middle articulating surfaces of the phalanges as well as between the middle and the distal articulating surfaces of the phalanges. All articulating surfaces are covered in hyaline cartilage, which provides a smooth, lubricated surface to allow of joint and relatively friction-free movements. This cartilage compresses when loaded, which allows for the articular surfaces to spread out forces acting within the joints (Nordin & Frankel, 2001). Articulations between the cuneiforms, navicular and cuboid are gliding joints where small amounts of rotation can occur (Nordin & Frankel, 2001). Between the cuneiforms, there is a small amount of vertical motion that can alter the shape of the transverse arch. The medial longitudinal arch (MLA) consists of the calcaneus, talus, navicular, cuneiforms and the first three metatarsals (Nordin & Frankel, 2001). The MLA is the highest arch of the foot (Nordin & Frankel, 2001). The MLA is supported via tibialis anterior and posterior, flexor hallucis longus and flexor digitorum longus. The lateral longitudinal arch is lower than the MLA and consists of the calcaneus, cuboid, 4th and 5th metatarsals (Nordin & Frankel, 2001). The lateral longitudinal arch is supported by peroneus brevis and longus and some parts of flexor digitorum longus. The transverse arch is located perpendicular to the longitudinal arches and consists of the cuneiforms, cuboid and all five metatarsals (Nordin & Frankel, 2001). The transverse arch is supported via peroneus longus. The phalangeal joints are located internally between the proximal and middle phalanges and the middle and distal phalanges. They are uniaxial and thus allow only for flexion and extension. The proximal phalange is concave whereas the distal phalange is convex.

1.3.2 Functional units of the foot

The forefoot acts to propel the body forward (Nordin & Frankel, 2001). This is done via the hallux and the attached sesamoids as this spring like structure creates the forward motion of the body during gait (Nordin & Frankel, 2001). The tibialis anterior acts to dorsiflex the ankle and assists with foot inversion (Nordin & Frankel, 2001). Peroneus longus acts to evert the foot and assist to plantarflex the ankle (Nordin & Frankel, 2001).

The fifth metatarsal provides the attachment for the peroneus brevis (Nordin & Frankel, 2001). There are two sesamoid bones located under the 1st metatarsal on the medial and lateral side (Nordin & Frankel, 2001). These are located between the 1st metatarsal and the tendon of flexor hallucis brevis (Nordin & Frankel, 2001). Peroneus brevis acts to evert the foot and assists in ankle plantar flexion (Nordin & Frankel, 2001). Flexior hallucis brevis acts to plantarflex the hallux during propulsion (Nordin & Frankel, 2001).

The midfoot which consists of the cuneiforms (3), navicular and cuboid serves to invert and evert the foot as well as plantarflex the ankle (Nordin & Frankel, 2001). Inversion occurs via tension on the attachment of tibialis posterior to the navicular, and eversion occurs via the tension in the peronus longus that passes over the cuboid (Nordin & Frankel, 2001).

The hindfoot consists of the calcaneus and talus. The hindfoot enables ankle plantar flexion via tension in the gastrocnemius, soleus and thus, the Achilles tendon (Nordin & Frankel, 2001). The flexor hallucis longus is located deep inside the calf and acts to plantarflex the hallux and ankle as well as foot inversion (Nordin & Frankel, 2001).

1.3.3 Major joints of the foot and ankle

The foot connects to the rest of the body at the ankle joint (talocrural joint) and the subtalar joints (talocalcanealnavicular joint). The ankle joint is the articulation between the tibia and fibula with the talus below. The fibula forms the lateral malleolus (bony point on lateral side of ankle) and the tibia forms the medial malleolus. The bony protrusions articulate with the medial and lateral sides of the talus which allows the talus to pivot (Nordin & Frankel, 2001).

The subtalar joint is the articulation between the talus and the calcaneus (Figure 1.1). The hindfoot and midfoot allow for movements of inversion and eversion. The ankle or talocrural joint has allows the movements of plantar flexion and dorsiflexion (Figure 1.2). Plantar flexion is as the motion of pointing the foot downwards. Dorsiflexion is the motion of lifting the foot upwards (Nordin & Frankel, 2001) (Figure 1.2).

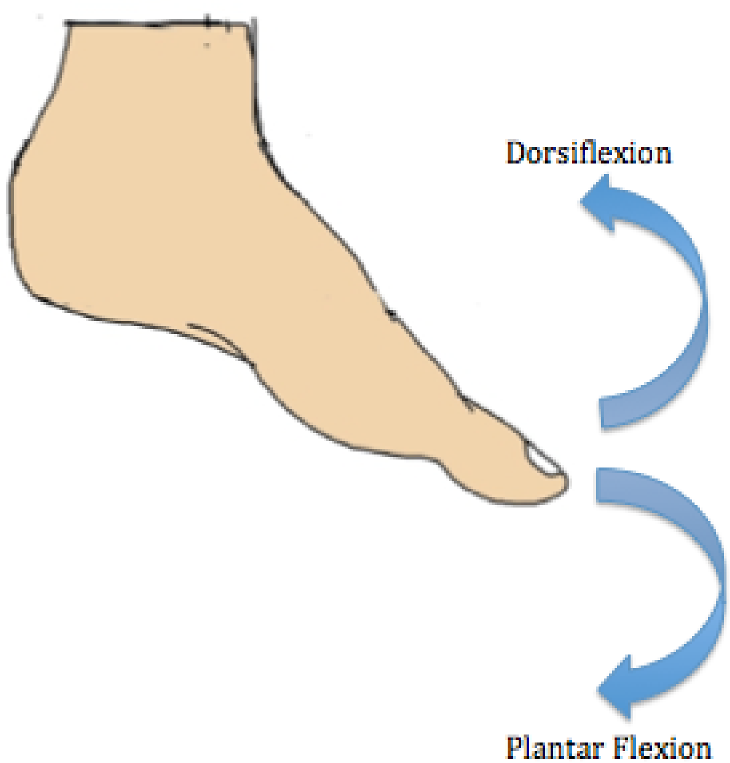


Figure 1.2: The movements of dorsiflexion and plantar flexion. Dorsiflexion involves lifting the foot up towards the anterior shin and plantar flexion involves pointing the toe downwards. This motion is fundamental throughout the gait cycle, used in order to propel the body forward during each swing phase

The talus articulates superiorly with the tibia and fibula at the talocrural joint forming the ankle.

Figure 1.3: Movements of the talocrural joint; plantar flexion and dorsiflexion (adapted from Nordin & Frankel, 2012).

The axis of the ankle joint is oriented between the malleoli (Nordin & Frankel, 2001), and cab be located by palpation. Movement about the ankle joint determines the plantar flexion and dorsiflexion of the ankle. Movement occurs in all three planes and occurs about the oblique axis (Nordin & Frankel, 2001).

The subtalar joint is the articulation between the talus and calcaneus (Figure 1.4). Here the talus articulates inferiorly with the calcaneus and navicular at the subtalar joint. It has one degree of freedom and has motions that consist of a combination of adduction/abduction and inversion/eversion during both walking and running (Nordin & Frankel, 2001)(Figure 1.4). Due to the nature of having one degree of freedom, movement within one action occurs simultaneously in the other (Nordin & Frankel, 2001).

Figure 1.4: Movement of the subtalar joint; adduction and abduction, inversion and eversion (adapted from Nordin & Frankel, 2012). The subtalar joint is made up of the talus and the calcaneus. The subtalar joint allows for pronation and supination of the foot.

The transverse tarsal joint consists of movement between two pairs of bones. The first pair being movement between the talus and the navicular and the second pair being movement between the calcaneus to the cuboid (Nordin & Frankel, 2001). This joint is very important to the overall function and movement of the foot as it allows for inversion and eversion of the midfoot relative to the hindfoot (Nordin & Frankel, 2001). The calcaneocuboid joint is located on the lateral side and the talonavicular joint is located on the medial side.

The calcaneocuboid joint is consists of the calcaneus and the cuboid. The anterior calcaneal surface is saddle shaped (convex transversely and concave vertically), and the articular surface of the cuboid is also saddle shaped, however it is convex vertically and concave transversely.

The longitudinal movement at the midtarsal joint occurs along an axis that passes through the postero-lateral calcaneus and the top of the cuboid. Movement here can be classified as pronation-abduction and supination-adduction when the calcaneus is fixed. The transverse axis consists of movement of dorsiflexion-abduction and plantar flexion-adduction when the talus and calcaneus are fixed.

Movement at the midtarsal joint depends on the position of the subtalar joint. When the subtalar joint is pronated, the axes are parallel, essentially unlocking the joint and creating greater flexibility. This greater flexibility is what allows for shock absorption and perhaps more importantly, the ability to adapt to uneven surfaces when running or walking. In contrast, when the subtalar joint is supinated, the midtarsal joint is locked, which creates rigidity. This rigidity is what allows the foot to propel the body forward during locomotion.

Medial longitudinal arch (MLA)
The medial longitudinal arch consists of the calcaneus, talus, navicular, cuneiforms and the first three metatarsals. It is supported via the calcaneonavicular ligament, plantar fascia and the plantar ligament. The MLA dampens the ground reaction forces through arch deflection. In specific, the plantar fascia and associated ligaments contract and expand with unloading and loading to take the foot from a load bearing platform to a rigid lever for propulsion (Kindred, Trubey, & Simons, 2011).

During the loading phase of stance, the MLA deflects interiorly, allowing muscles surrounding the arch to store energy and be used during push off (Fukano & Fukubayashi, 2009). The windlass effect involves tightening of the plantar fascia, which is a thick band of connective tissue that spans the calcaneus to the metatarsal heads (Figure 1.5). The plantar fascia originates on the medial tuberosity of the calcaneus and inserts on the metatarsophalangeal plantar plates (Dugan & Bhat, 2005; Nordin & Frankel, 2001). The windlass mechanism describes tightening the plantar fascia due to the extension of the hallux and pulling calcaneus and metatarsals closer, creating a structural support to propel the body forward. The midfoot is locked via the internal rotation of the lower leg segment driving the head of the talus medially and the cuboid and navicular subtly change their relative positions. It is also thought that the windlass mechanism aids in transferring energy and momentum from the gastrocnemius and soleus to the foot to propel the body forward on toe off (Kindred et al., 2011).

The lateral longitudinal arch
The lateral longitudinal arch is formed by the calcaneus, cuboid, and metatarsals four and five. This arch runs parallel and laterally to the MLA. This arch is very shallow and for people with a high arch (or pes cavus) foot type, this arch may touch the ground. This arch is not as flexible as the MLA and it transmits weight during locomotion.

Transverse arch
The transverse arch is formed by the tarsals and the base of the metatarsals. This arch will flatten during weight bearing, resulting in the forefoot to spread. This spreading allows the forefoot to support the body weight as it prepares for toe off.

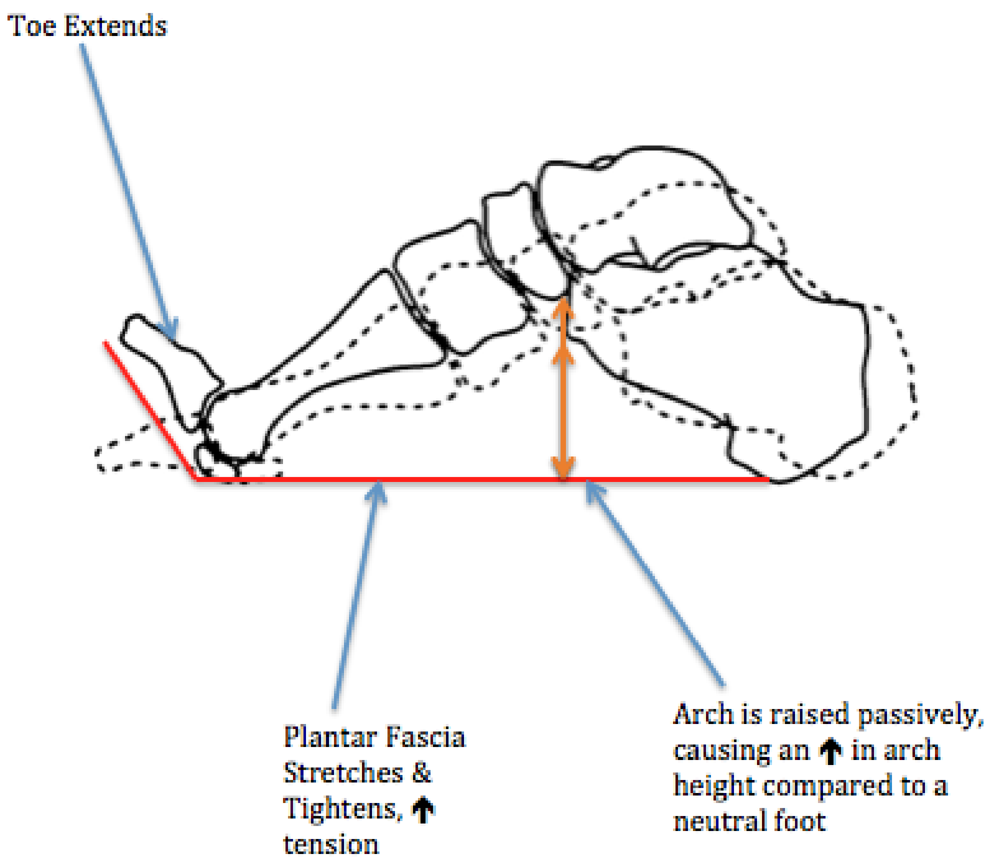


Figure 1.5 The Windlass Mechanism. The toe extends, causing tension in the plantar fascia, resulting in a raised arch.

MLA height to length ratio

The most appropriate way to measure the MLA position has been debated. A simple method uses the length and the height of the MLA to characterize its shape via the truncated foot length. The truncated foot length is measured by the length from the heel to the first metatarsal head. This has been shown to be a valid approximation of MLA structure compared to radiographic images of the bones of the arch (Saltzman, Nawoczenski, & Talbot, 1995). The arch height to arch length ratio is commonly reported (Saltzman et al., 1995)(Figure 1.6). This ratio was used in the model validation study by Jenkyn & Nicol (2007).

As the arch goes up, the height to length ratio increases (narrows) and as the arch goes down, the height to length ratio decreases (widens). In both walking and running gait, as the foot reaches toe off, the arch narrows (height to length ratio increases). Researchers and clinicians find that the height to length ratio of the MLA is an important measure and it is often used in the determination and classification of foot pathologies and biomechanical gait analyses (Caravaggi et al., 2018).

The main function of the arch is to aid in spreading ground reaction forces upon contact with the ground. By absorbing some of the force, the foot is protected from repeated strains and potential damage to soft tissues (Nachbauer & Nigg, 1992). Typically, arch height determines susceptibility to injury (Mei-Dan et al., 2005). Pes planus or flat footed people are more likely to pronate; over time this can lead to foot pain, heel pain and plantar fasciitis (Mei-Dan et al., 2005). Mei-Dan et al., (2005) found that those with low arches are more likely to have a higher rate of ankle sprains than in normal arched patients (55% to 39%). High arched or pes cavus people are more likely to stand supinated which can also lead to plantar fasciitis (Mei-Dan et al., 2005).

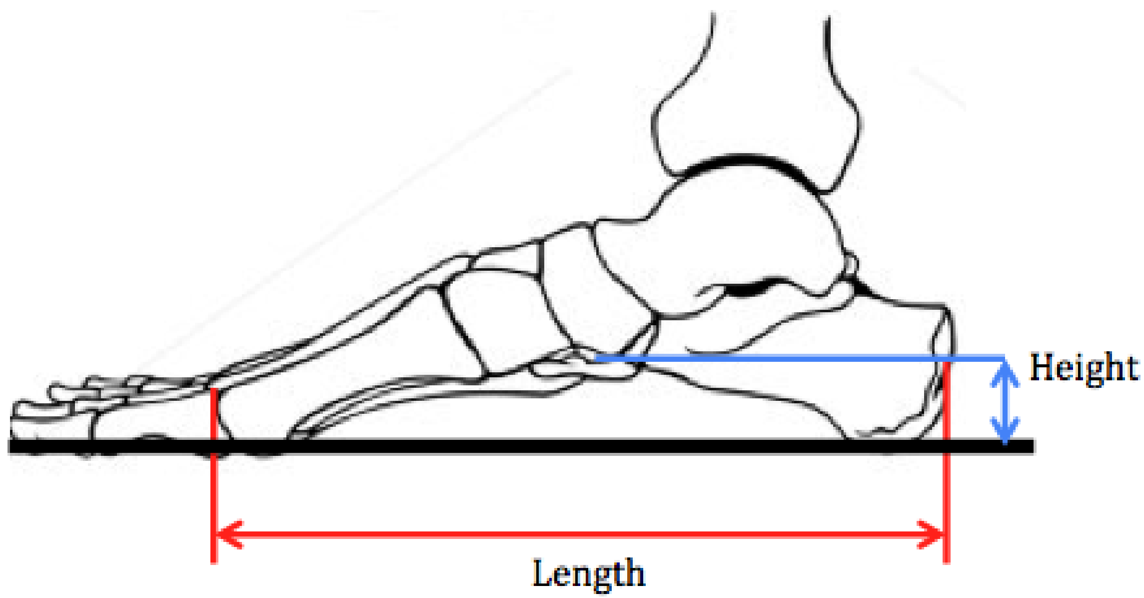


Figure 1.6 Medial Longitudinal Arch (MLA) height to length ratio (height/length) (Adapted from (Saltzman et al., 1995)). Here, height was the distance from floor to the inferior aspect of the talar head. Length was the distance from the posterior surface of the calcaneus to the anterior surface of the first metatarsal head (Saltzman et al., 1995).

1.3.4 Ligaments of the ankle and foot

Ligaments are a form of connective tissue which hold bones together. Each of the joints and articulations previously defined are held together via ligaments. There are several ligaments that hold the ankle together. The anterior and posterior talofibular ligament connect the talus to the fibula in the front and back via the fibular malleolus to the talus at the front of the lateral articular facet. The talofibular ligaments prevent the foot from excessively sliding forward relative to the tibia. In reference to injury, this is the most commonly injured ligament (Oae, Takao, Uchio, & Ochi, 2010).

The calcaneofibular ligament connects the calcaneus to the fibula via the apex of the fibular malleolus down and back to a lateral facet of the calcaneus. This ligament limits inversion. The talocalcaneal ligament connects the talus to the calcaneus via the articular surface of the talus to a depression on the upper articulating surface of the calcaneus.

The deltoid ligament is made up of the tibiocalcaneal ligament, the anterior and posterior tibiotalar ligaments and the tibionavicular ligaments. The tibionavicular ligament connects the tibia to the navicular via the navicular tuberosity to the medial malleolus and the tibiocalcaneal ligament connects the tibia to the calcaneus via the sustentaculum tali on the calcaneus to the medial malleolus. The deltoid ligament provides stability to the hindfoot.

The spring ligament is attached to the anterior sustentaculum tali of the calcaneus to the plantar surface of the navicular. The function of the spring ligament is to hold up the bones of the MLA.

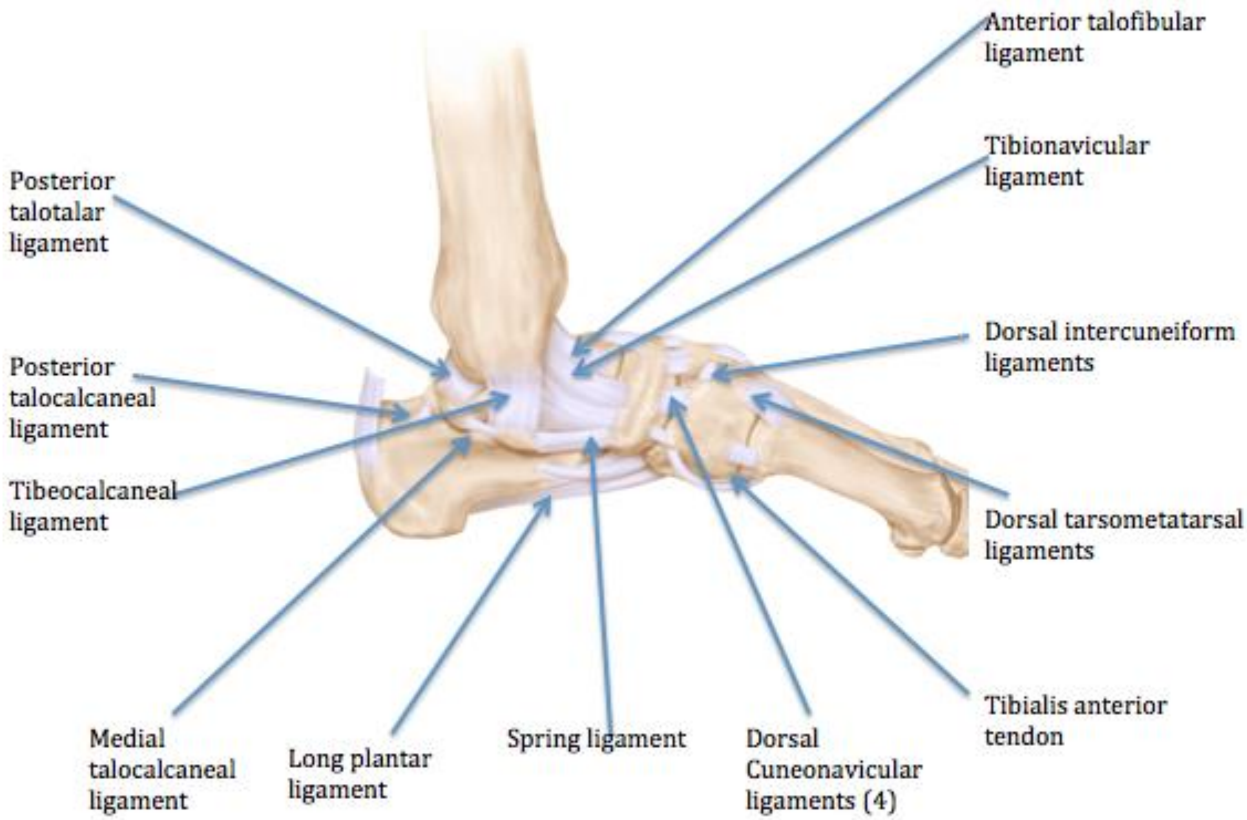


Figure 1.7 Medial view of the tendons and ligaments of the foot and ankle.

Within the midtarsal joint, there are five supporting ligaments (Figure 1.7). Stability for the calcaneocuboid joint comes from the plantar calcaneocuboid joint, the bifurcate ligament and the dorsal calcaneocuboid ligament. The talonavicular joint receives stability solely from the talonavicular ligament, and the cuboidnavicular joint is stabilized solely by the cuboidnavicular joint.

Metatarsophalangeal joints are supported and stabilized via metatarsophalangeal collateral ligaments. These ligaments limit plantar and dorsiflexion. The 1st metatarsal is also stabilized via the medial and lateral metatarsosesamoid ligaments. The 2nd to 5th

metatarsals are stabilized on the medial and lateral sides via the metatarsoglenoid ligaments. Interphalangeal joints are stabilized both medially and laterally via interphalangeal collateral ligaments.

1.4 Gait cycle

Walking and running are cyclic activities where the legs and feet repeat specific movement patterns (Nordin & Frankel, 2001). The gait cycle is the repeating pattern of gait, such as from initial ground contact with one leg to the subsequent ground contact with the same leg. There are different gait cycles that characterize walking and running. However, for both walking and running, one leg moves forward and strikes the ground, the body moves forward over top and then that leg is swung forward while the other leg is on the ground. Walking and running are characterized by the relative proportion of time that each leg spends on the ground (Nordin & Frankel, 2001). During running, both legs will spend less time on the ground, and there are two phases of double limb no support. In contrast, at least one leg is always on the ground during the gait cycle of walking.

1.4.1 Biomechanics of gait

The components of gait and the gait cycle require coordination of many movements of the legs including the hip, knee and ankle and foot (Nordin & Frankel, 2001). At initial contact, it is essential for the lower limb to be stable to allow for proper force absorption (Dugan & Bhat, 2005). This is provided by the hip adductors that remain contracted through the whole gait cycle (Dugan & Bhat, 2005).

In order for heel off to occur in the stance phase, controlled movement of the tibia occurs which allows dorsiflexion (Dugan & Bhat, 2005). It is during this phase that pelvic rotation occurs in order to create some external rotation of the stance limb (Dugan & Bhat, 2005). It is this external rotation that inverts the calcaneus and supinates the foot (Dugan & Bhat, 2005).

Toe off is caused by the contraction of the gastrocnemius and soleus to create supination in heel off that creates the rigid subtalar joint (Dugan & Bhat, 2005). In combination with the tightening of the plantar fascia, the intrinsic muscles of the foot play a large role, such as the abductor hallucis, flexor hallucis brevis, abductor digiti minimi and flexor digiti minimi (Dugan & Bhat, 2005). Muscles that cross the transverse tarsal joint also act to stabilize and solidify this joint (Dugan & Bhat, 2005). If knee flexion at this stage is not enough for the foot to clear the ground, the quadriceps muscle is activated to ensure the knee has enough flexion (Dugan & Bhat, 2005).

During initial swing, as the knee is flexed and the body is propelled forward, the iliopsoas is activated in order to create hip flexion (Dugan & Bhat, 2005). This is what allows the entire limb to swing forward. The hip abductors are the muscles creating pelvic stability during the swing phases (Dugan & Bhat, 2005). During the terminal part of the swing, hip flexion slows down to prepare for leg contact (Dugan & Bhat, 2005). Hip adductors are recruited here to bring the thigh closer to midline further stabilizing the leg (Dugan & Bhat, 2005).

1.4.2 Walking gait cycle

For each leg, walking and running are both made up of a stance phase and the swing phase (Dugan & Bhat, 2005). For walking, the stance phase accounts for 60-65% of the walking cycle while the swing phase accounts for the remaining 35-40% of the cycle (Dugan & Bhat, 2005; Nordin & Frankel, 2001) (Figure 1.8). Since each leg and foot spends more time in stance phase than in swing phase, there are two portions of the walking gait cycle where both feet are on the ground and are impacted by the opposing ground forces (Nordin & Frankel, 2001). This is called double limb support (Nordin & Frankel, 2001) and it accounts for approximately 24% of the gait cycle (12% for each instance). The rest of the gait cycle is therefore single leg support with only one foot on the ground.

The stance phase for each leg begins at initial foot contact (heel strike) followed weight acceptance (Perry, Thorofare & Davids, 1992). Of these two events, weight acceptance is more demanding on the foot (Perry et al., 1992) and would be considered the foot strike, causing the most demand on the foot and thus, the shoe. These first two events of stance phase occur during double limb support. Single limb support consists of the next two events of stance phase: mid-stance and terminal stance events.

Midstance phase consists of the single limb support that happens after double limb support (Perry et al., 1992). One foot remains planted firmly on the ground as the other foot prepares for the swing phase (Perry et al., 1992). The swing phase consists of the pre swing, initial swing, mid swing and finally, terminal swing. Terminal stance is what propels the body forwards (Perry et al., 1992). After terminal stance, there is another occurrence of double limb support (Perry et al., 1992). The limb advances and prepares the body for the next gait cycle.

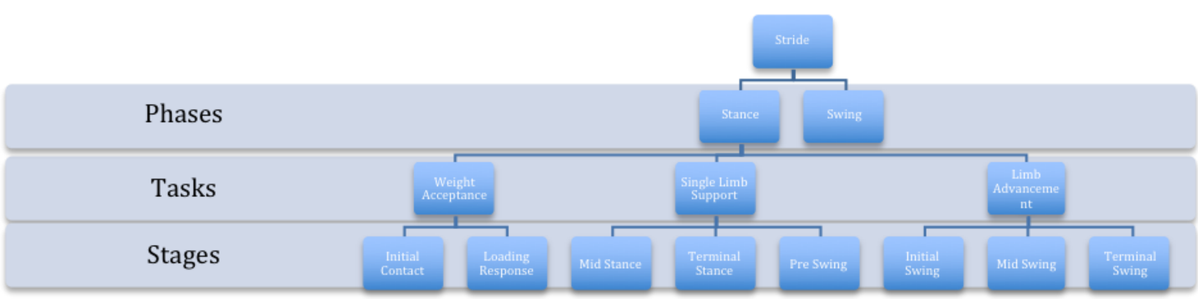


Figure 1.8: Gait cycle break down into stance and swing. The stance phase is broken down into weight acceptance and single limb support and the swing phase is also known as limb advancement (adapted from Perry et al., 1992).

During walking, the initial contact occurs at 0-2% of the gait cycle and is characterized by heel strike with the ankle at 90° (Perry et al., 1992). After initial contact shock absorption occurs. This is known as the loading response and occurs from 2-10% of the gait cycle and commences right after initial contact and continues until the foot is flat (Perry et al., 1992). This is characterized by controlled ankle plantar flexion, knee flexion and hip stabilization (Perry et al., 1992). Shock absorbing in this stage creates stability and allows for the continuation of forward movement (Perry et al., 1992). There is approximately 10° of ankle plantar flexion and subtalar valgus rolling the heel to flat foot (Perry et al., 1992).

The next stage of the gait cycle is from 10-30% of the gait cycle where the ankle is in dorsiflexion, knee extension and hip stabilization (Perry et al., 1992). This is where the body weight goes over the planted foot and contributes to limb and trunk stability (Perry et al., 1992). Midstance phase is defined as the other foot is lifted and continues until the body weight is aligned over the forefoot (Perry et al., 1992). Both the heel and forefoot remain on the ground however the tibia rotates above the ankle rocker.

After midstance, from 30-50% of gait cycle and begins with a heel rise and a free forward roll of the body forward (Perry et al., 1992). This phase continues until body weight is transferred over to the other foot or until heel strike of the other foot (Perry et al., 1992). The forefoot will be the only body weight support as the heel rises and the ankle is now in 5-10° of dorsiflexion.

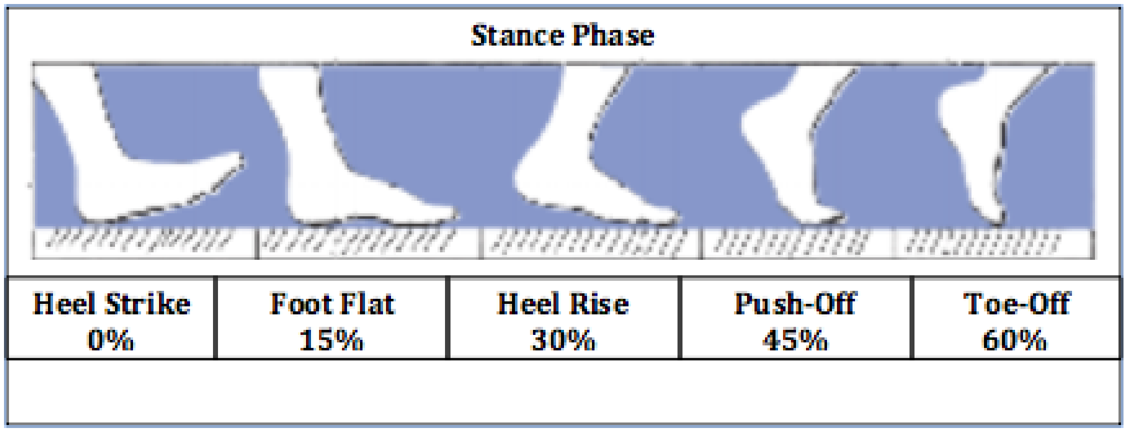
Pre-swing occurs at 50-60% of the gait cycle and is defined with knee flexion (Perry et al., 1992). It begins with initial contact on the opposite foot and ends with toe off (Perry et al., 1992). The weight transfer is the cause of the high heel rise seen (Perry et al., 1992).

Initial swing occurs from 60-73% of the gait cycle and is defined by knee and hip flexion (Perry et al., 1992). The purpose of this stage is for foot clearance and subsequent limb advancement (Perry et al., 1992). The stage begins with the foot lifting off the floor and ends with the opposite foot swinging (Perry et al., 1992).

Mid swing occurs from 73-87% and has the purpose of limb advancement and foot clearance from hip flexion and ankle dorsiflexion (Perry et al., 1992). This phase starts with the swinging limb in line with the planted limb and ends with the tibia vertical on the stance leg (Perry et al., 1992).

Terminal stance occurs from 87-100% and are defined by hip and knee deceleration, knee extension and ankle dorsiflexion (Perry et al., 1992). Here, the limb advances and prepare the limb for stance (Perry et al., 1992). This phase is characterized by the vertical tibia and ends with the foot striking (Perry et al., 1992). All of these phases would be consistent regardless if running in shoes or barefoot.

A



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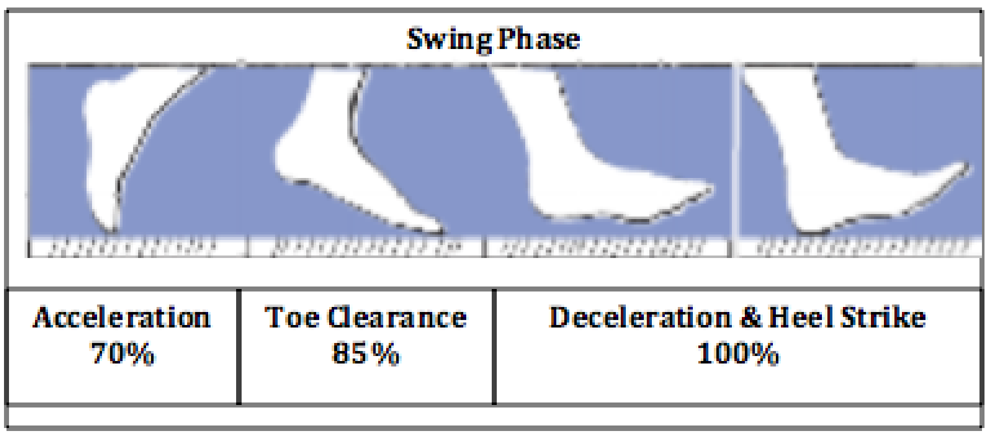


Figure 1.9: One complete gait cycle during walking shown in percentage as A) the stance phase and B) the swing phase for one foot (adapted from Nordin & Frankel, 1992).

1.4.3 Running gait cycle

The running gait cycle is different from the walking gait cycle. The stance phase has a decreased duration, accounting for 40% of the running cycle, while the swing phase takes up more of the gait cycle, accounting for 30% (Nordin & Frankel, 2001). In comparison to walking, there is another distinct component. Between the stance and the swing phase are two series of float phases, each encompassing 15% of the gait cycle (Nordin & Frankel, 2001) (Figure 1.9). The float phase is the time during running when both feet are off the ground (Nordin & Frankel, 2001).

Furthermore, the running gait cycle can be broken down past the stance phase, float phase and swing phase. Between initial contact and midstance, there are peak forces acting on the feet and lower legs. Weight is then transferred to the forefoot between midstance and toe off by the way of propulsion through pronation. Pronation of the foot during running is essential to gain the power needed to propel the body forward using the strength of the 1st metatarsal, also known as the big toe (Nordin & Frankel, 2001).

Toe off allows the body to go into it's first phase of floating. This is where both legs are off of the ground using momentum from the big toe to still move forwards followed by the other leg on the ground, starting the gait cycle again. After mid swing comes terminal swing and lastly, initial contact occurs once again. Terminal swing is the follow through forward of the swing that allows the foot to touch the ground again and start the gait cycle again.

More specifically, during the initial contact to midstance (Figure 1.9), the foot is supinated (Bates, Osternig, Mason, & James, 1978; R. A. Mann, Baxter, & Lutter, 1981). Compared to walking, after heel strike there is no plantar flexion at the ankle as the ankle immediately goes into dorsiflexion which is what allows the foot to move into pronation (R. Mann & Inman, 1964). Proper absorption during this initial contact is attributed to a conjunction of cartilage compression, joint motion and muscle contraction (Dickinson, Cook, & Leinhardt, 1985). At this stage hip and knee flexion play a large role in the absorption of ground reaction forces (R. A. Mann et al., 1981). For those who consistently run in running shoes, the initial peak force or impact peak occurs at around 10% of the stance and lasts less than 30ms (Hreljac, 2004). Also known as a braking force, this is the initial force used to slow the body down upon heel strike (Goss & Gross, 2012). In the ankle, movement of the subtalar joint allows for the ankle to absorb and dissipate many forces during running. As the subtalar joint pronates within the first 20% of the stance phase, the transverse tarsal joint axes become parallel. Accordingly the forefoot and thus the forefoot is now more mobile and can adjust to uneven terrain (Rodgers, 1988).

During the period from midstance to toe off, or the propulsion part of the stance phase, dorsiflexion occurs as the foot is now fixed to the ground (Dugan & Bhat, 2005). A maximum in dorsiflexion is seen here as soon as the tibia has gone past the ankle (Dugan & Bhat, 2005). It is also important to note that as maximum dorsiflexion and thus pronation is reached, there are many forces acting on the knee (Dugan & Bhat, 2005). As these forces come to their maximum the quadriceps and hamstrings will contract thus stabilizing the knee (Dugan & Bhat, 2005). Heel lift can occur after maximum pronation as the pelvis is forced to rotate causing the tibia to rotate externally, the calcaneus to rotate internally and supination occurs forcing the heel off of the ground (Dugan & Bhat, 2005).

Acceleration of the stance leg is initiated as the foot is now in plantar flexion (Adelaar, 1986). This plantar flexion causes the foot to go into supination, locking the transverse tarsal joint and creating a rigid base for propulsion (Dugan & Bhat, 2005). This propulsion is where the second maximum in ground reaction force occurs (Dugan & Bhat, 2005) (Figure 1.8).

Initial swing occurs as the body is propelled forward and the ground reaction forces pass posterior to the knee, forcing the knee to flex (Dugan & Bhat, 2005). Knee flexion that occurs here cancels out the need for the ankle to become dorsiflexed in order for the foot to clear the ground (Dugan & Bhat, 2005).

The second float phase occurs as the swinging leg is preparing to once again strike the ground. Ground reaction forces on the knee are at a maximum here and force the knee to flex, thus the need for contraction of the quadriceps to protect the knee joint (Dugan & Bhat, 2005). The second peak force known as the active peak occurs at about 0-75% of the stance phase and lasts considerably longer (up to 200ms) (Hreljac, 2004). This is due to the amount of force used to propel the body forward into the float phase, where both feet are off of the ground (Goss & Gross, 2012). Once the foot has touched the ground the body is prepared to commence another gait cycle.

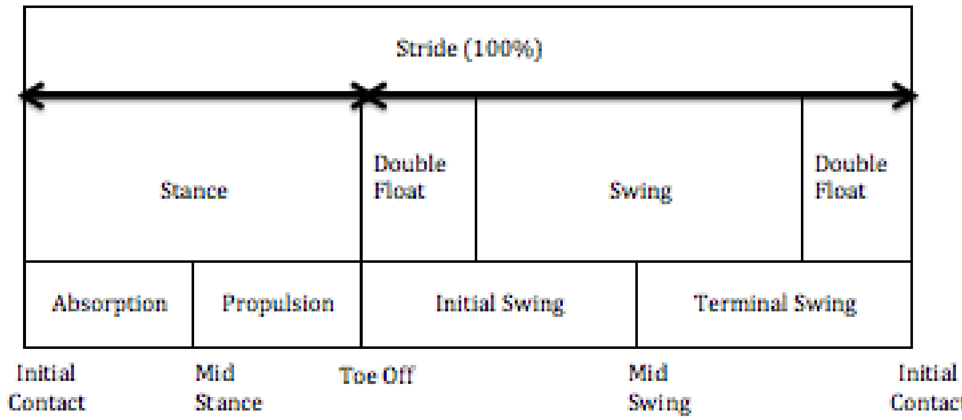


Figure 1.10 Running gait cycle as adapted from Onupuu, 1994. During running there are two series of float phases seen where both feet are off of the ground in comparison to walking where there are none. There is also an increase of time spent in the swing phase and a decrease of time spent in the stance phase.

With increasing running speeds, the forces caused by the feet and lower body during running during different phases of the gait cycle will also change (Nordin & Frankel, 2001). During a higher speed, it is common that there are two peaks of force, one at the beginning of stance phase and one at the end of approximately three to five times the body weight of the runner (Nordin & Frankel, 2001) (Figure 1.11). During lower speeds, it is common that there is only one peak force found of approximately five times body weight, and is reached near the end of the stance phase (Nordin & Frankel, 2001). Excessive impact forces, such as those found in running, can lead to injury over time such as mechanisms behind many injuries such as tibial stress fractures and plantar fasciitis and have been found to show differences between groups of injured and non-injured runners (Hreljac, Marshall, & Hume, 2000; Milner et al., 2006; Pohl et al., 2009; van Gent et al., 2007).

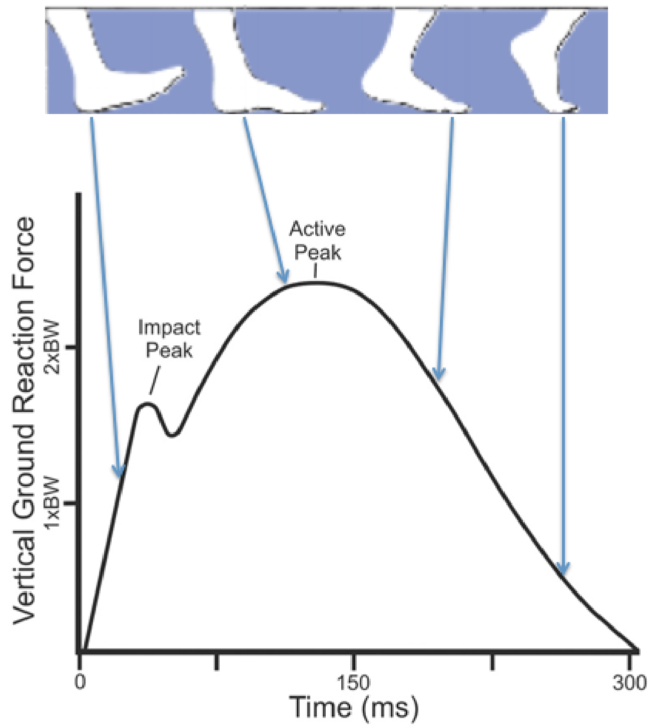


Figure 1.11 Vertical ground reaction forces seen during one gait cycle (adapted from Nordin, 2012). There are two instances were GRFs peak, once on impact and the second towards the end of stance phase, as the body is propelling forwards.

1.5 Running

1.5.1 Running patterns

There are three distinct running patterns. The first is a forefoot strike, which can be characterized by initially striking on the forefoot, followed by rocking and transferring weight to the heel (Lieberman et al., 2010). As the foot approaches the ground the ankle is plantarflexed, just like hindfoot or heel strike runners, however forefoot strikers will generally land on the outer edge of the foot landing under the 4th and 5th metatarsals. The hip and knees will remain flexed. After the initial strike, the ankle will naturally begin to dorsiflex as the weight and load are dispersed throughout the foot. This loading will force the foot to flatten and thus the arch begins to stretch. As the foot and ankle prepare to toe off, the foot will start to evert and pronation begins. Pronation is a combination of eversion, dorsiflexion and arch flattening (Dugan & Bhat, 2005). The ankle will then plantarflex as a result of the contraction of the calf muscles and the toes flex propelling the body into float phase.

The second pattern is a heel strike or hindfoot strike. This involves the last third of the foot hitting the ground upon initial contact and transferring weight to the toes (Lieberman et al., 2010). Heel striking patterns are the least commonly seen pattern among experienced barefoot and minimalist shoe runners (Lieberman et al., 2010). During a heel strike, the hip and knees are still in a flexed position, the ankle is dorsiflexed and runners typically land on the heel or just below the ankle joint. Once the initial contact has taken place the ankle will plantarflex however the arch remains unloaded as the forefoot comes down with the help of the muscles of the anterior shin. The foot will then evert and the foot finally comes to a flattened position, loading the arch. The foot is now in pronation so that the Achilles and calf muscles can now plantarflex the ankle, flex the toes and propel the body into the air into float phase.

The third running pattern is a midfoot strike. It is characterized by the foot landing flat or the midfoot hitting the ground upon initial contact followed by transferring weight to the heel (Lieberman et al., 2010). Midfoot strikers have the same amount of hip and knee flexion as both forefoot and hindfoot strikers. In regards to the initial contact, contact is going to take place when both the heel and forefoot land simultaneously. As the initial strike happens, the foot is already flattened so the arch is loaded immediately and the ankle goes into plantar flexion and propels the body forwards into float phase and prepares the body for another gait cycle.

It is also important to note that there is a decreased ground contact time seen in runners with a forefoot strike and midfoot strike (De Wit, De Clercq, & Aerts, 2000; C Divert et al., 2008; Nunns et al., 2013). Forefoot striking also increases pressure put on the metatarsal heads (Altman & Davis, 2012). This decrease in ground contact time can be attributed to the fact that during forefoot striking and midfoot striking there is less dorsiflexion at the ankle. With less dorsiflexion occurring, the ankle becomes flattened quicker and thus the arch becomes loaded sooner. This means that the foot is able to plantarflex and propel the body forward sooner preparing the legs for the next gait cycle.

Although, characterizations in strike patterns have been studied in relation to energy expenditure and injury rates, some studies have shown that there is no direct benefit from transitioning a strike pattern either from forefoot to rearfoot, or from rearfoot to forefoot (Hamill & Gruber, 2017). Perhaps, changing footstrike pattern can actually lead to injuries sustained from stressing the feet (Hamill & Gruber, 2017). This study identifies that foot strike pattern appears to depend on the type of run, as running longer distances may cause a rearfoot strike and faster, shorter runs may cause a forefoot strike (Hamill & Gruber, 2017). On a harder surface, runners were less likely to rearfoot strike, have a higher step cadence and thus, were able to run faster (Lieberman et al., 2015).

1.5.2 Barefoot runner's patterns

Experienced long distance barefoot runners most often have similar forefoot running patterns (Lieberman et al., 2010). Forefoot runners will exhibit a flatter foot placement (ie. more plantar flexion, less dorsiflexion) and the flatter the foot, the smaller pressure on the heel (De Wit et al., 2000). The ankle is more plantarflexed at landing (Squadrone & Gallozzi, 2009) and there is an increase in vertical arch height during stance (ie. more positive MLA height-to-length ratio) during ground contact (Perl, Daoud, & Lieberman, 2012). For forefoot runners, rearfoot eversion was lower than that of rearfoot strikers (Pohl & Buckley, 2008). where they land on the ball of the foot before transitioning to the heel (Lieberman et al., 2015) or the heel never touches the ground (Hamill & Gruber, 2017). It has also been shown that barefoot runners exhibit a higher cadence or stride rate compared to traditional shod runners (Moody, Hunter, Ridge, & Myrer, 2018).

Runners are likely to run barefoot if they have grown up barefoot running such as tribesmen in remote locations in the world, those who have transitioned from minimalist shoes into full barefoot running, and those who believe it is better for their feet. Many runners currently wear or have tried training in minimalist shoes. This has motivated the purpose of this study: to see whether barefoot mimicking shoes are significantly different than running in regular cushioned running shoes.

1.5.3 Shod Runner's patterns

Shod runners are runners that wear running shoes that are typically cushioned. They often initially strike the ground with the hindfoot, followed by the ball of the foot (Lieberman et al., 2010). Research suggests the reason that shod runners rearfoot strike is that this may have a lower metabolic cost associated for longer distances (Hamill & Gruber, 2017).

Rearfoot runners strike the ground in inversion and change to eversion at midstance (Stackhouse, Davis, & Hamill, 2004). At heel contact, the heel and forefoot will be slightly supinated, that is, inversion, dorsiflexion of the ankle and flattening of the arch (Rodgers, 1988). At midstance, the foot will be pronated from 55-85% of support phase and the foot will supinate and return to neutral around 70-90% of support (Rodgers, 1988). The foot will be supinated for push-off (Rodgers, 1988) and the ankle is more dorsiflexed upon landing (Squadrone & Gallozzi, 2009).

1.5.4 Impact force & injury

Impact forces upon initial contact cause sudden, large forces to travel up the body (Milner et al., 2006; Pohl et al., 2009; van Gent et al., 2007) (Figure 1.11). Repetitive high loading on the joints and tissues are thought to explain many injuries such as tibial stress fractures and plantar fasciitis. The ground reaction force experienced during running can be nearly double that of walking, such as 1.5-2 times body weight during running (Milner et al., 2006). This initial peak force is seen in the first 50ms of the heel strike (Bobbert, Schamhardt, & Nigg, 1991). During the strike phase, the internal leg muscles are lengthened, creating forces and stress (Bobbert et al., 1991).

Although there is a decreased force associated with barefoot running, shorter strides leads to an increased cadence or frequency of foot strike, so any potential decreases in force upon impact associated with barefoot running could be counteracted by the increased frequency of foot strike and thus, potential injury (Collier, 2011). A study by Shih et al., (2013), found that barefoot runners had a significantly higher cadence than shod runners.

1.6 Injury

Approximately 24-65% of runners are injured annually between training errors, slips, falls and repeated strain (Macera et al., 1989). Injured tissues include bone, connective tissue, ligaments and muscles of the foot and leg (Macera et al., 1989). These injuries include, but are not limited to, stress fractures, plantar fasciitis, tendonitis, iliotibial band syndrome and knee tracking problems.

One study, described 2002 running injuries and determined that the four most common sites of injury ranked from highest to lowest were the knee (42.1%), foot and ankle (16.9%), lower leg (12.8%), hip and pelvis (10.9%) with the top three injuries in this study were patellar femoral pain syndrome, iliotibial band friction syndrome, and plantar fasciitis (Taunton et al., 2002). As the feet are exposed to approximately 10,000 steps an hour, poor biomechanics, such as excessive pronation or supination during heel strike, can lead to injury over time (Macera et al., 1989).

In a study done by Messier et al., 2018, 300 runners were studied over the course of a 12 month training period. The findings were that at least one injury occurred to 66% of the runners within the initial 12 months and more interestingly, more women than men were injured (Messier et al., 2018). The most common sites for injuries were the knee and foot (Messier et al., 2018).

1.6.1 Training errors

Training errors play a large role in injury occurrence (Shorten, 2000). These are most acknowledged as anything that causes an increase load being placed on the body, such as sudden increases in duration and intensity, constant intense training, a single intense training day or race and/or sudden return (Shorten, 2000). Training errors can be attributed to many running injuries, specifically repetitive strain injuries (Shorten, 2000). Cross training, small mileage increases (of about 10% per week) and most importantly rest can help to prevent overtraining, training errors and repetitive strain injuries (Shorten, 2000). A study by Altman & Davis (2016), was one of the first scientific study to look at habitually shod and barefoot runners over the course of a year and track rates of injury. The study noted that there was a significantly lower number of injuries in the barefoot group, although when normalized for distance, there was no significant difference between the two groups (Altman & Davis, 2016).

1.6.2 Intrinsic & extrinsic factors to injury

Running injuries can be caused by a variety of intrinsic and extrinsic factors (Taunton et al., 2002). Intrinsic factors are those that are concerned with the runner themselves, such as experience or anatomical limitations, while external factors are training errors, old shoes and unpredictable running terrains (Taunton et al., 2002).

Intrinsic factors arise due to repetitive strain from overloading or due to poor running biomechanics (Richards et al., 2008). Poor biomechanics may include over pronation and supination, pes planus or flat footedness, pes cavus or high arches. Pes cavus can change movements of the hindfoot and forefoot as it is typically caused by a muscle imbalance of the leg and foot causing the foot to plantarflex and become more inverted (Wapner, 2000). Pes planus, or flat footedness is characterized by having a low medial longitudinal arch. It is thought that those who have flat feet are prone to having less transverse plane forefoot motion, less adduction at toe off and plantar flexion of the hindfoot instead of dorsiflexion (Taunton et al., 2002).

Extrinsic factors are those that are concerned with the environment, such as running terrain, type (i.e. long distance, sprinting, racing, etc.), and age of shoe (Taunton et al., 2002). Running terrain includes running on trails, sand, concrete and asphalt. If the terrain is bumpy or sloped to one side for a large part of a run, over a prolonged period of time the hips could be uneven, causing repetitive stress on one side of the knee, causing pain such as iliotibial band syndrome (Taunton et al., 2002). Running on sand inhibits the ability to effectively arch the foot required to propel the body forward (Taunton et al., 2002). This is because as the body is arching the foot trying to pronate to excel forward, the sand underneath starts to give, causing the arch to work harder to propel. Concrete is going to be harder to run on for the body, as there is very little give and therefore any force put on the concrete will come right back up into the lower limbs (Taunton et al., 2002). This repetitive stress can cause things such as shin splints, iliotibial band syndrome, etc. Asphalt has a little more give to it as it is not as dense. This means that as the body exerts force, some of it can be dissipated into the ground and less comes back to the lower limbs (Taunton et al., 2002).

1.6.3 Mechanisms behind injury

High instantaneous loading, (during heel strike and toe off), high ground reaction forces and lower medial longitudinal arch heights were shown to be some key factors in the development of plantar fasciitis (Pohl et al., 2009). As ground reaction forces and loading increases, so does the strain placed on the plantar fascia, potentially resulting in plantar fasciitis over time (Pohl et al., 2009).

Excessive varus and valgus movements, eversion, and tibial rotation have been suggested to cause injury in long distance runners (James, Bates, & Osternig, 1978; B. M. Nigg, 2001). Excessive movements can occur in any form; shod, barefoot or in barefoot mimicking shoes can lead to repetitive strain injuries (B. M. Nigg, 2001). Correcting alignment of the lower limb and ankle through proper running shoe design, orthotics or other inserts could correct these risk factors (B. M. Nigg, 2001). Having correct alignment of the body can reduce or eliminate excessive movements (B. M. Nigg, 2001).

As the majority of the population does run shod, studies concentrate on this group of runners. Within the past few years as barefoot and minimalist running becomes more popular, more studies have emerged that looked into barefoot running and minimalist running. One study by Decker, Torry, Wyland, Sterett, & Steadman, 2003, has shown that barefoot running and minimalist shoes cause a decrease in hip and knee flexion which have been linked to increases seen in ACL injuries. This decrease in hip and knee flexion can be attributed to the decrease in stride length, suggesting that the lower leg is more centred beneath the body on foot strike (Soares et al., 2018). Another study by Sinclair, Butters, & Stainton, 2018, found that running in barefoot and minimalist shoes cause an increase in knee adduction on instantaneous loading. This medial knee compartment loading is a strong predictor for knee osteoarthrits (Sinclair et al., 2018).

1.7 Anatomy of the running shoe

Athletic shoes typically consist of a leather upper portion, but other fabrics such as nylons or microfiber may also be used (Mcpoil, 2000). The upper portion may also consist of a combination of different materials to create various effects of stability, flexibility and motion control. The cushioned midsoles are made of EVA foam with inserts of gels, dual density foam, air or fluids in attempts to increase cushioning (Mcpoil, 2000). These are seen is models such as the Nike Air. The use of cushioning systems in modern day running shoes, help to attenuate impact and ground reaction forces acting on the foot (Mcpoil, 2000). However, cushioning systems do not completely absorb forces causing the body to bear the remaining impact repeatedly over the course of a bout (Mcpoil, 2000).

Heel wedges are made up of polymeric foam and the outsole is made of hard rubber (Frederick, 1984). Reinforcements are common throughout the rest of the shoe, mostly in places such as the toe, heel and midsole (Mcpoil, 2000). This helps lengthen the longevity of the shoe and thus, the time to replace is also lengthened. Running shoe design has the ability to affect both speed and energetics based on factors of age, model, materials and quality of the shoe (Frederick, 1984).

1.12 Anatomy of the standard running shoe. A strong shank of a shoe can increase torsional strength and aide in motion control.

1.7.1 Heel counter & wedge

Most running shoes are designed with a high wedge at the heel, located underneath the midsole (Lieberman et al., 2010) (Figure 1.12). This forces the runner to have a rear foot strike (Lieberman et al., 2010). The heel strike is forced as there is a heel-toe offset whereas the heel is higher than the forefoot, naturally guiding the runner to heel striking. Many shoes have arch supports and/or stiff soles that decrease the runner's ability to strengthen the intrinsic muscles of the foot (Lieberman et al., 2010). This potentially leads to injuries such as plantar fasciitis (Lieberman et al., 2010). Although heel wedges are shown to potentially be the cause of some injuries, shoes have a heel wedge as this is where shock absorbing materials are found, providing cushioning to the heel. This cushioning is vital as repeated impacts such as those found in exercise and more specifically running help protect the heel.

The heel counter supports the back of the ankle and Achilles during running and activity and is located between the heel tab and the midsole (Figure 1.12). The heel counter aids in ensuring a good fit between the shoe and the foot, preventing excessive motion during dorsiflexion of the foot at heel strike also known as motion control (Mcpoil, 2000).

1.7.2 Motion control

Both anti pronation and motion control has been a factor in running shoe design since the early 1980's (Shorten, 2000). It was originally incorporated into shoes via heel wedging or flaring the heel (Stacoff, Nigg, Reinschmidt, van denBogert, & Lundberg, 2000). Motion control aids in preventing the foot from excess supination or pronation of the foot during ground contact. The foot needs a level of pronation in order to propel the body forward as the heel strikes, the foot pronates, and then supinates along the outside edge and ends in pronation by pushing off of the big toe. This motion control is accomplished by adding stiffer cushioning, heel counters, insoles and wedges (Shorten, 2000). Repeated and excessive pronation or supination of the foot during running can result in repetitive strain injuries such as plantar fasciitis and iliotibial band syndrome (Shorten, 2000). Individuals with pes planus, or flat feet are more likely to over pronate, requiring a stabilizing shoe, while individuals with pes cavus, or high arches are more likely to have reduced pronation requiring a cushioned shoe (Shorten, 2000).

Another form of motion control comes from traction of the outersole (Mcpoil, 2000). Without traction, activities performed could be dangerous especially over difficult terrains (Mcpoil, 2000). This suggests the idea that shoes should be monitored on a regular basis for wear, as treads will degrade over use, time, and frequency, thus signalling to the user to replace them.

Another motion control factor is the motion control over the rear and midfoot (Mcpoil, 2000). This is because foot pronation is controlled by heel stabilization and midfoot support (Cheung, 2009; Mcpoil, 2000). The forefoot region has to be stabilized because side to side motion can lead to instability as well as stretching the upper portion of the shoe potentially causing excessive shear and compressive forces which affects the medial and lateral aspects of the foot and the balls of the foot (Mcpoil, 2000). Cushioning is used in the midsole to decrease impact forces however, they do not decrease the magnitude of the forces (Cheung, 2009; Mcpoil, 2000). Cushioning for overpronators is typically softer on the lateral sides, and harder on the medial aspect of the midsole. This aims to attenuate and dampen some of the initial ground contact force and control over pronation (Cheung, 2009).

1.7.3 Shoe use

Mcpoil, (2000) suggests that runners be vigilant about recording shoe usage. Under each step, the midsole of the shoe undergoes stress and compression (Mcpoil, 2000). Over the course of a run, the compression can decrease midsole shock effectiveness by 25-30% (Mcpoil, 2000).

Many people are not aware of the miles they run, weather they train in or surfaces trained on (Mcpoil, 2000). All of these factors have an impact on shoe durability and lifespan (Mcpoil, 2000). Wearing shoes that have had too many miles run on them mean that the cushioning systems may be ineffective at helping to dissipate ground reaction forces, thus more force is acting on the joints (Mcpoil, 2000). This could have implications for injury so runners should be vigilant about recording mileage to help decrease the likelihood of injury.

1.7.4 Lacing design of running shoes

The way that shoes are laced and the placement of the lacing eyelets are an important design of all running shoes (Hong, Wang, Li, & Zhou, 2011). Most importantly, laces allow the foot to be secured into the foot bed or the sole of the shoe and against the heel counter (Hong et al., 2011). This also allows the individual user to create a custom fit to the shoe and increases comfort (Frey, 2000).

It has been suggested the purpose of laces are to create a better fit and more importantly, distribute stress evenly over the dorsum of the foot (Hong et al., 2011). Hong et al. (2000), also found that lacing conditions have a significant influence on shoe-foot coupling during running. That is, the better laced they are for a specific foot type, the better the shoe will fit. The same goes for the more the lacing technique does not match a foot type, the worse the shoe will fit.

1.8 Barefoot simulation shoes

1.8.1 Pressure and ground reaction force during barefoot running

Ground reaction force is the force the body exerts when the foot strikes the ground. Barefoot running can be characterized as having increased loading rates due to minimal external protection and shock absorption as compared to running shoes (De Wit et al., 2000). The flatter the foot is upon impact, the smaller the pressure on the heel as the foot better dissipates forces upon impact (De Wit et al., 2000). This suggests that perhaps barefoot runners aim to flatten their feet as they run to diminish heel impact (De Wit et al., 2000). There is also evidence that barefoot running creates the highest vertical ground reaction forces (De Wit et al., 2000). These high vertical ground reaction forces repeated over time can be a leading cause of injury for both barefoot runners and minimalist runners.

1.8.2 Stride frequency and impulse

Barefoot running was shown to increase stride frequency, lower contact on the ground, decreased flight time, and quicken braking time (C Divert et al., 2008). Stride length was shorter and stride frequency was increased with barefoot running (Squadrone & Gallozzi, 2009). This was found to be immediately assumed when either barefoot or minimalist shoes were worn (Soares et al., 2018).

Pressure under the toes was significantly higher with VFF compared to barefoot running (Squadrone & Gallozzi, 2009). Perhaps this can be attributed to the fact that the VFF have a thick rubber sole, limiting it's ability to dissipate pressure compared to running barefoot. This could have injury implications for the metatarsals, as increases in pressure over time can be a factor in the development of a repeated strain injury.

1.8.3 Characterization of a barefoot running shoe

Barefoot running shoes or minimalist shoes or barefoot mimicking shoes seek to create a thinner, more flexible rubber sole (Lieberman et al., 2010). This sole acts to protect the foot from any abrasions from the ground (Figure 1.13). Bonaccini, et al. (2013), classify minimalist shoes as having a low profile, increased sole flexibility, reduced offset between the heel and forefoot (<4 mm), little motion control, lightweight and the minimalist part really coming from having very little cushioning and motion control (Bonacci et al., 2013; Cauthon, Langer, & Coniglione, 2013). This small heel-toe offset results in runners adopting a forefoot strike pattern. Other characteristics a light outer sole and little or no heel counter (Mcpoil, 2000). VFF do have an upper, however it acts solely to hold the foot to the foot bed, as does the Velcro strap. The Velcro strap also holds the heel to the back of the shoe so the toes do not slide forward during running.

Figure 1.13 VFF barefoot mimicking minimalist shoes. These shoes typically contain little to no support however offer a protective barrier between the sole of the foot and the ground.

Nigg, (2009), suggests that there is indirect evidence that barefoot running strengthens various muscles crossing at the ankle joint as well as big muscle groups (i.e. quadriceps and hamstrings) and small muscle groups (i.e. peroneus longus and soleus)(B. Nigg, 2009). Running shoes use more energy as the acceleration and deceleration power required with the weight of a shoe, energy used to deform the shoe during running, energy used to rotate the shoe on ground contact, energy absorbed in the midsole and energy lost in stiffness of the joints of the foot (B. M. Nigg, 2001; Warburton, 2001).

1.8.4 Hazards of barefoot running

The largest external hazard of running barefoot (and in barefoot simulated (i.e. minimalist) running shoes) is the potential of injury caused by debris (Murphy et al., 2013). This includes nails, glass, wood chips (slivers) and stones (Squadrone & Gallozzi, 2009). A study by Lieberman et al., (2016) found that habitually barefoot runners exhibit significantly more injuries to the plantar surface of the foot.

In general, it is thought that those transitioning from a cushioned midsole to either barefoot running or barefoot simulated running shoes may take a longer time than those simply transitioning from a minimalist shoe to barefoot running (Murphy et al., 2013). In order to properly transition to barefoot running, barefoot or minimalist shoes should slowly be integrated into the workout, just as a new exercise would be (Murphy et al., 2013).

1.8.5 Why transition to barefoot running?

Barefoot enthusiasts and historical barefoot runners claim that running barefoot strengthens the intrinsic muscles of the feet and that shod running makes the intrinsic muscles of the feet potentially weaker (Rossi, 1999; Wikler, 1961). It is thought that even with greater improvements in cushioning and motion control, injury rates are not declining (Lieberman et al., 2010).

Theories behind advantages of barefoot running include; a reduction in impact forces (C Divert, Mornieux, Baur, Mayer, & Belli, 2005; Caroline Divert, Baur, Mornieux, Mayer, & Belli, 2005; Squadrone & Gallozzi, 2009), reduced oxygen consumption (Burkett, Kohrt, & Buchbinder, 1985; Catlin & Dressendorfer, 1979), increased proprioception (Lieberman et al., 2010), and increased intrinsic muscles (Rao & Joseph, 1992).

1.9 Studies and kinematic outcomes

1.9.1 Previous studies

Previous barefoot and shoe testing methods have used external markers on the shoe (Morio, Lake, Gueguen, Rao, & Baly, 2009; Stacoff et al., 2000). This has been shown to overestimate skeletal movements within the shank and foot (Cappozzo, Catani, Leardini, Benedetti, & Della Croce, 1996; Reinschmidt et al., 1997). Two dimensional studies have also been conducted, by utilizing reflective markers and simply taking pictures in different planes (Areblad, Nigg, Ekstrand, Olsson, & Ekström, 1990). This is affected by the alignment of the foot relative to the camera position as there is very little consistency between subjects (Areblad et al., 1990).

Another study was conducted with intracortical Hofmann pins (Stryker Howmedica AB Sweden, 3.2 mm diameter, #5038-5-80) with reflective markers attached to the ends. These pins were inserted in the calcaneus (Morio et al., 2009). This invasive and painful study required surgical intervention, allowing for accurate results but few subjects. This study suggests that various shod conditions alter the foot motion pattern during the push off phase of gait (Morio et al., 2009). However, this study only compared the differences between shod and barefoot conditions (Morio et al., 2009). This model simply used 17 markers that were placed over various anatomical landmarks, to create two segments (Morio et al., 2009). It was found that shoes constrained eversion/inversion and abduction and adduction of the foot (frontal and horizontal planes respectively) (Morio et al., 2009). These findings are aligned with Wolf, et al. (2008) that found footwear constrained foot motion in forefoot spreading and foot pronation during push off. They concluded that having room for the forefoot to spread was crucial to mimic barefoot gait (Morio et al., 2009).

1.9.2 Current testing methods

Currently according to Fong, et al. (2013), there is a gap within the literature surrounding shod and barefoot running. They have suggested that barefoot running be compared to shod conditions to attribute potential differences in motions of the foot, running patterns or injury rates (Fong, Sinclair, Hiller, Wegener, & Smith, 2013). There is also little literature to support the claims that minimalist shoes are different from traditional running shoes and that they mimic barefoot running (Bonacci et al., 2013). Bonaccini, et al. (2013), did find however, that barefoot running increases the work done at the ankle and that minimalist shoes may not mimic barefoot running (Bonacci et al., 2013). It is important to note that shoes for this mentioned study were not qualified barefoot mimicking shoes-they contained elevated heels and cushioning. The same study did confirm that barefoot running does change biomechanics of traditionally shod runners (Bonacci et al., 2013).

Literature has studied females and VFF (McCarthy, Fleming, Donne, & Blanksby, 2014). This study showed that it is possible to change motor patterns in running over a 12 week training period (McCarthy et al., 2014). This study however, did not measure kinematics of the MLA of hindfoot/forefoot pronation/supination (McCarthy et al., 2014).

A study looking at midsole thickness and running patterns found that angles in the sagittal plane between the ankle and the foot is lower in barefoot versus shod conditions (Chambon, Delattre, Gueguen, Berton, & Rao, 2014). The shod shoes studied were of various midsole thicknesses however, they all had a 0 mm heel-toe offset (Chambon et al., 2014). Knee and ankle flexion range of motion was also lower in the barefoot conditions which could be a factor in developing shin splints or plantar fasciitis (Chambon et al., 2014; Taunton et al., 2002).

Another study looking at kinematic and kinetics between shod and minimalist shoes, found that minimalist shoes had greater knee flexion and dorsiflexion upon foot strike during running (Willy & Davis, 2013). Impact peaks and vertical loading rates were also greater while wearing minimalist shoes, thus suggesting an increased rate of overall loading while running in minimalist shoes vs shod which could lead to stress fractures, shin splints, etc. (Taunton et al., 2002; Willy & Davis, 2013).

In terms of increasing the likelihood of injury, (Ryan, Elashi, Newsham-West, & Taunton, 2013) found that semi minimalist shoes (Nike Free 3.0 V2) and full minimalist shoes (Vibram Bikila) to cause more injury than regular shod running shoes. More specifically, the Vibram Fivefingers caused more shin and calf pain in runners training over a 12 week period (Ryan et al., 2013).

When running in shoes with heel-toe offsets of less than 6 mm, it is recommended that there is a gradual introduction of minimalist or barefoot shoes as well as gradually increasing mileage as an injury prevention mechanism (Cauthon et al., 2013).

1.9.3 Vibram Five Fingers & Barefoot running

Only one study has compared the VFF (classic model) and barefoot running (Sqaudrone, 2009). This study was the first to use habitual barefoot runners and looked at sagittal plane kinematics using a treadmill (Sqaudrone, 2009). Measuring only sagittal plane kinematics possibly misses the majority of the motion of the foot as many motions of the foot such as pronation, involves eversion (located in the frontal plane), abduction (located in the transverse plane), and dorsiflexion (located in the sagittal plane). This study also analyzed pressure distribution between Vibrams and barefoot conditions (Sqaudrone, 2009). Data was collected at 60 Hz, however markers used for digitization were simply mounted on the lateral condyles of the knees, head of fibula, lateral malleoli, tuber calcaneum and 5th metatarsal (Sqaudrone, 2009). The same study found that 60 Hz was sufficient when collecting kinematic running data at 12 km/h (Squadrone, 2009; Van Gheluwe, 1995). The study by Squadrone (2009), calculated angles between the foot and ground, knee and ankle joints, peak knee flexion and ankle plantar flexion. No specifics between the MLA angles and forefoot and hindfoot pronation/supination were studied. Contact time was significantly decreased in Vibrams than shod running, and the magnitude of impact peak forces were significantly higher in shod than Vibram or barefoot conditions (Squadrone, 2009).

1.9.4 Summary of kinematic outcomes

Many studies have tried to measure and quantify foot strike patterns during running and walking. Few have been able to quantify movements between segments of the foot in order to determine hindfoot/forefoot movement and pronation/supination.

Studies done by Lieberman et al. (2010), studied the effect of strike type on GRF impacts by using an infrared camera system with skin-mounted markers. They found that habitually shod runners adopt a flatter foot on impact between barefoot and shod runners. They did not look at forefoot, hindfoot or midfoot movement between the two groups.

A study by Morio et al. (2009) used a two segment foot model (forefoot and rearfoot) and compared barefoot and shod runners running in barefoot, shoes and sandals. They found that barefoot runners adopt more eversion of the forefoot and that running in sandals decreased the amount of eversion. They concluded that the soles of standard running shoes affect and constrain inversion, eversion and adduction during impact and throughout stance phase. They also saw that shoes constrain natural foot movement compared to barefoot running. Studies done by Morio et al. (2009) and Wolf et al. (2008) both supported the idea that shoes affect the amount of forefoot spreading and pronation during toe off.

A two dimensional study done by Bates et al. (1978) determined that there was a longer time spent in pronation, and that pronation began sooner and finished later, in the stance phase during barefoot running. Another two dimensional study by De Wit et al. (2000) used video tape and skin mounted markers determined that during barefoot running, there is a significantly smaller initial eversion at impact, a higher step frequency and a flatter foot at touchdown.

See Appendix D for a table summarizing the kinematic outcomes of the various studies.

Chapter 2: Kinematic Models

Kinematics of the foot and ankle are simply concerned with the movements that are occurring without paying attention to the loads or forces causing movement. Kinematic studies are useful for collecting information about complex anatomy.

2.1 Kinematic models

2.1.1 Fluoroscopy and X-ray

Radiographic images or x-rays are produced via film and are typically static in nature. These images show any bone or boney fragments within the scope of view. Fluoroscopic images are produced via a live feed, thus able to film video and also show bone or boney fragments. In both x-ray and fluoroscopy, depending on the angle, field of view and/or location, it may be difficult to locate specific bones or landmarks accurately. Fluoroscopes can be combined to create 3D video via bi-planar fluoroscopes. One of the best advantages of this method are instant feedback in the form of video or pictures. Video and static pictures can be digitized at the time of capture and there is no film used. Anatomical landmarks are tracked with digitized markers and can be compared to skin mounted markers. Bi-planar fluoroscopes can track 3D images therefore, this method of kinematic analysis is extremely accurate as specific landmarks can be seen quite easily and compared from trial to trial to see differences. The lack of markers required decreases placement error and leads to more accurate data collection and analysis. Exposing subjects to radiation is an issue with this method, as many trials are required to get specific landmarks to appear in the field of view as the field of view is quite limited. Post processing time is time intensive, as bone models must be created via CAT scans and matched to static images from trials. Bones hidden from the field of view or stacked bones may be difficult to see.

2.1.2 Optical motion capture

Optical motion capture requires the use of multiple motion capture cameras mounted on the wall to track 3D movements and static images. They use reflective or light based markers to locate landmarks on the body. Using infrared or visible red light, markers are located within the field of view. Markers must be seen by at least 2 cameras in order for the marker to appear as images appear in 3D. Using specialized software, 2D views or static images use direct linear transformation to integrate and create a 3D view or views available in the x, y and z planes of movement. This 3D transformation allows markers to be identified in 3D space and thus movements can be compared between trials. Optical motion capture systems are capable of collecting at rates of up to 500Hz. They also have many benefits including the use of analog data to capture other information simultaneously such as EMG or force plate data. Accuracy with this system is dependent on the size of the area, number of markers, field of view and the number of cameras being used. This is overall an accurate way of collecting kinematic data. There are many advantages to using an optical motion capture system. Firstly, there is no wire interference as passive markers are used. This allows for faster and more complex motions such as running and jumping to occur with no concern over interference from equipment. There is no electromagnetic interference as seen in other optical motion capture and cameras are not influenced by light. One of the disadvantages of this method is there can be too many markers in one area and the camera may not pick up on all them in a cluster and a long post processing time associated with data analysis. Markers must be manually located if they are in a cluster or frames must be deleted if too many markers are missing from the field of view. If all the markers are visible this method is accurate in comparing any differences present.

2.1.3 Video analysis

Singular or multiple video cameras can be set up at varying heights and angles to capture kinematic data with or without markers. Speeds and landmarks are typically tracked with video analysis. Video is typically captured at 60Hz. Once captured, still frames are digitized using software used to locate each of those markers in space. This process is lengthy compared to that of optical motion capture. Similar to option motion capture, linear transformations are used to take multiple 2D images and convert them into 3D marker positions. Advantages and disadvantages are the same as optical motion capture, however two advantages of video analysis are that it is a more portable method, allowing data to be collected anywhere a camera would fit and secondly, video capture is a cost effective way to capture kinematic data. This allows less constraint compared to a lab setting and data can also be collected at competitions, practice, etc.

2.1.4 Electromagnetic tracking

Electromagnetic tracking uses a low frequency electromagnetic field to measure and locate positions of segments. A static transmitter is fixed to a lab frame of reference and is required to be relatively close to the area being studied. Receivers are required to be attached to the segments of interest. Interactions between the electromagnetic fields allow the 3D position of segments to be determined within a field of view. The positions are determined by the interaction of the electromagnetic field and the receiver, compared to the 3D orientation of the transmitter with the receiver. This system requires no markers and no markers need to be seen to collect data. Lighting within the data collection room does not interfere and data collection is relatively quick. There is great resolution with this system allowing for a more accurate digitization process as guessing landmarks is greatly reduced with a higher resolution. There is a large interference with any metallic objects as well as interference with the subject as there are wires connecting markers to the body. Complex calibrations are required at the beginning of each data collection and limitations on the space between the transmitter and receivers.

2.1.5 Bone pins

All of the data collection methods so far can benefit from utilizing skin mounted markers of some form. The issues with skin mounted markers is there will always be some movement between the skin to which the marker is attached and the underlying bone. One way to decrease this error is by surgically inserting metallic pins into desired locations on the bones that protrude through the skin. Pins such as Hoffman pins (Stryker Howmedica AB Sweden, 3.2 mm diameter, #5038-5-80) are typically used. These stick out of an incision point and insert into the bony landmark to be studied. These bone pins landmarks points to be studied in order to reduce error in capturing data and during post processing. Beads can also be implanted into the bones during surgery to be studied in the future. Once pins or metal beads are inserted, methods such as fluoroscopy and x-ray can be utilized to get an extremely accurate comparison of movement of bones underlying the skin. This is the gold standard in terms of motion comparison. However, this method is extremely invasive and potentially painful. Complications such as infection and loose beads in joint spaces as beads can be smaller than 1 mm can be detrimental to patients.

2.1.6 Skin motion artifact

Skin motion artifact is when markers mounted on the skin move differently compared to their underlying landmarks. This is one of the greatest sources of error in motion capture (Andriacchi, Alexander, Toney, Dyrby, & Sum, 1998; Cappozzo, 1991). There are a few sources of error due to skin motion artifact. The first being misplacement of markers on anatomical landmarks.

Palpation of landmarks to place skin mounted markers creates a lot of error. If the markers are placed incorrectly over landmarks, all data collected with that landmark as a result will be inaccurate. Inertial effects or landmarks moving around during trials affects accuracy of data collection. Most markers protrude and bounce during kinematic trials and thus, can be knocked with other body parts. Bouncing of markers will falsify resulting landmark movement under the skin, Skin deformation over anatomical landmarks can also create a false sense of landmark movement as skin can become taut or reverse, relaxed and sag down. Sliding often occurs close to joints and skin has more movement here and muscle contraction forces markers to draw in different direction (Cappozzo et al., 1996; Leardini, Chiari, Croce, & Cappozzo, 2005).

Skin motion artifact is going to occur in any studies of motion besides that where intracortical bone pins or beads are inserted into specific landmarks. This further emphasizes the need to keep the markers in exact locations between trials. Skin motion artifact cannot be reduced with skin mounted markers, however there are ways to reduce added error. This is a drawback of using a multi segment foot model to analyze movements of the feet during running. In order to reduce error, each segment of the multi segment foot model has it's own coordinate system to compare between trials. This model was validated using bi-planar fluoroscopy in comparison to the multi segment foot model used by Jenkyn et al. 2007, and determined no significant differences in fluoroscopy landmarks to multi segment foot model landmarks.

There is a higher rate of error in skin motion artifact during running than walking (Leardini et al., 2005). Errors seen can cause marker displacement of up to 20 mm relative to the underlying bone (Leardini et al., 2005). Mean displacements seen between anatomical bone landmarks and skin markers can vary from 2.7 to 14.9 mm during running with malleoli having the largest artifact as found by via xray (Maslen & Ackland, 1994). Error is reduced by keeping markers consistent between trials and never moving them from static to running trials. This way, if placed consistently differences between trials and conditions can be seen.

2.2 Multi-segment foot model

There are many segmented foot models that track various segments and movements of the foot. This particular model allows for tracking and study of clinically important movements, including the medial longitudinal arch and the windlass mechanism of the hallux. This model has to be compatible with the optical motion analysis system currently in place at the Wolf Orthopedic Biomechanics Laboratory (WOBL) in the Fowler-Kennedy Sports Medicine Center. A system with too many markers to define key bones would clutter the foot but too few markers mean risking missing some crucial movements of the foot. The markers used were chosen as they represent critical movements of the foot as well as it is clinically applicable for many different shoe sizes, even children.

Figure 2.1 Multi segment foot model developed by Jenkyn, et al (2007) used to track five segments of the foot. This model is broken down into the hallux, the medial forefoot, the lateral forefoot, midfoot and hindfoot.

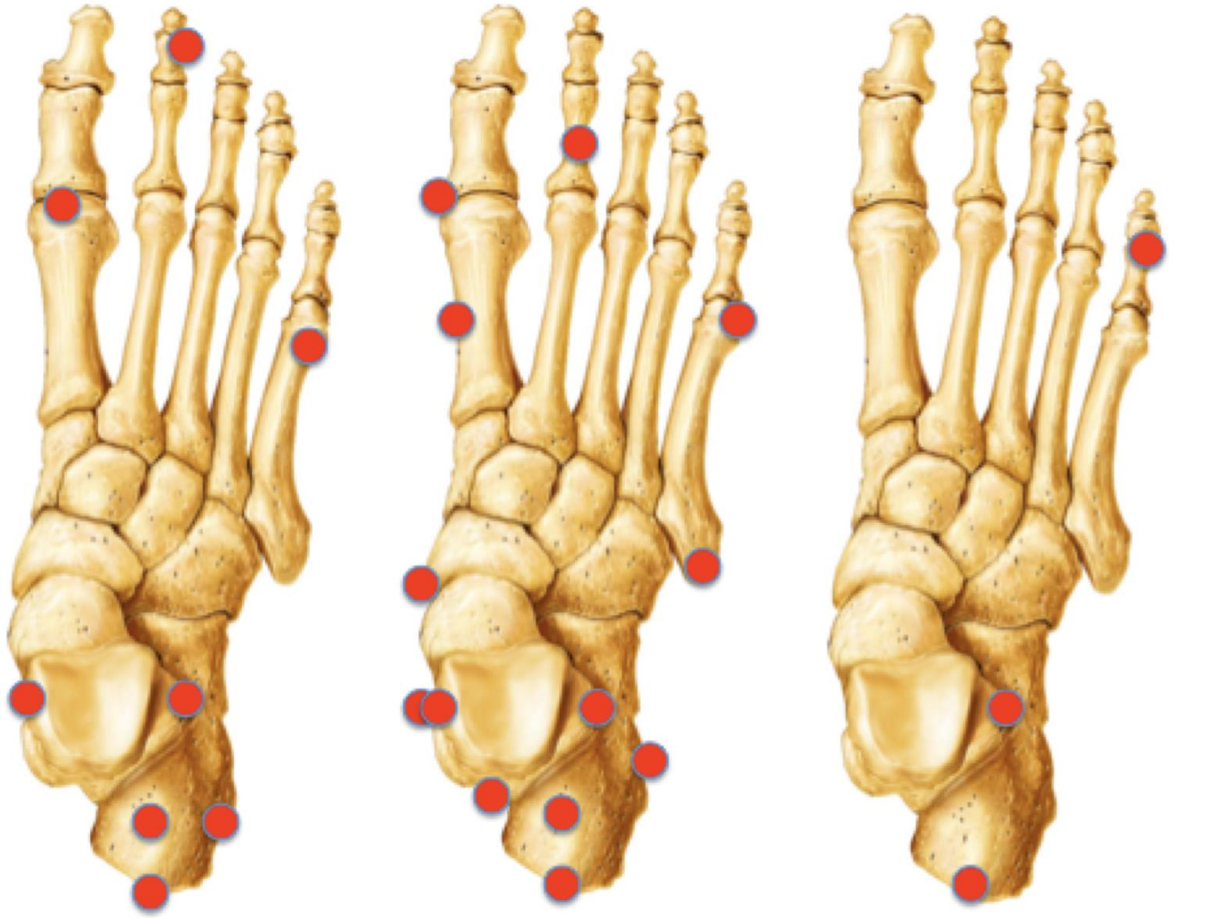


Figure 2.2 Illustration of the three multi segment foot models used for 3D gait analysis. Markers, illustrated with red circles, were singular and reflective. All studies did not break down markers into clusters or the foot into segments. On the left, a multi segment foot model as utilized by Milner, et al (2006). In the middle, a multi segment foot model used by Pohl, et al. (2005) and Morio, et al. (2009). On the right, a model used by Lieberman, et al. (2009).

The multi-segment foot model is one developed by Jenkyn & Nicol 2007, which tracks five segments of the foot separately using optical motion capture (Table 1). The segments are the hindfoot (calcaneous), talus, midfoot (cuneiforms I-III, navicular, cuboid), medial forefoot (metatarsal I and II), and lateral forefoot (Figure 2.1)(metatarsal III-V) (T R Jenkyn & Nicol, 2007; T R Jenkyn, Shultz, Giffin, & Birmingham, 2010; Thomas R Jenkyn & Anas, 2009). There are many different versions of the multi-segment food models that track different segments of the foot (Table 1).

Table 1 Cluster names and locations of the multi-segment foot model

Cluster	Cluster Location
Hindfoot	Posterio-lateral calcaneus, lateral to Achilles tendon
Midfoot	Dorso-medial foot over the navicular tuberosity
Medial forefoot	Medial-dorsal foot over midshaft of first metatarsal
Lateral forefoot	Lateral-dorsal foot over midshaft of fifth metatarsal
Hallux	Dorsal over the proximal phalangeal of the hallux

To track all 26 bones of the feet, bone pins would need to be used. Thus, the reasoning for segmenting the foot into five sections that can be assumed to be rigid. Defined segments needed to be large enough to attach a cluster marker and have enough distance in between as to be seen by motion capture cameras. Markers located in close proximity would result in missing, merged and lost clusters. Clusters of three markers are used to track the three-dimensional position and orientation of each segment (T R Jenkyn & Nicol, 2007; Thomas R Jenkyn & Anas, 2009).

Within this study, the current kinematic model represents four joint motions: medial longitudinal arch ratio, hallux-medial forefoot, forefoot-midfoot and hindfoot-midfoot. The markers are custom made, spherical 8 mm diameter, covered in reflective tape (3M, Minneapolis, MN) attached by 3 mm carbon fiber rods to a 20 mm polyethylene base (Figure 2.3) (T R Jenkyn & Nicol, 2007; T R Jenkyn et al., 2010; Thomas R Jenkyn & Anas, 2009). Typically markers used in other studies consist of singular reflective spheres attached to the body (Figure 2.2).

Figure 2.3 Custom made clusters of spherical markers used in the multi-segment foot model.

The kinematic model of the foot is reported with three motions: the medial longitudinal arch ratio, forefoot-midfoot and midfoot-hindfoot. The MLA was described as a height to length ratio. It was defined by the position of the segments of the hindfoot, midfoot and medial forefoot. Both forefoot segments represented motions of the midfoot as a compound twisting motion in the midfoot frontal and transverse planes. The hindfoot segment was compared to the midfoot segment and represented as orthogonal rotations in the midfoot frontal and transverse planes.

Bony landmarks are the identical landmarks used by Jenkyn et al. (2009) & Jenkyn & Nichol (2007) and can be seen in the following table as reproduced with permission (Table 2). The x, y, z coordinates of each landmark was located in a lab frame of reference. One vector was created using the points on the stylus and averaged over 1 second at 60Hz in order to define the orientation of the stylus. This was then used to calculate the coordinates of the metal end in the lab frame of reference, which represents the anatomical landmark.

Table 2 Bony landmarks digitized for each segment used to define segment fixed axes. Note that the medial and lateral forefoot segments share landmarks (Jenkyn, et al. 2009; Jenkyn & Nichol, 2007).

Segment	Tracked landmarks
Thigh	FLE: lateral epicondyle (most lateral point)
	FGT: greater trochanter (most lateral point)
	FME: medial epicondyle (most medial point)
Lower leg	LLM: lateral malleolus (most lateral point)
	LFH: fibular head (most lateral point)
	LMM: medial malleolus (most medial point)

Hindfoot	CAER: eminentia retrotrochlearis (greatest lateral elevation) CALT: lateral tuberosity (lateral to the achilles tendon attachment) CAMT: medial tuberosity (medial to achilles tendon attachment)
Midfoot	MCI: first cuneiform (distal dorsal crest) MNT: navicular tuberosity (most medial point) MCU: cuboid (lateral dorsal edge at joint with calcaneous)
Medial forefoot	MIH: first metatarsal head (most dorsal point) MIB: first metatarsal base (most dorsal point)
Lateral forefoot	MVH: fifth metatarsal head (most dorsal point) MVB: fifth metatarsal base (most dorsal point)

Landmarks are palpated during quiet standing (T R Jenkyn & Nicol, 2007; T R Jenkyn et al., 2010; Thomas R Jenkyn & Anas, 2009). Once landmarks were located a stylus instrumented with three auto-reflective markers was used to establish measure its three dimensional position with respect to the corresponding segment reference frame (Table 2) (Figure 2.4) (T R Jenkyn et al., 2010; Thomas R Jenkyn & Anas, 2009). This is the same method as used by Jenkyn et al. (2010). Defining the landmarks during quiet standing allows these static bone-fixed reference frames to be considered as the weight bearing neutral positions and thus, the measurements made can be compared to movement trials (T R Jenkyn et al., 2010).

The stylus is 26.4 cm in length, made up of a metal rod, 5 mm in diameter, with 1 cm spherical balls covered in reflective tape (3M, Minneapolis, MN). Distance between distal and middle markers was 6.4 cm, distance between the middle and proximal markers was 8.3 cm and distance between the proximal marker and the end of the stylus was 11.1 cm (Figure 2.4).

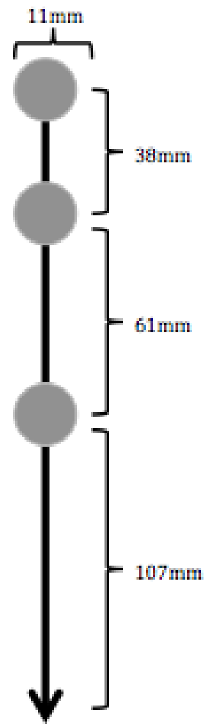


Figure 2.4 Custom made stylus, used in locating underlying bones and landmarks.

2.3 Windowing shoes

In order for the multi-segment foot model to work with any type of running shoes, the cluster markers must be visible to the optical motion tracking system, thus holes or windows must be created within the shoe (Shultz & Jenkyn, 2012). Windowing shoes decreases the amount of motion artifact that occurs when foot motion is measured using markers applied to shoes. These windows must be large enough to allow the cluster markers to fit, as well as the shoe should not interfere with any movement of the foot (Shultz & Jenkyn, 2012). More importantly, the windows cannot affect the structural integrity of the shoe itself (i.e. cut too large) (Shultz & Jenkyn, 2012).

It was found by Schultz & Jenkyn (2011), that the maximum size of a hole cannot exceed 1.7 cm x 2.5 cm (Shultz & Jenkyn, 2012). Larger holes will affect the integrity of the shoe (Shultz & Jenkyn, 2012). Any size smaller than 2.7 cm x 2.3 cm will not change the structural integrity or performance of the shoe (Shultz & Jenkyn, 2012).

Although there is very limited research, studies conducted on shoe windowing have been focused on the integrity and support of solely windowing on motion control shoes (Shultz & Jenkyn, 2012). This should not present itself as a great concern, as the VFF shoes do not contain stability or motion control measures that will affect the integrity of the shoe. Holes were cut on areas where there was no structural support, such as on the upper that is simply made of a synthetic stretchy material.

The aim of this study was to see if differences exist between VFF and barefoot running in habitually shod and VFF runners. Currently, there is a gap in literature within this topic, with little-kinematic data available.

Chapter 3: Methods

3.1 Experimental equipment

All kinematic data collection and testing was done at the Wolf Orthopaedic Biomechanics Laboratory (WOBL) at the Fowler-Kennedy Sports Medicine Clinic at The University of Western Ontario in London, Ontario, Canada. This study was approved by the Health Science Research and Ethics Board (HSREB #104830) at the University of Western Ontario.

Prior to each subject participating in this study, subjects gave written consent after reading information on the data collection process, goals and methods (Appendix A). Subjects then completed a Physical Activity Readiness Questionnaire (PAR-Q) regarding basic information on their health and exercise status (Appendix E). If subjects answered 'yes' to any of the physical activity questions, they were not eligible to participate in the study.

Three groups of participants were studied: habitual shod runners, habitual VFF runners and habitual barefoot runners. The habitual shod runners comprised four individuals (2 female, 29 ± 8.5 years and 2 male, 26 ± 4.2 years) who ran for an average of 23.8 ±11 km per week over the past 2 years. The habitual VFF runners comprised three (males, 24.3 ± 0.6 years), who ran for an average of 13.3 ±7.6 km per week over the previous 2 years. The habitual barefoot runner comprised one (male) age 22, who ran for an average of 100-120 km per week over the past 2 years.

Subjects were recruited from the Fowler Kennedy Sport Medicine Clinic and the local community. Participants did not have any musculoskeletal disorders or current diagnosed injuries. Individuals were excluded if they had a current musculoskeletal disorder such as osteoporosis or muscular dystrophy, were under 18 years of age, were not habitual runners (< 2 years and < 10 km/week), or had current lower body injuries such as broken bones, sprains or strains.

Subjects were assigned a random participant ID, using a random number generator between numbers 1-100. Names of participants were not used during analyses, they were referred to their ID numbers. Anthropometric data of sex, weekly running mileage, age, weight, height, knee widths, ankle widths, feet widths and feet lengths were collected and measured prior to testing.

Table 3.1 Demographic data of eight subjects of the study describing group, average running mileage per week, sex, age at time of data collection, weight and height.

Subject	Group	Km/week	Sex	Age	Weight (lb)	Height (in)
27	Shod	30	F	23	120	66.5

55	Shod	20	M	23	155	68
3	Shod	35	F	35	230	66
87	Shod	10	M	29	175	72
90	Vibram	10	M	24	180	70
50	Vibram	20	M	25	210	73
7	Vibram	10	M	24	190	69
31	Barefoot	100-120	M	22	140	69

3.2 Motion analysis equipment

WOBL is equipped with real-time 3D optical motion capture cameras from Motion Analysis Corporation (Santa Rosa, USA). Kinematic data was collected using 8 Hawk motion-capture cameras (Motion Analysis Corp, Santa Rosa, USA) and 4 Eagle motion-capture cameras (Motion Analysis Corp, Santa Rosa, USA) at a sample rate of 120 Hz, using infrared technology.

The frame reference origin was established as the northwest corner of the force plate. Axes were established during the calibration process, with the x axis (long axis of the lab (North) and y-axis laying perpendicular to the x axis (East). Both x and y-axes were parallel to the ground. The z-axis was pointed vertically and perpendicular to the x and y axes (Figure 3.1).

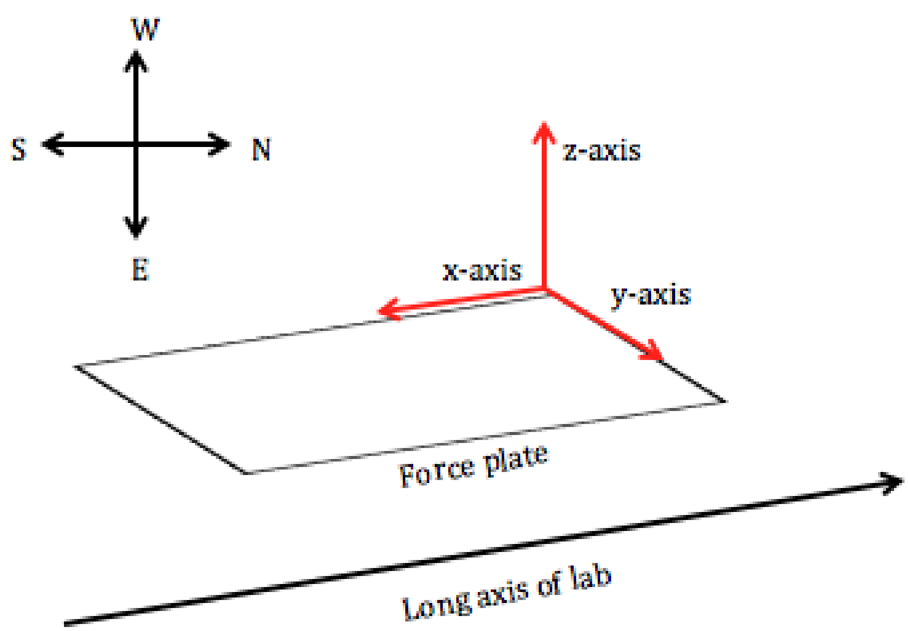


Figure 3.1 Lab coordinates are based off of the NW corner of the force plate (as defined by the corner with the red arrows). The x axis is pointing south, the y axis is pointing east and the z axis is perpendicular to both the x and y axes.

The system allows for the capture of movements in real-time. Templates are created in Cortex 2.6.8 (Motion Analysis Corp, Santa Rosa, USA) to allow quick and easy viewing of movements of the body.

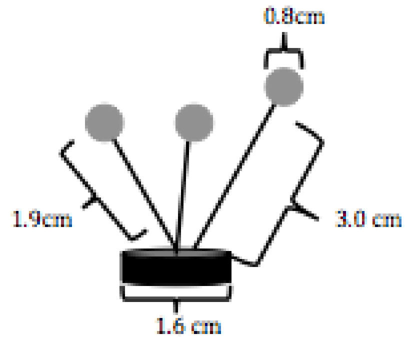


Figure 3.2 Triad cluster markers used in the multi-segment foot model. All trial clusters were custom made at the University of Western Ontario (London, ON, Canada) and markers were covered in reflective tape (3M, Minneapolis, MN, USA).

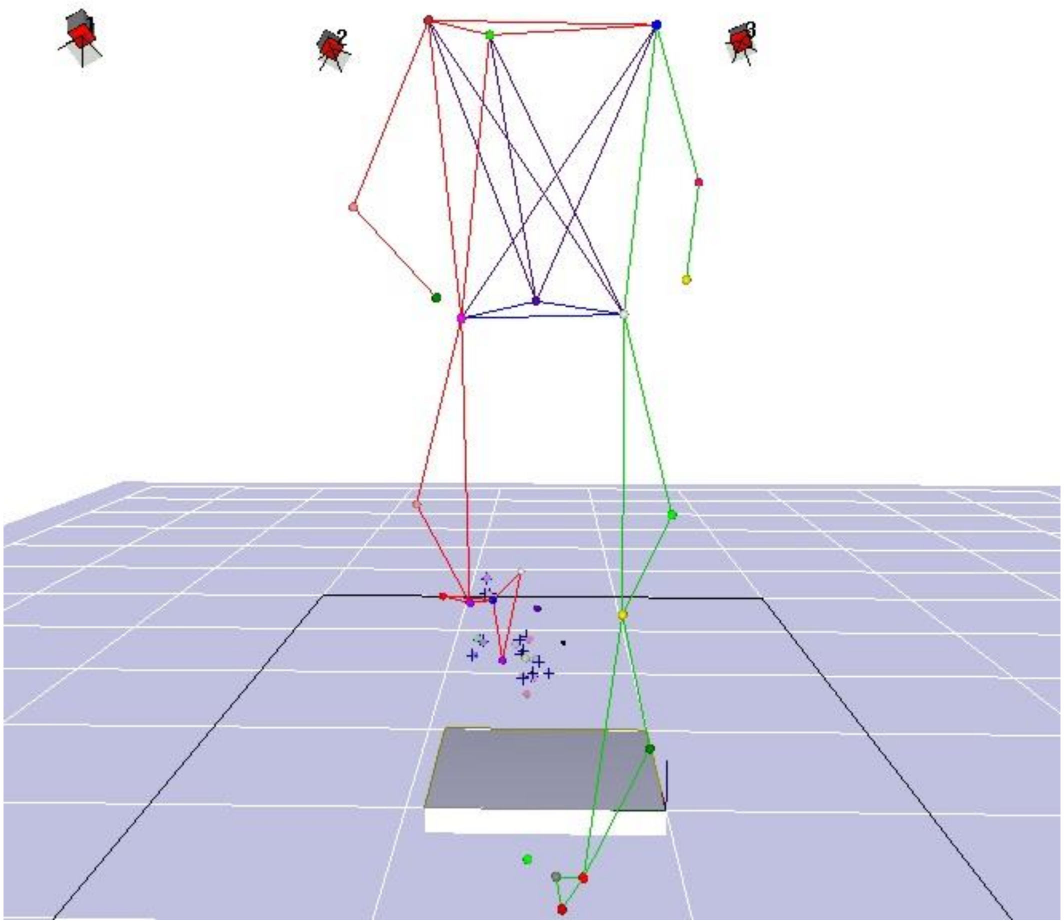


Figure 3.3 MSFM markers and Helen Hayes modified body marker sets identified in Cortex 2.6.8 during a running trial (Motion Analysis Corp, Santa Rosa, USA).

The 3D optical motion capture cameras used infrared light to record reflection from the markers. In order to be seen and recorded by the cameras, a marker must appear to a minimum of two cameras to decrease positional errors. Each camera in Cortex located the markers in 2D views and used an algorithm to triangulate 3D coordinates.

Feet were carefully marked at the commencement of data collection to ensure that markers were reattached to the correct locations between trials. Prior to dynamic movement trials, marker placement was carefully studied in 3D view to ensure optimal placement

3.3 Experimental procedures

3.3.1 Calibration

Prior to the commencement of testing each day, the cameras were calibrated using Cortex 2.6.8 (Motion Analysis, Santa Rosa, USA). The calibration allows for an accurate location of each marker by creating a relationship between the markers attached to the subject and each of the markers location within the lab frame of reference.

First, the calibration frame (i.e. the seed frame) was placed onto the NW corner of the force plate (Figure 3.3). This L shaped frame comprises of four fixed reflective markers and gives the optimal position for the four markers to be seen by the 12 cameras.

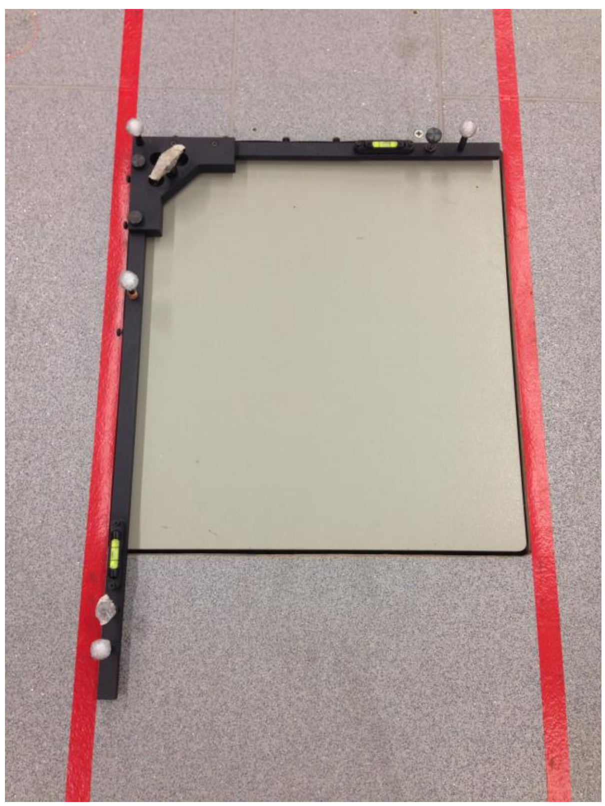


Figure 3.4 L-shaped calibration frame (seed frame) creating the axes of the lab. The four fixed markers help create a lab frame of reference.

A dynamic calibration wand with three markers was 'waved' through the lab for two minutes. This expands the calibrated seed area and allows for a more accurate marker location, according to the recommendations by the manufacturer.

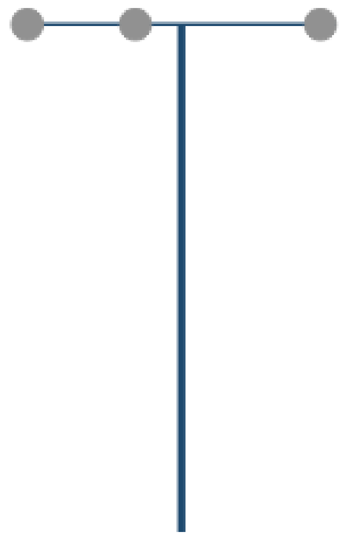


Figure 3.5 Calibration wand fitted with three fixed reflective markers used to calibrate the data collection area.

3.3.2 Helen Hayes marker set

Additional individual markers were arranged in a modified Helen Hayes marker configuration as seen in Table 3.2.

Table 3.2 Positions of reflective markers based off of the Helen Hayes marker set.

Description	Placement
Left shoulder	Tip of acromion process
Right shoulder	Tip of acromion process
Right scapula	Superior angle
Left elbow	Lateral epicondyle of the humerus
Right elbow	Lateral epicondyle of the humerus
Left wrist	Centered between the styloid processes of the radius and ulna
Right wrist	Centered between the styloid processes of the radius and ulna
Left superior iliac spine	Top of crest
Right superior iliac spine	Top of crest
Sacrum	Superior aspect at the L5-sacral interface
Left lateral knee	Along the flexion/extension axis of rotation at lateral femoral condyle
Right lateral knee	Along the flexion/extension axis of rotation at lateral femoral condyle
Left lateral ankle	Along the flexion/extension axis of rotation at lateral malleolus
Right lateral knee	Along the flexion/extension axis of rotation at lateral malleolus
Left top of foot	Centre of the foot between the 2nd and 3rd metatarsals
Right top of foot	Centre of the foot between the 2nd and 3rd metatarsals
Left heel of shoe	Posterior calcaneus at same height from floor as toe marker
Right heel of shoe	Posterior calcaneus at same height from floor as toe marker
Left thigh wand	Mid lateral thigh
Right thigh wand	Mid lateral thigh
Left shank wand	Mid lateral shank
Right shank wand	Mid lateral shank

3.3.3 Cluster marker setup

The feet were tracked with an additional 'Multi-segment Foot Model' marker set. Clusters of three markers were placed on the hallux, first metatarsal, fifth metatarsal, the heel and navicular (Table 3.3).

Table 3.3 Multi-segment foot model marker set and their respective locations. The same marker locations were used for each subject and for each condition.

Cluster	Cluster Location
Hindfoot	Posterio-lateral calcaneus, lateral to Achilles tendon
Midfoot	Dorso-medial foot over the navicular tuberosity
Medial forefoot	Medial-dorsal foot over midshaft of first metatarsal
Lateral forefoot	Lateral-dorsal foot over midshaft of fifth metatarsal
Hallux	Dorsal over the proximal phalangeal of the hallux

Figure 3.6 Cluster marker configuration of the multi-segment foot model. 5 triads consisting of 15 reflective markers represent 5 segments of the foot; hallux, midfoot, medial forefoot, lateral forefoot and calcaneus.

3.4 Static trial

Prior to dynamic trials, a static trial of 10 seconds was collected. This static trial is used to create a template for each subject. The static trial was used to determine the position underlying bones of the foot and allowed for anatomical frames of reference to be created.

After collection, the static trial was studied to ensure all 22 of the modified Helen Hayes markers and the 15 multi-segment foot model markers were visible. This was essential, as virtual markers need to be created in post-processing. For each subject, virtual markers were created using the location of an origin marker, a long axis marker and a plane marker. 'Snap-to' markers used the virtual marker to be created to ensure a proper triangulation between the three reference markers to find a location accurately.

Origin markers were always chosen as another marker attached the same cluster. Long axis markers were chosen based on markers most consistently viewed over a trial. Plane markers were chosen based on an adjacent marker or a marker of the same height of another triad (i.e. long marker on a triad with another long marker on a triad). No data could be missing in any of these definitions or the virtual marker could not be created.

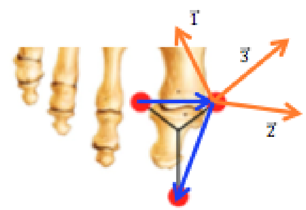
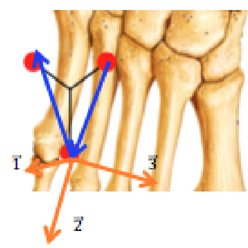
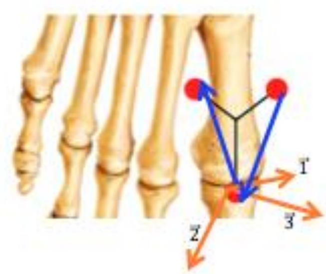
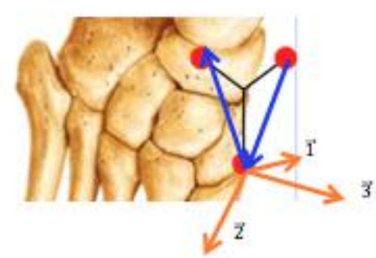
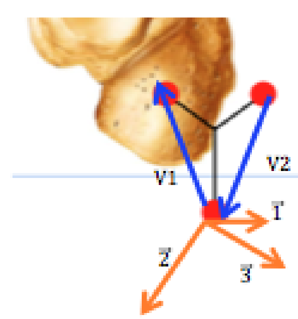
In some instances where markers were missing for a significant number of frames (>10 frames), trials were discarded. Clusters were named in the same sequence for each of the static and dynamic trials.

Having all markers present during the static trial ensured virtual marker placement to be accurate. Each subject had identical virtual marker definitions to ensure consistency throughout dynamic trials (Table 3.4).

Table 3.4 Virtual marker definitions of 5 MSFM triad clusters; Hallux, 1st metatarsal (midfoot), navicular (medial forefoot), 5th metatarsal (lateral forefoot) and calcaneus. Origin markers, long axis markers and plane markers were used consistently through each subject using static trials to define virtual marker locations.

Marker name	Origin marker	Long axis marker	Plane marker	'Snap-to' marker
Hallux Long	Hallux Medial	Navicular Lateral	1st Lateral	Hallux Long
Hallux Lateral	Hallux Medial	1st Lateral	Hallux Long	Hallux Lateral
Hallux Medial	Hallux Long	Navicular Lateral	1st Medial	Hallux Medial
1st Long (Midfoot)	1st Lateral	Navicular Lateral	Hallux Long	1st Long
1st Lateral (Midfoot)	1st Medial	Hallux Medial	Hallux Long	1st Lateral
1st Medial (Midfoot)	1st Lateral	Hallux Long	Navicular Lateral	1st Medial
Navicular Long (Medial Forefoot)	Navicular Lateral	Hallux Long	1st Long	Navicular Long
Navicular Lateral (Medial Forefoot)	Navicular Medial	Hallux Long	Navicular Long	Navicular Lateral
Navicular Medial (Medal Forefoot)	Navicular Lateral	Hallux Long	1st Medial	Navicular Medial
Calcaneus Long	Calcaneus Lateral	5th Long	5th Medial	Calcaneus Long
Calcaneus Lateral	Calcaneus Medial	5th Long	5th Lateral	Calcaneus Lateral
Calcaneus Medial	Calcaneus Long	5th Long	5th Medial	Calcaneus Medial
5th Long (Lateral Forefoot)	5th Medial	Calcaneus Long	Navicular Long	5th Long
5th Lateral (Lateral Forefoot)	5th Long	Calcaneus Long	Calcaneus Medial	5th Lateral
5th Medial (Lateral Forefoot)	5th Long	Calcaneus Medial	Calcaneus Lateral	5th Medial

Table 3.5 Cluster markers used in creating two vectors for each foot segment. Each set of vectors allows for a separate coordinate system to be defined.

Segment	Vectors	Example
Hallux	V1=H1-H3 V2=H3-H2 O=H3	
5th Metatarsal (Lateral Forefoot)	V1=LFF2-LFF1 V2=LFF1-LFF3 O=LFF1	
1st Metatarsal (Medial Forefoot)	V1=MFF2-MFF1 V2=MFF1-MFF3 O=MFF1	
Navicular (Midfoot)	V1=MF2-MF1 V2=MF1-MF3 O=MF1	
Calcaneus (Hindfoot)	V1=HF2-HF1 V2=HF1-HF3 O=HF1	

Trials of one second were taken at 120 Hz with the stylus pointing at each of the landmarks (Table 3.6). These landmarks were used to calculate the transformation matrix that showed the cluster motion in the segment frame of reference.

Table 3.6 Stylus landmarks used to calculate bone locations.

Segment	Description	Placement
Hindfoot	CAER	Eminentia retrotrochlearis (greatest lateral elevation)
	CALT	Lateral tuberosity (lateral to achilles tendon attachment)
	CAMT	Medial tuberosity (medial to achilles tendon attachment)
Midfoot	MCI	First cuneiform (distal dorsal crest)
	MNT	Navicular tuberosity (most medial point)
	MCU	Cuboid (lateral dorsal edge at joint with calcaneus)
Medial forefoot	MIH	First metatarsal head (most dorsal point)
	MIB	First metatarsal base (most dorsal point)
Lateral forefoot	MVH	Fifth metatarsal head (most dorsal point)
	MVB	Fifth metatarsal base (most dorsal point)
Ankle JCS	LMM	Medial malleolus (most medial point)
	LLM	Lateral malleolus (most lateral point)
Hallux	DH	Most distal point of the hallux
	LPH	Lateral head of the hallux
	MPH	Medial head of the hallux

3.5 Vibram Five Finger Shoes

Vibram Five Fingers Bikila shoes (Vibram Inc., Albizzate, Italy) were used for all subjects. The shoes were chosen based on minimalism of straps, laces and constraints over the top of the foot, allowing the foot to move as naturally as possible. This model was chosen because of accessibility to purchase to the general public, as they are available at many local sporting goods locations.

This model is a marketed as a running-specific model and is also promoted as providing a barefoot running experience (i.e. enhancing natural gait and forefoot striking). The shoe is supposed to fit snug to avoid blisters and the sole is designed to allow for optimal flexibility for simulating barefoot running. The sole is a 4 mm TC1 performance rubber, which is thickest under the ball of the foot and has a 0 mm heel-toe offset.

According to the manufacturer, Vibram Five Finger shoes do not require a break in period and the top materials are designed to stretch and fit most foot shapes. Sizing of the shoes are said to be true to size and were worn without socks.

Figure 3.8 Vibram Five Fingers Bikila without holes for the multi segment foot model. Reflective patterns on the shoes were covered with athletic tape as to avoid interference with infrared lighting and marker recognition.

3.6 Dynamic trials

Once all landmarks were captured and the template was formed, dynamic trials of both running and walking were conducted. Four each of walking, running barefoot and running in VFF were captured.

During the dynamic trials, one of the researchers was responsible for keeping track of dynamic trial type and ensuring all markers remained on the body between trials. The other researcher was responsible for operating the computer software, ensuring trials were being captured correctly and ensuring all markers were visible for each trial.

For both the walking and running trials (barefoot and VFF), subjects were walking and running at a self-selected pace. The 10m path in the WOBL lab was similar to a study done by Hashish et al., (2015), who had subjects run at a self-selected speed across a 9 m path.

Time allowed for a warm up before the running trials. Participants were instructed to run at a pace comfortable for them. As shown by Bates, (1978), there are no differences in rearfoot motion for running speeds between 3.3 and 4.5 m/s.

For conditions wearing shoes, double sided adhesive tape was applied to the foot. The front of the triad cluster markers were identified by drawing their locations on the foot. These identified locations allowed the markers to be removed between shoe conditions as shoes cannot be changed without removing the markers.

No randomization of the order of conditions was performed as this would necessitate removing and replacing cluster markers on the feet between trial conditions, which would introduce a source of error. The trails were not fatiguing, so the order of conditions should not have an effect. A study by Hreljac et al., (2000), had participants run at a testing speed of 4m/s, which, was greater than the training speed of most participants in their study. The study analyzed biomechanical factors that contribute to injury using a four marker system to identify the leg and rearfoot. Their subjects were tested at a variety of running speeds, and upon analysis of these preliminary tests, data was similar in all speeds tested (Hreljac et al., 2000). This is why speed was not measured in this study.

3.7 Kinematic Analysis

Kinematic data was analyzed for each frame for each trial, and trials of barefoot running and VFF were compared. Every stance phase was normalized over 100 data points. The 100 data points represents the stance phase from initial ground contact to toe off. Changing each gait cycle to 100 points allowed for a direct comparison to be made between each trial since data was collected at 120Hz and subjects were running at a self-selected pace, therefore having a varied number of frames both between trials and between subjects.

The five kinematic measures output by the multi-segment foot model were: the medial longitudinal arch height to length ratio (MLA), forefoot angle (FFA-F) in the frontal plane (pronation/supination of the forefoot with respect to the hindfoot), forefoot angle (FFA-T) in the transverse plane (narrowing and widening of the MLA), hindfoot angle (HF-F) in the frontal plane (pronation/supination with respect to the lower leg) and hindfoot angle (HF-T) in the transverse plane (adduction/abduction with respect to the lower leg).

3.7.1 Medial Longitudinal Arch (MLA)

Kinematics of the MLA is defined as the height to length ratio. For each frame the length of the arch was defined as the distance between the medial tuberosity of the calcaneus (CAMT) to the head of the first metatarsal (MIH). The height of the MLA is determined by the vertical vector stopping at the navicular tuberosity (NT), located perpendicularly to the vector between the CAMT and MIH.

The height to length ratio was compared between trials of barefoot running and trials of VFF running between subjects. Positive values indicate narrowing of the MLA, while negative values represent widening.

Figure 3.9 Length of the MLA is defined as the medial tuberosity of the calcaneus (1) to the head of the first metatarsal (2), and the height is determined by a perpendicular vector to the navicular tuberosity (3).

3.7.2 Forefoot pronation/supination

The forefoot angle is important in quantifying the angle of pronation and supination between trials of barefoot running and VFF running. The angle is quantified by the motion of the vector between the metatarsal heads of the 1st (MIH) and 5th (MVH) (1) and a second vector between the metatarsal bases of the 1st (MIB) and the 5th (MVB) (2).

Angles were developed as the twisting between the metatarsal heads and bases as compared to the midfoot. The angle was therefore reported as the angle of movement of the midfoot segment. In the frontal plane, positive angle values suggested pronation of the forefoot, while negative angle values suggested supination of the forefoot.

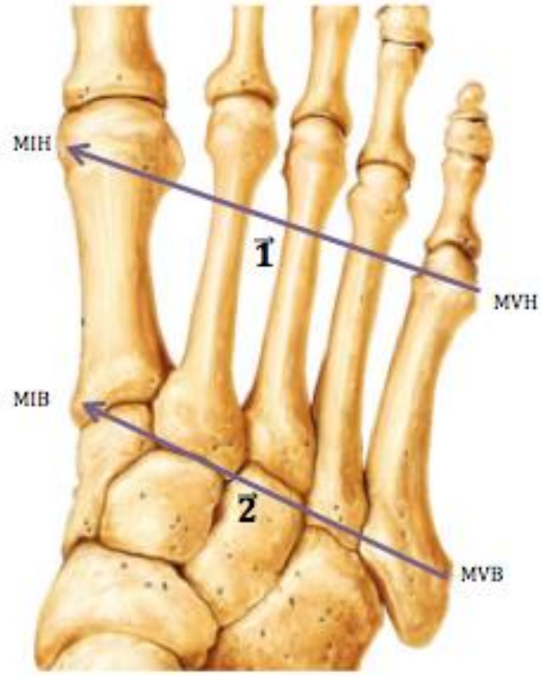
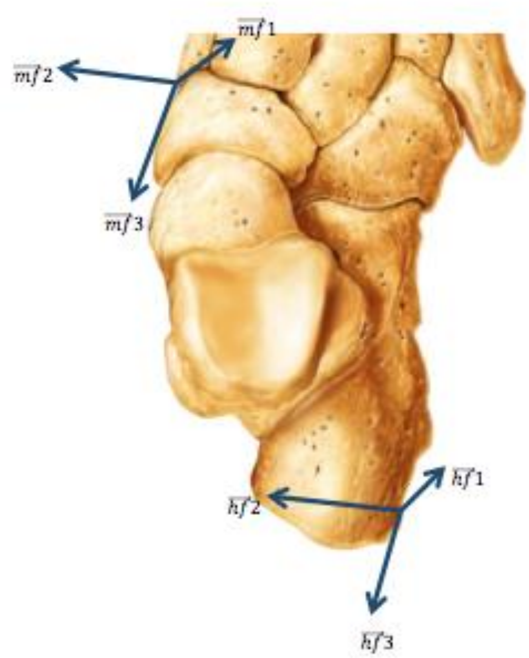


Figure 3.10 Two forefoot angle vectors as defined by the distance between the head of the 5th metatarsal (MVH) to the head of the 1st metatarsal (MIH) and the second vector defined as the distance between the base of the 5th metatarsal (MVB) to the base of the 1st metatarsal (MIB).

3.7.3 Hindfoot pronation/supination

Movement of the hindfoot (calcaneus) was defined compared to the midfoot. Analyzing this movement used floating axis angles (Grood & Suntay, 1983) in Matlab (Mathworks, Natick, USA). These angles allow for 3D angles to be studied in three planes of motion using rotation and translation matrices between two rigid bodies (Grood & Suntay, 1983).

Comparing angles of the hindfoot relative to the midfoot used the orientation of the hindfoot (distal) with respect to the midfoot (proximal) by creating a floating axis perpendicular to the primary axis of both the distal and proximal segments. Positive angles reflect supination of the hindfoot segment and negative angles reflect pronation of the hindfoot segment.



3.11 Calcaneus (hindfoot) and navicular (midfoot) markers and their vectors used to determine supination and pronation of the hindfoot segment.

3.7.4 Kinematic Variables

Kinematic variables used in this study are those of the MLA height to length ratio, forefoot movement and hindfoot movement. Positive values of the MLA height to length ratio (i.e. higher/shorter arch) suggests supination and a negative value of the MLA height to length ratio (i.e. lower/long) arch suggests pronation of the hindfoot. The overall size of the foot does not matter due to the ratio used.

With respect to the forefoot, motion here consists of twisting between the heads with respect to the bases at the midfoot. With respect to the frontal plane, a positive angle reflects pronation and a negative angle reflects supination. The angle in the sagittal plane was not quantified. In the transverse plane, a positive angle reflects spreading of the MLA as well as flattening of the 1st metatarsal and a negative angle reflects a rise in the MLA.

The hindfoot segment is described relative to the midfoot. In the transverse plane, a positive angle reflects adduction of the hindfoot and a negative angle reflects abduction of the hindfoot. With respect to the frontal plane, any movement here is considered rotation. A positive angle suggests supination and a negative angle suggest pronation.

3.8 Post Processing

A static trial captured the reference landmarks of the lower body with external surface markers. This was later used to link all virtual landmarks and build the coordinate frame of reference. To ensure that a full gait cycle was visible, the initial foot strike was located. The gait cycle for each subject was started ten frames prior to the initial foot strike and ended 5 frames after the subsequent toe off.

In the case of frames missing markers (i.e. when the marker could not be seen by a minimum of two cameras), virtual markers created in the static trials were used to identify those missing. Virtual joins were used when markers were missing from three consecutive frames. This most often happened when markers were too close together and the cameras simply picked up overlapping markers as one or when the swing leg passed by the stance leg, blocking markers from the field of view.

Virtual joins are an effective way of determining another markers position in the field of reference by taking an origin and the long and plane axis. The origin is the marker most rigidly attached to the body, the long and plane axes were chosen as markers known nearby. Without these three known markers, virtual joins are not possible. Once all the markers were identified for each trial, .trc files containing the x, y, z coordinates for each marker were exported. Files were compiled for consistency to be used in Matlab (Mathworks, Natick, USA).

Cluster markers were named in the same order for every trial. Attention was paid to naming these markers, as the importance for correctly identifying each location determines the underlying bone's position in the frame of reference (Table 3.5).

Matlab (Mathworks, Natick, USA) was used to process the marker data. Once markers were digitized, data was put into Matlab (Mathworks, Natick, USA) in order to be processed through a multi segment foot model developed by Shultz, et al. (2006)**.** Each trial was low pass filtered with a fourth-order Butterworth filter with a cutoff of 6 Hz as used in previous research (Jenkyn T. A., 2009). This removed the vibration movement artifacts from the triad marker clusters and from the marker positions (Jenkyn et al., 2010).

Chapter 4: Results

A total of nine subjects successfully completed the testing sessions with the testing session lasting an average of 35 minutes.

The five segment foot model was successfully tracked for eight of the nine subjects. The data of one subject (subject 029) was removed from the analysis since the static wand landmark data of the CAER landmark was not properly collected. The missing CAER landmark meant that the hindfoot could not be tracked and so this subject was removed. Of these eight test subjects, 4 runners trained habitually with running shoes, 3 trained with VFF and 1 trained barefooted. All test subjects were tested wearing VFF shoes and in barefoot.

To test the hypotheses, two subjects groups were created. Data from the habitual shod runners (n=4) formed the habitual shod group. Data from the habitual VFF runners (n=3) plus the habitual barefoot runner (n=1) were combined and formed the habitual VFF group (n=4). The five kinematic measures from the multi-segment foot model were reported comparing the trials wearing VFF and barefoot. Differences were compared for significance using independent samples t-tests. None of the findings were significantly different.

4.1 Habitual shod and habitual VFF running barefoot

4.1.1. MLA (barefoot)

The motion of the medial longitudinal arch for the habitual shod runners was not significantly different from the habitual VFF runners when running barefoot, as shown by the averaged MLA height to length ratios ($p > 0.05$; Figure 4.1). From heel strike to midstance (0-50% of stance), the habitual shod runners had negative ratios meaning the MLA was lower and they had a wider arch. The habitual VFF and barefoot runners had higher ratios and therefore their MLA was higher and they had a narrowed arch.

The shapes of the two averaged curves tended to mirror each other, with the shod runners not peaking during the stance phase and the VFF runners peaking soon after heel strike. The largest positive ratio for VFF runners occurred at midstance (0.024) and the largest negative ratio for shod runners occurred at heel strike (0% of stance; -0.12) for their respective average curves.

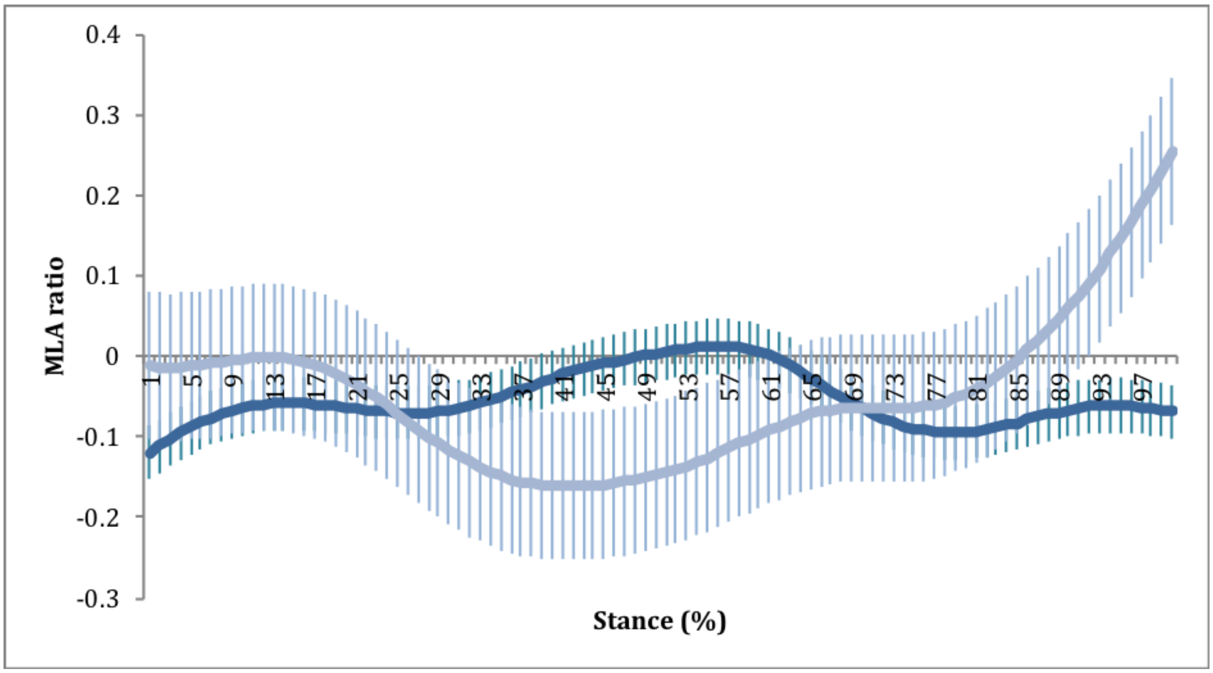


Figure 4.1 MLA data as represented by the height to length ratio during the stance phase of running barefoot. Positive ratios reflect a narrowing of the arch and negative ratios reflect a widening of the arch. The light blue and navy traces illustrate the VFF and shod conditions respectively. MLA is defined as a dimensionless ratio of arch height to length.

4.1.2 Forefoot in the frontal plane (barefoot)

The motion of the forefoot for the habitual shod runners showed no significant differences from the habitual VFF runners when running barefoot, as shown by the averaged pronation and supination ($p > 0.05$; Figure 4.2). Throughout the stance phase both the habitual shod runners and the habitual VFF runners had negative angles, thus the forefoot was supinated.

The shapes of the two averaged curves do follow the same pattern, however, the habitual shod runners are more consistently supinated throughout the stance phase and the habitual VFF runners are the most supinated at foot strike (0% of stance; -10.19°).

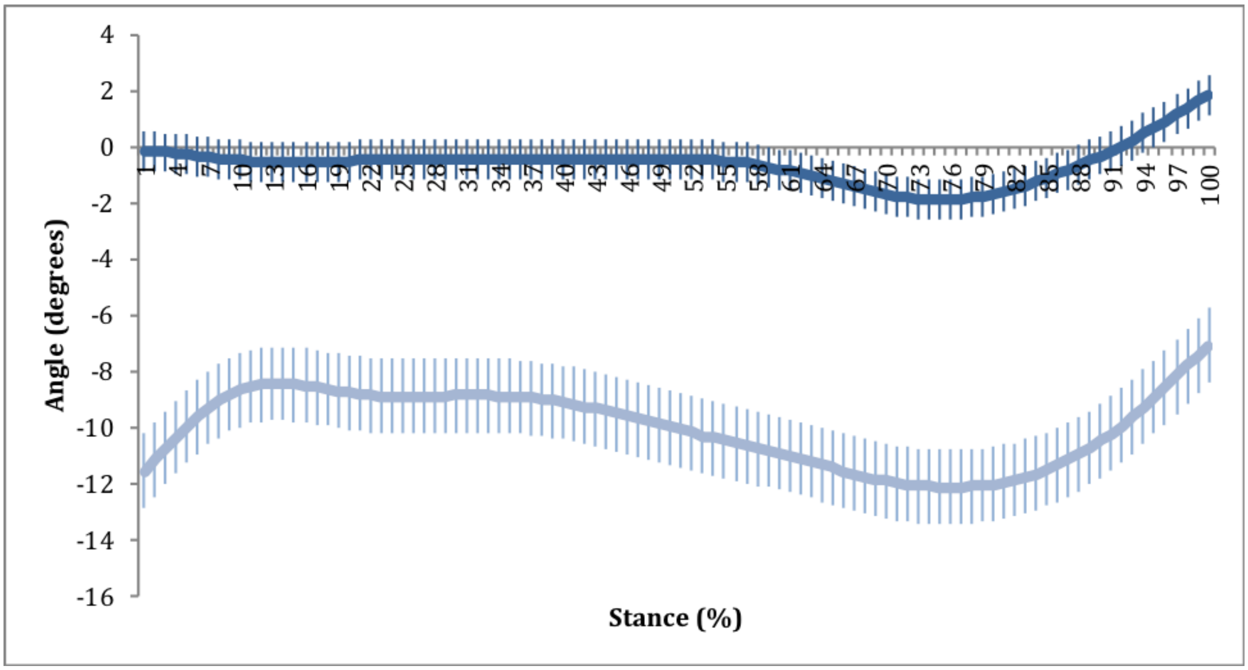


Figure 4.2 Frontal plane forefoot angle with respect to the midfoot during the stance phase of running in the barefoot condition. Positive angles reflect pronation and negative angles reflect supination. The navy and light blue represent the shod and VFF conditions respectively.

4.1.3 Forefoot in the transverse plane (barefoot)

The motion of the forefoot for the habitual shod runners was not significantly different from the habitual VFF runners when running barefoot, as shown by the rise and spread of the MLA ($p > 0.05$; Figure 4.3). From foot strike to toe off (0-100% of stance), the habitual barefoot runners and the habitual VFF runners had positive transverse plane forefoot angles with respect to the midfoot. However, the shod runners have a greater spread of the MLA (smaller positive transverse plane forefoot angles with respect to the midfoot).

The shapes of the two averaged curves do not follow the same pattern towards toe off. At foot strike (0% stance), both VFF and shod runners had their peak in MLA with shod runners totalling 18.2° and VFF runners totalling 26.5°.

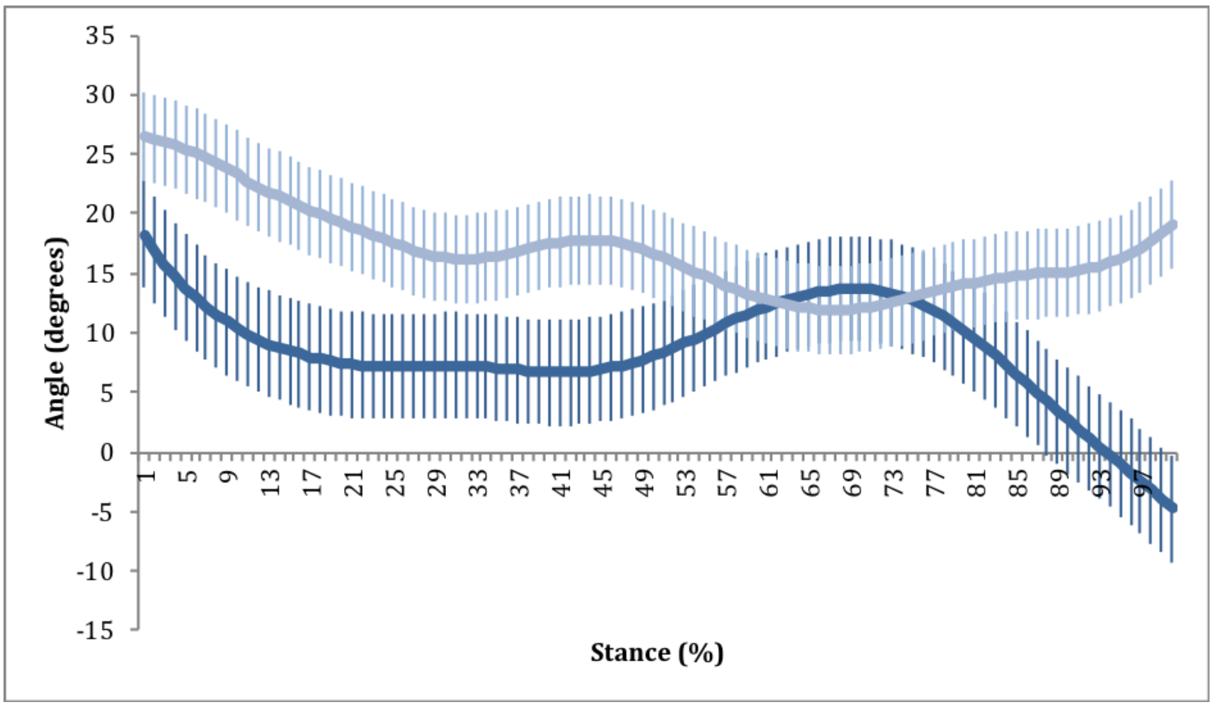


Figure 4.3 Transverse plane forefoot angle with respect to the midfoot during the stance phase of running in the barefoot condition. Positive angles reflect a narrowing in the MLA and negative angles are seen as the widening of the MLA. The navy and light blue represent the shod and VFF conditions respectively.

4.1.4 Hindfoot in the frontal plane (barefoot)

The motion of the hindfoot for the habitual shod runners was not significantly different from the habitual VFF runners when running barefoot, as shown by pronation and supination ($p > 0.05$; Figure 4.4).

At foot strike, the VFF runners supinated at the calcaneus and habitual shod runners pronated at the calcaneus. At the end of toe off (100% of stance), both the habitual shod runners and the habitual VFF runners both decreased their pronation and supination respectively. The habitual shod peaked at 3.2° of supination and habitual VFF peaked at 11.3° of supination.

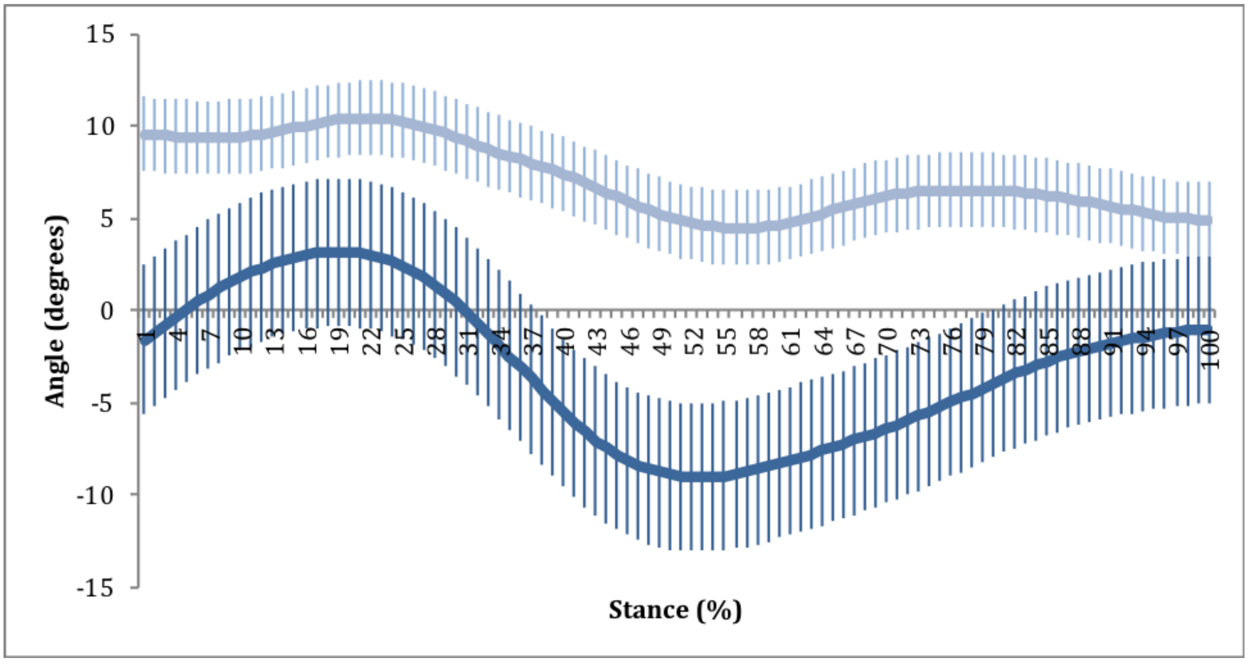


Figure 4.4 Frontal plane hindfoot angle with respect to the midfoot during the stance phase of running in the barefoot condition. Positive angles reflect supination and negative angles reflect pronation. The navy and light blue represent the shod and VFF conditions respectively.

4.1.5 Hindfoot in the transverse plane (barefoot)

The motion of the hindfoot for the habitual shod runners was not significantly different from the habitual VFF runners when running barefoot, as shown by the adduction and abduction ($p > 0.05$; Figure 4.5). The habitual shod runners peaked at -6.4° abduction and habitual VFF runners peaking at 4.1° adduction.

The shapes of the two averaged curves do follow a similar pattern, however the VFF runners were mostly adducted throughout the stance phase while the shod runners were abducted.

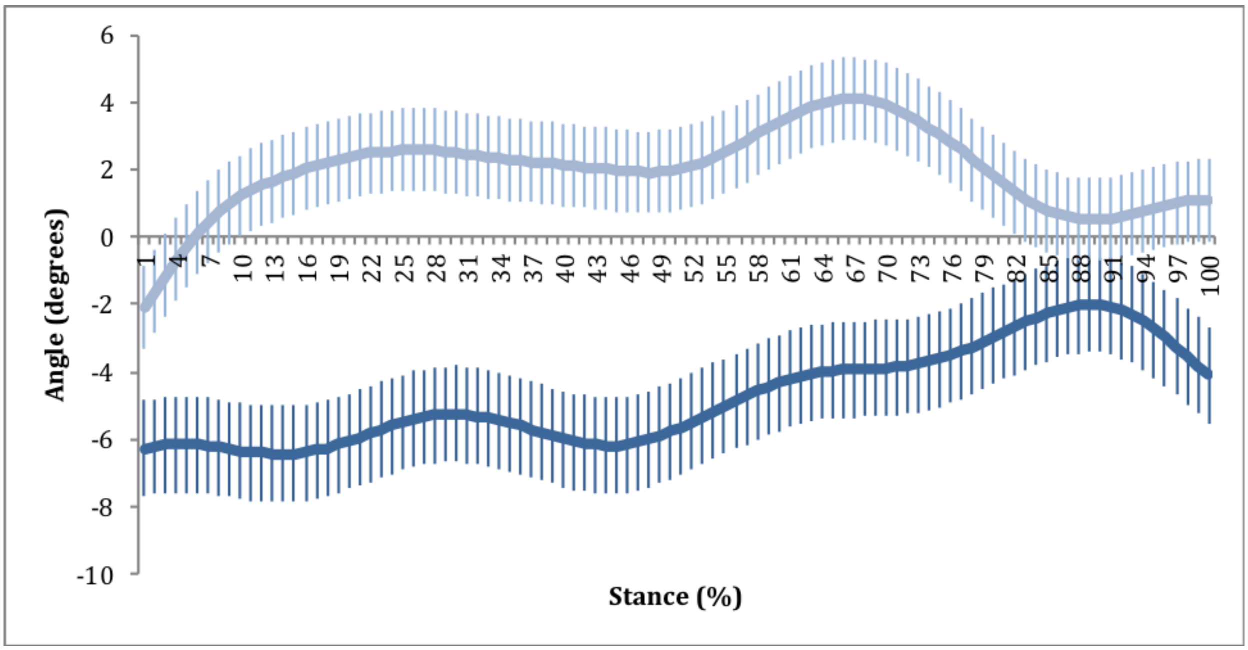


Figure 4.5 Transverse plane hindfoot angle with respect to the midfoot during the stance phase of running in the barefoot condition. Positive angles reflect adduction and negative angles reflect abduction. The navy and light blue represent the shod and VFF conditions respectively.

4.2 Habitual shod and habitual VFF running in VFF

4.2.1 MLA (VFF)

The motion of the medial longitudinal arch for the habitual shod runners were not significantly different from the habitual VFF runners when running in VFF, as shown by the averaged MLA height to length ratios ($p > 0.05$; Figure 4.6). From foot strike to midstance (0-50% of stance), the habitual shod runners had negative ratios and arches were widened. Habitual VFF runners had higher positive ratios, thus arches were narrowed.

The shapes of the two averaged curves are opposing, as the VFF runners do not peak until the end of toe off, whereas the shod runners peak soon towards midstance. The largest positive ratio for VFF runners occurred at toe off (100% of stance; 0.19) and the largest negative ratio for shod runners occurred soon after foot strike (-0.24) for their respective average curves.

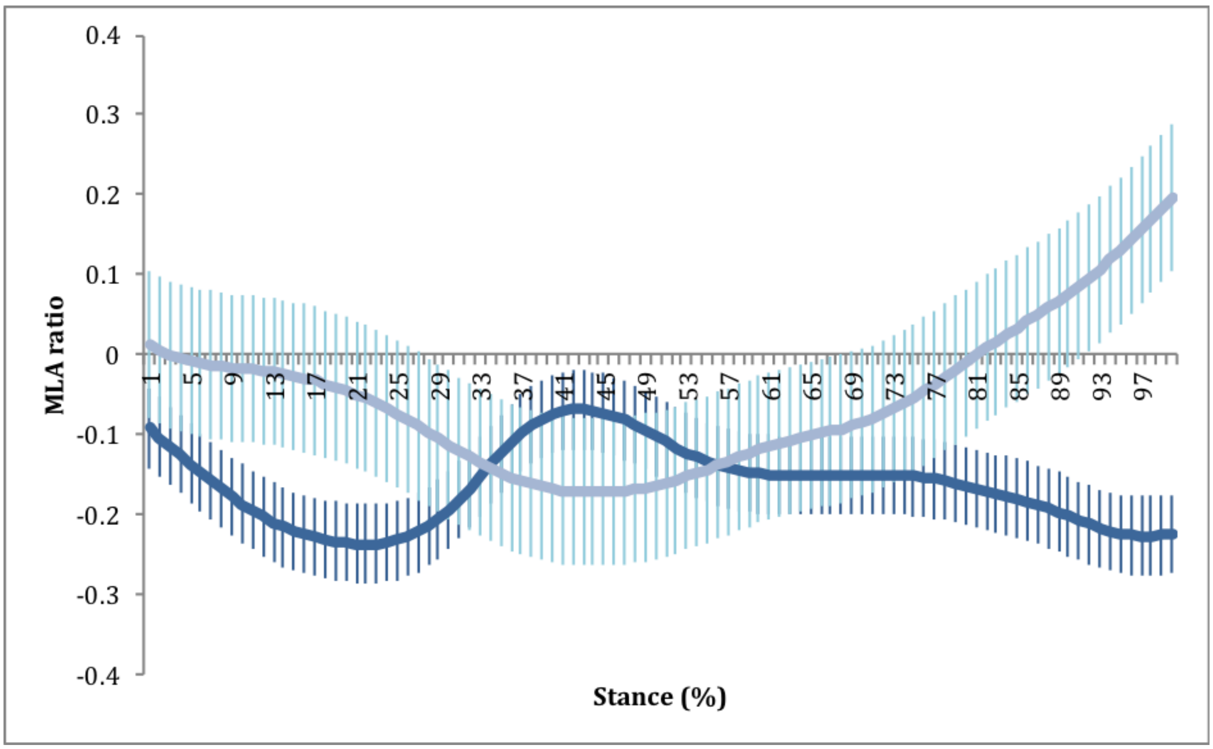


Figure 4.6 MLA data as represented by the height to length ratio during the stance phase of running in VFF. Positive ratios reflect a narrowing of the arch and negative ratios reflect a widening of the arch. The light blue and navy traces illustrate the VFF and shod conditions respectively. MLA is defined as a dimensionless ratio of arch height to length.

4.2.2 Forefoot in the frontal plane (VFF)

The motion of the forefoot for the habitual shod runners was not significantly different from the habitual VFF runners when running in VFF, as shown by the averaged pronation and supination ($p > 0.05$; Figure 4.7). From foot strike to toe off (0-100% of stance), the habitual shod runners maintained a pronated stance or a positive angle and the habitual VFF runners maintained a supinated stance or a negative angle.

The shapes of the two averaged curves do follow the same pattern, with the habitual shod runners peaking their pronation at 0% stance or foot strike (7.6°) and habitual VFF runners peaking their supination at foot strike (0% of stance; -13°).

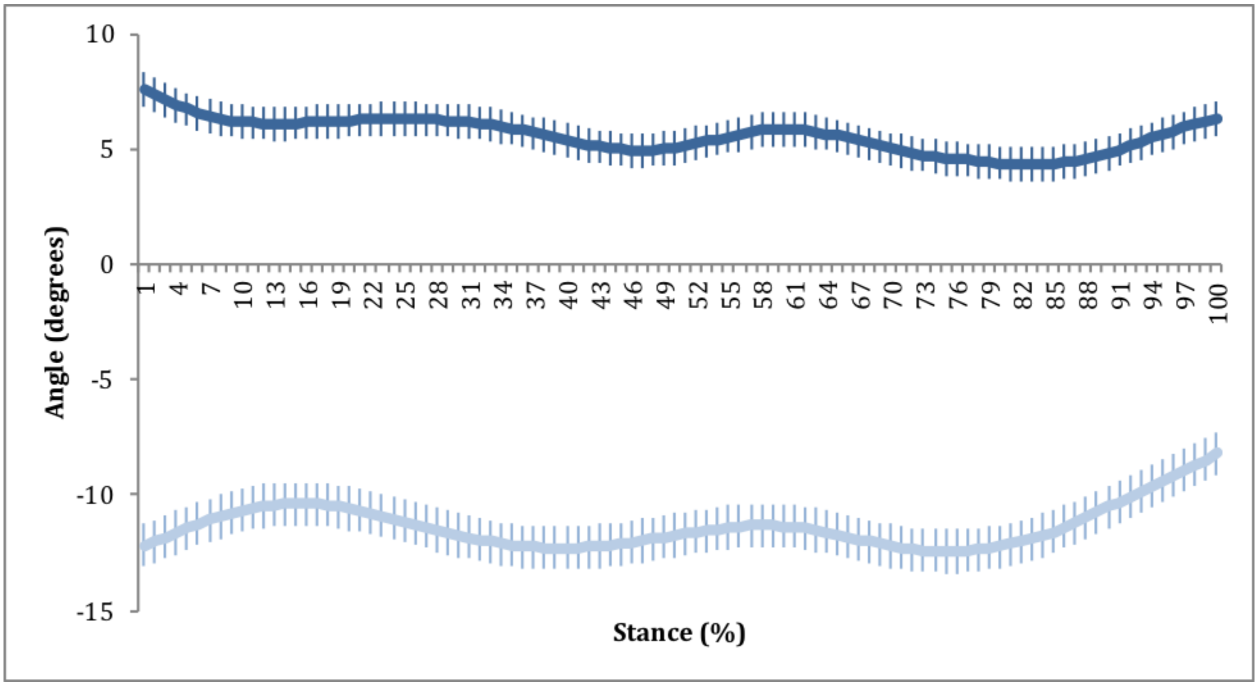


Figure 4.7 Frontal plane forefoot angle with respect to the midfoot during the stance phase of running in the VFF condition. Positive angles reflect pronation and negative angles reflect supination. The navy and light blue represent the shod and VFF conditions respectively.

4.2.3 Forefoot in the transverse plane (VFF)

The motion of the forefoot for the habitual shod runners was not significantly different from the habitual VFF runners when running in VFF, as shown by the rise and spread of the MLA ($p > 0.05$; Figure 4.8). From foot strike to toe off (0-100% of stance), the habitual shod runners had negative outcomes (widening of the MLA) and the habitual VFF runners had positive outcomes (narrowing of the MLA).

The shapes of the two averaged curves do not follow the same pattern. The VFF runners have a rise in the MLA and then a slight decrease and then the MLA peaks at toe off (100% of stance) at 14.7°. The habitual shod runners tend to maintain a widened MLA and throughout all of stance phase (0-100% of stance).

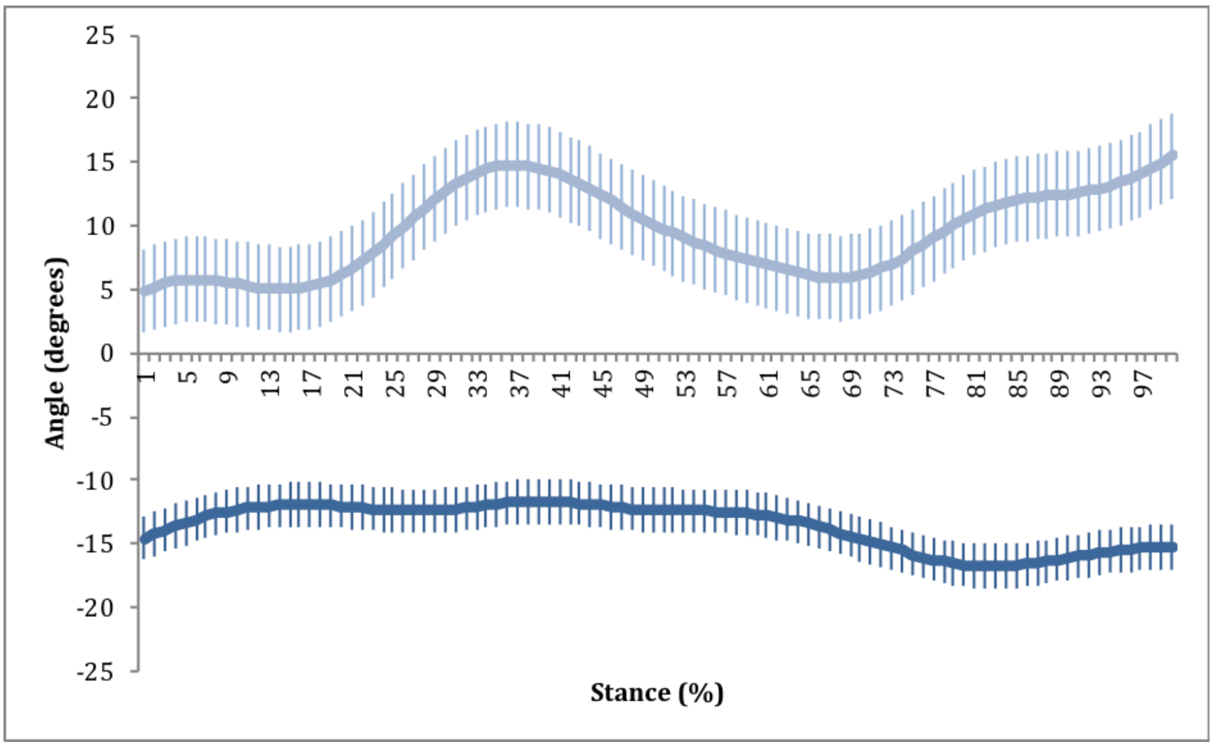


Figure 4.8 Transverse plane forefoot angle with respect to the midfoot during the stance phase of running in the VFF condition. Positive angles reflect a narrowing in the MLA and negative angles are seen as the widening of the MLA. The navy and light blue represent the shod and VFF conditions respectively.

4.2.4 Hindfoot in the frontal plane (VFF)

The motion of the hindfoot for the habitual shod runners was not significantly different from the habitual VFF runners when running in VFF, as shown by the pronation and supination ($p > 0.05$; Figure 4.9). From foot strike to toe off (0-100% of stance), the habitual shod runners peaked in supination at foot strike (0% stance; 7.4°) and the habitual VFF runners peaked in pronation at midstance (-5.9°).

The shapes of the two averaged curves do not follow the same pattern. The habitual VFF runners are pronated at the calcaneus during foot strike, followed by supinated throughout the midstance phase and lastly, return to a pronated calcaneus upon toe off. The habitual shod runners have a supinated calcaneus followed by a pronated calcaneus up until toe off when it returns to being supinated.

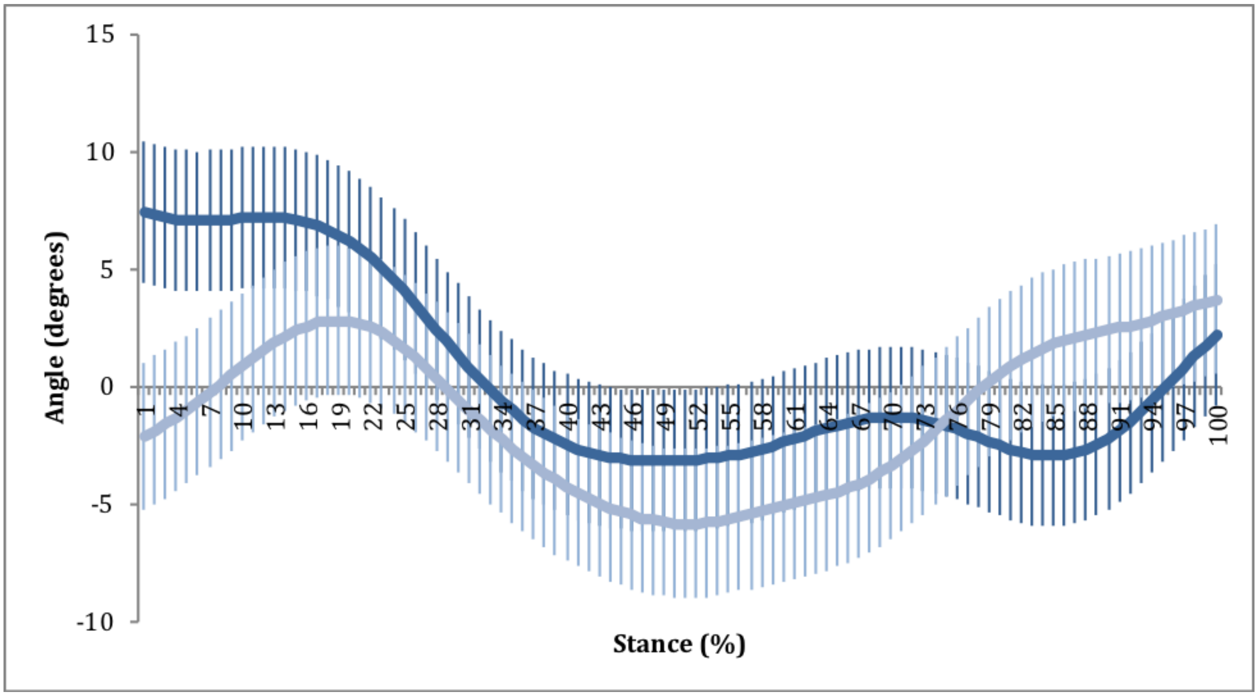


Figure 4.9 Frontal plane hindfoot angle with respect to the midfoot during the stance phase of running in the VFF condition. Positive angles reflect supination and negative angles reflect pronation. The navy and light blue represent the shod and VFF conditions respectively.

4.2.5 Hindfoot in the transverse plan (VFF)

The motion of the hindfoot for the habitual shod runners was not significantly different from the habitual VFF runners when running in VFF, as shown by the adduction and abduction ($p > 0.05$; Figure 4.10). The habitual shod runners peaked at -8.2° abduction and habitual VFF runners peaking at 1.9° adduction.

The shapes of the two averaged curves do not follow the same pattern. The calcaneus of the habitual VFF runners was adducted throughout the stance phase until it abducted on toe off and the calcaneus of the habitual shod runners was abducted, with the calcaneus becoming increasingly abducted towards toe off.

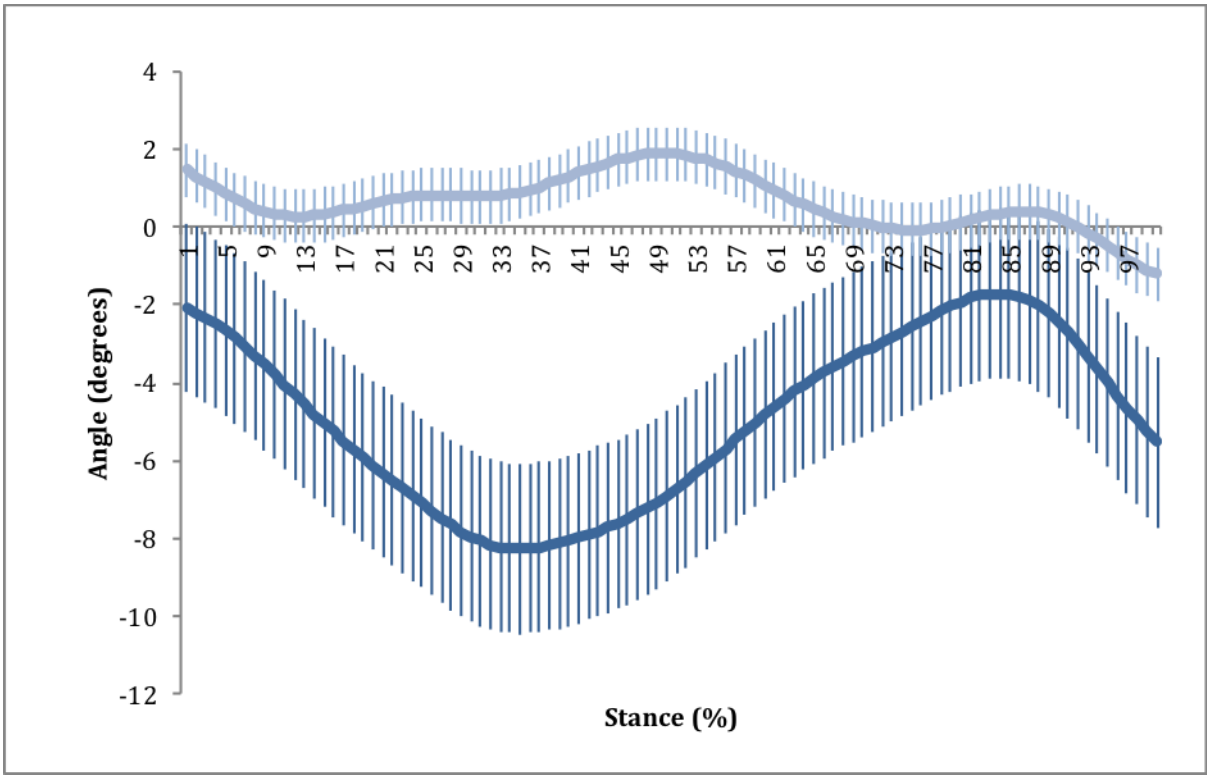


Figure 4.10 Transverse plane hindfoot angle with respect to the midfoot during the stance phase of running in the VFF condition. Positive angles reflect adduction and negative angles reflect abduction. The navy and light blue represent the shod and VFF conditions respectively.

Table 4.1 Habitual shod runner trials compared to habitual VFF runners in barefoot (BF) and VFF conditions. For the hypothesis agreement to occur, the habitual shod running gait should be significantly different from the habitual VFF running gait.

	Type	Habitual Shod		Habitual VFF		Difference		95% CI for diff		t	p	Hypothesis Agreement
		Mean	SD	Mean	SD	Mean	SD	Lower	Upper			
MLA (dimensionless)	BF	-0.053	0.133	-0.049	0.328	-0.003	0.177	-0.50	0.49	-0.020	0.985	No
Forefoot frontal (°)	BF	-0.624	1.24	-9.99	6.20	9.38	3.16	-0.27	19.0	2.97	0.054	No
Forefoot transverse (°)	BF	8.39	19.5	16.9	27.9	-8.50	17.0	-51.36	34.4	-0.499	0.637	No
Hindfoot frontal (°)	BF	-4.73	17.6	1.97	8.21	-6.70	9.70	-33.0	19.6	-0.690	0.526	No
Hindfoot transverse (°)	BF	-3.08	14.3	7.15	3.47	-10.2	7.33	-32.3	11.8	-1.39	0.249	No
MLA (dimensionless)	VFF	-0.162	0.23	-0.054	0.24	-0.11	-0.11	-0.52	0.304	-0.643	0.643	No
Forefoot frontal (°)	VFF	5.57	14.9	-11.3	10.18	16.9	9.01	-5.87	39.7	1.88	0.116	No
Forefoot transverse (°)	VFF	-13.6	20.4	9.54	29.1	-23.1	17.8	-67.8	21.6	-1.30	0.247	No
Hindfoot frontal (°)	VFF	-4.89	19.4	0.64	10.6	-5.54	11.0	-34.5	23.5	-0.502	0.639	No
Hindfoot transverse (°)	VFF	0.49	19.3	-1.16	11.4	1.62	11.2	-27.5	30.7	0.144	0.891	No

Chapter 5: Discussion & Conclusions

5.1 Discussion

We have observed that there were trends towards significant differences in the forefoot kinematics when running barefoot between the habitual shod and VFF runners as exhibited by pronation and supination. There were no additional significant differences between habitual shod runners and VFF when running either in barefoot or VFF. These findings add to literature as injuries can be seen in any type of running and one type of running may not be associated with a higher risk of injury. This also supports findings that injury rates were not significantly different between VFF and shod running (Altman & Davis, 2016).

Previous studies between barefoot and shod foot kinematics did not define movement of the forefoot and hindfoot relative to the midfoot (James et al., 1978; Wolf et al., 2008). One great benefit of our study was the specific definition of the midfoot, which, allowed for movements of the forefoot, hindfoot and subtalar joint to simultaneously be captured. Most importantly, splitting the forefoot into medial and lateral aspects allowed for pronation/supination to be quantified. There is little literature studying forefoot movement (Messier et al., 2018).

A total of 23 kinematic variables were calculated, representing the motions of the five segments of the foot in three planes. These kinematic variables represent data to clinicians to better understand how segments of the foot move in relation to each other in various planes which could give insight into biomechanical causes of injury.

5.2 MLA Kinematics

The MLA height-to-length ratios of the shod and VFF runners do not follow a similar pattern during the stance phase when running in VFF or barefoot. Both the shod and VFF runners exhibited narrowed arches on initial ground contact. This means that they have a widened arch during initial contact followed by a narrowing of the arch during midstance. Approaching toe off, the shod runners widened their arches and the VFF runners narrowed their arches.

The pattern of arch movement between the VFF runners and shod runners in VFF is different, suggesting that perhaps running in VFF are not the same for habitual shod and VFF runners. This could emphasize the need for the gradual implementation of VFF into training rather than abrupt changes in footwear. The results show that habitual VFF and shod runners should be cautious when barefoot running and this aligns with previous research (Warne & Gruber, 2017).

The pattern of arch movement between the VFF runners and shod runners in barefoot is different, suggesting that perhaps running barefoot is not the same for habitual shod and VFF runners. This could emphasize the need for the slow implementation of barefoot running into training.

The MLA has implications for running as Robbins & Hanna (1987) suggest that MLA shortening or rising (i.e. positive magnitude) allows the foot to act as an impact absorber, decreasing force and impact and thus decreasing potential injury. The same study suggests that MLA shortening aids in the prevention of plantar fasciitis due to the increased shock absorption capabilities of the arch when shortened (Robbins & Hanna, 1987). Knowing that plantar fasciitis affects a large portion of runners (Macera et al., 1989; Taunton et al., 2002; van Gent et al., 2007), and that barefoot runners have a greater risk of fasciitis (Morales-Orcajo et al., 2018), determining shoe mechanics which aid in MLA shortening may improve injury prevalence in long distance runners.

Our results show that habitual VFF runners running in VFF and barefoot do not have the same patterns of MLA height to length ratio as habitual shod runners. The results of the habitual shod runners agree with the findings that forefoot striking during stance phase increases arch height, during ground contact or load acceptance during running (Perl et al., 2012). This increasing of the arch height seen could increase the risk of plantar fasciitis, knee and ankle pain (Robbins & Hanna, 1987; Pohl, 2009). Barefoot runners likely experience arch shortening or narrowing following 4 months of integrating barefoot training (Nachbauer & Nigg, 1992; Robbins & Hanna, 1987). This could have implications to responses to training as this may be a measure for preventing the development of plantar fasciitis (Robbins & Hanna, 1987; Nachbauer & Nigg, 1992).

This study was in line with findings by Shroyer at al., (2011), there were no significant changes in arch height between VFF and shod running. This could allude to the idea that arches will not fail when wearing minimalist shoes. Therefore, runners trained in habitual shod and VFF are subjected to the same injury factors (Shroyer, Etheredge, & Weimar, 2011).

MLA arch height has implications as it seems that there are indications that habitual barefoot runners can train the MLA. This would be done via strengthening of the intrinsic foot muscles using VFF. This would cause the MLA to shorten during the stance phase. However, this preliminary result indicates that habitual barefoot runners can utilize VFF as a training modality for barefoot running, as the movement patterns of the MLA are not significantly different.

5.3 Forefoot kinematics

There was a trend towards significant differences between the habitual shod and VFF runners when running barefoot. Shod runners exhibited a neutral stance on ground contact and pronated their feet at toe off. VFF runners exhibited supination on ground contact and remained in supination on toe off.

There was no significant difference between the habitual shod and VFF runners when running in VFF. However, shod runners exhibited a pronated forefoot on ground contact and remained pronated at toe off. VFF exhibited supination on ground contact and remained supinated on toe off.

With regards to forefoot kinematics, there is some information on forefoot movement during running (Morio et al., 2009), but less literature that specifically compares the forefoot movement of a heel striker and a forefoot striker (Soares et al., 2018; Takabayashi, Edama, Yokoyama, & Kanaya, 2018). The foot is often referred to as one rigid segment. This quantifies how the foot moves with respect to the shank rather than exploring the forefoot as a separate segment allowing for the exploration of pronation/supination and forefoot spreading.

There is a decreased ground contact time seen in runners with a forefoot strike (i.e. VFF and barefoot runners), compared to those with a rearfoot or midfoot strike (Nunns et al., 2013), however, forefoot striking also increases pressure put on the metatarsal heads (Morales-Orcajo et al., 2018). This increased pressure has shown to increase plantar connective tissue stress which can lead to plantar fasciitis (Chen, Wong, Wang, Lin, & Zhang, 2019).

Pronation seen in shod running in both barefoot and VFF conditions emphasizes the need for shod runners to slowly integrate barefoot mimicking or barefoot running into training while switching to a forefoot or midfoot strike (Warne & Gruber, 2017) as it may alter their biomechanics and could lead to injury.

These findings are not in alignment with (Kurup, Clark, & Dega, 2012), who stated that regular users of soled shoes tend to show more forefoot supination. Upon toe off, our results are mixed compared with previous research which stated that supination occurs to propel the body forwards (Dugan & Bhat, 2005; Rodgers, 1988).

The habitual shod runners (i.e. rearfoot strikers), typically have slightly supinated forefoot (Rodgers, 1988), which was evident in barefoot but not in VFF running. Rearfoot strikers have an increased amount of forefoot pronation, which was evident with the habitual shod runners running barefoot and VFF (Stacoff, Kaelin, Stuessi, & Segesser, 1989). This is further supports the necessity for habitual shod runners to slowly implement barefoot running into training routines.

There was no significant difference seen between the habitual shod and VFF runners when running barefoot. Shod and VFF runners exhibited similar patterns of arch widening on ground contact, followed by arch narrowing during midstance, and then were opposite on toe off.

There was no significant difference seen between the habitual shod and VFF runners when running in VFF. Shod and VFF runners exhibited different patterns. The habitual shod runners showed arch widening on ground contact, which increased until midstance, before narrowing and then widening on toe off. The habitual VFF runners showed arch narrowing on ground contact and remained consistent until toe off. It appears as though VFF in all conditions changed the way that the MLA rises and falls during the stance phase and more specifically, during toe off. This is inconsistent with findings that barefoot runners exhibit a wider arch on foot strike (De Wit et al., 2000).

While the VFF runners raise the arch approaching toe off by tightening the windlass mechanism (i.e. extending the hallux), the habitual shod runners do the opposite in both barefoot and VFF. Shod runners exhibited an even flatter arch at toe off when running in VFF and barefoot. This is inconsistent with previous findings (De Wit et al., 2000), and is important as tightening and narrowing of the arch is necessary to propel the body forward into the swing phase of running gait. Our results indicate that the VFF runners effectively use the windlass mechanism, and that shod runners do not, both in running barefoot and in VFF. In both cases, the habitually shod runners had widened arches, which could suggest that intrinsic muscle training happens when habitually running in VFF that increases the propulsion caused by the windlass mechanism. The widening of the arch is also attributed to larger external loading rates and can be assumed that this is a mechanism to reduce initial pressure on ground contact (De Wit et al., 2000). This could have implications on injuries for habitual shod runners running barefooted or in VFF as they have not adopted this compensation strategy.

5.4 Hindfoot Kinematics

There was no significant difference between the habitual shod and VFF runners when running barefoot. Movement patterns were similar except on initial ground contact where VFF runners exhibited supination and habitual shod runners were pronated.
There was no significant difference seen between the habitual shod and VFF runners when running in VFF. Movement patterns were similar except on initial ground contact where shod runners exhibited supination and habitual VFF runners were pronated.

With regards to the hindfoot, it appears as though conditions peak in pronation around midstance. This data provides support that shod and VFF runners do not mimic hindfoot movement patterns of each other.

Typically, movement of the foot while running follows the same pattern: supination during foot strike, followed by pronation halfway through stance and then supinates until toe off (Clark, Frederick, & Hamill, 1983). The pronation seen by shod runners running barefoot and the VFF running in VFF is a means of decreasing force upon ground contact and subsequently the force placed on the body (Clark, 1983; Dugan, 2005). Excessive pronation causes the tibia to rotate medially, and can be connected to increased rates of injuries of the lower body (Bates et al, 1978; James et al, 1978; Nigg & Luethin, 1980; Schuster, 1978; Subotnick, 1981; Clark, 1983). This could have implications for the VFF running barefoot and shod runners running in VFF as they were supinated on ground contact and suggest that they are not the same. This suggests and goes against findings that the body alters the foot strike to limit ground reaction forces as a mechanism to reduce injury (Clark et al., 1983; Dugan & Bhat, 2005).

It is important to note that research done by Pohl et al. (2006) found that there was a lack of coupling between rearfoot inversion/eversion and forefoot inversion/eversion while running barefoot. This means that the frontal plane motion of the forefoot has little to do with movements that occur in the frontal plane of the hindfoot. Anatomically during foot strike, the heel hits the ground and causes the foot to immediately supinate after which it pronates (Clarke, 1983). This was observed in the shod runners running in VFF only, but was not consistent with results seen by the VFF runners running either barefoot or in VFF. This indicates that VFF do not mimic the natural movement pattern of the foot during running. This is congruent with findings stating that the more stiff a shoe, the more the natural movement of the foot was constrained (Stacoff, et al. 1989;1991). However, in comparison to what is stated above with respect to overall intersegmental foot movement over the stance phase, results do agree with work done by Nachbauer & Nigg (1992), stating that increased ankle supination is used to decrease force when runners are adapted to running barefoot. This study attributed this increased ankle supination to neuromuscular control mechanisms that work to keep the force at the same magnitude, thus injury rates should not be different (Nachbauer & Nigg, 1992). These mechanisms include larger knee flexion at initial ground contact, increased ankle supination at ground contact and larger pronation (Nachbauer & Nigg, 1992). This may lead towards VFF mimicking barefoot running as increased ankle supination is present.

Motion patterns of rearfoot strikers show that the heel will be slightly supinated upon heel strike and at midstance, the foot will be pronated from 55-85% of the stance phase and the foot will supinate and return to neutral around 70-90% of the support phase (Rodgers, 1988). This was seen in the habitual shod runners running in VFF which suggests that foot strike patterns did not change from rearfoot to forefoot and VFF do not mimic barefoot running.

The pronation seen the conditions of the habitual shod runners running barefoot and the shod runners running in VFF could be a mechanism for injury as pronation is thought to cause knee pain and shin splits (Rodgers, 1988). Clark, et al. (1983) suggests that more than 13 degrees of pronation is excessive and could lead to injury. There were some shod runners running barefoot that were close to 13 degrees of pronation who should be cautious when barefoot running. Differences seen in shod runners between the barefoot and VFF running suggest that these are not similar.

Supination seen in the habitual shod runners running in VFF and the VFF running barefoot during the first half of stance phase is controlled by the tibialis posterior (Stackhouse, et al. 2004). Reducing the supination upon the initial phases of stance could alleviate injuries such as posterior tibial tendonitis (Stackhouse, et al. 2004).

The increases in hindfoot pronation seen with the habitual shod runners running barefoot, and the habitual shod and VFF runners running in VFF is similar to results found in a similar study and shown to be a factor in overuse injuries (Stacoff, et al. 1989; 1991). Movement of the forefoot has shown to impact pronation of the rearfoot and thus torsional stiffness needs to come from the foot and importantly, the shoe (Stacoff et al., 1989). Having a stiff shoe sole can decrease torsional movement, which, would lead to increased pronation and injuries (Stacoff et al., 1989). This suggests that habitual runners running barefoot or in VFF can be at risk for injury (Clark et al., 1983).

There were no significant differences seen in the habitual shod and VFF runners running barefoot. Movement patterns opposed each other with the shod runner being more abducted throughout the stance phase and the VFF runners being more adducted.

There were no significant differences seen in the habitual shod and VFF runners running in VFF. Movement patterns mirrored each other in that both trended to be more adducted on initial ground contact and abducted on toe off. In hindfoot adduction/abduction, shod runners appeared to mimic that of the VFF runners when running in VFF especially during the last quarter of the stance phase. The hindfoot of the habitual VFF runners maintains adduction, while habitual shod runners maintain abduction in both the VFF and barefoot conditions, which could help prevent injuries (Morio et al., 2009). This study found that the sole of the shoe constrained adduction, which is interesting to note that the shod runners experienced this when running barefoot and in VFF.

Results here disagree with that of Freychat et al., (1996), that there is less forefoot abduction and adduction when running barefoot. When comparing movement patterns during stance phase after initial impact, conditions were opposite in VFF and barefoot running. Opposite with results found by Morio et al., (2009), minimalist shoes seemed to interfere with the amount of abduction seen in this study with regards to habitual shod runners. These findings highlight that VFF may dampen the ability of the foot to abduct during the stance phase of running and suggesting that therefore there is may be an increased risk of injury from excessive adduction or abduction during running (Morio et al., 2009).

5.5 Limitations

Our study findings were limited to looking at barefoot and VFF conditions but it would be interesting to compare the findings to that of habitual VFF running shod. Due to time constraints, trials of shod running were not analyzed. Analysis of this condition would provide interesting comparisons.

There were limitations as we did not measure ground reaction force data. This data may provide insight in whether shod runners adapt a barefoot running pattern by decreasing force at time of impact. This would have relevance into the use of barefoot simulated shoes, as GRFs could be compared between shod, barefoot simulating shoes and barefoot running. The use of GRF data could add to integrating barefoot running into running plans based on injury studies.

Skin motion artifact, movement of markers may not be an accurate description of movements of the underlying bones.

There were limitations on current data that compared the movements of the forefoot separate to the hindfoot. Planes of motion are not usually studied. The lack of data should be further explored as the trend of barefoot running is becoming more prominent (Rodgers, 1988).

Camera placement affects the accuracy of motion capture and their ability to see and define markers that are in motion, small and close together. This was especially important on the medial side of the right foot. The MSFM requires a marker on the hallux, the shaft of the 1st metatarsal and the navicular, thus the three marker clusters appear in very close proximities especially during running. Moving the 12 cameras towards the lower half of the body may have been more accurate for viewing the trajectory of the markers, however this lab and camera set up was created and is used for the purpose of clinical gait analysis using the whole body. Having more habitual barefoot runners would be beneficial to represent typical foot motions of that group.

The relatively short running path (approximately 10 m) may have hindered the ability of the participants to achieve steady-state running mechanics.

The last limitation was the small sample size and thus a low statistical power. Our two sample groups each contained four subjects. In addition to these statistical considerations, the small sample size likely makes it difficult to extrapolate the findings to the general running population. It would be difficult for any conclusions or suggestions to be made in regards to injury prevention based on these findings.

5.6 Strengths

This five segment foot model allowed for the separate description of movements of the hindfoot, forefoot and midfoot in habitual shod and VFF runners. This model also allowed for transverse and frontal planes to be examined and specific movements to be studied. Thus we were able to explore movements in two planes and not simply a comparison of the forefoot relative to the hindfoot. This model also outputs 23 kinematic variables so a thorough comparison was achieved.

5.7 Conclusion

Overall, we observed altered movement patterns between the segments of the foot of habitual VFF runners compared to habitually shod runners. This study shows that VFF may affect the movement on the foot. This is seen with the trend for significant difference in forefoot pronation/supination between the habitual barefoot runners and VFF runners running barefoot. This suggests that VFF may constrain the way the forefoot and hindfoot are able to move due to the reduced torsional movement caused by a stiff shoe sole (Stacoff et al., 1989). Possibly, the elastic nature and webbing of the VFF is enough to add this reduced torsional movement. Motion of the hindfoot was not significantly different when transitioning from a running shoe to either barefoot or VFF. Together, this reflects that there is almost immediate adaptation in response to switches between running shoes. Given the biomechanical challenges of running barefoot or wearing VFF, these rapid adaptations reinforce the necessity to slowly implement this training modality. The slow implementation should decrease potential injury as shown with altered movement patterns. We suggest that habitual VFF and shod runners should be cautious when barefoot running. This recommendation aligns with previous research (Warne & Gruber, 2017). Runners should also be cautious when integrating VFF as a substitution to barefoot running.

References

Adelaar, R. S. (1986). The practical biomechanics of running. *Am J Sports Med*, *14*(6), 497–500.

Altman, A. R., & Davis, I. S. (2012). Gait & Posture A kinematic method for footstrike pattern detection in barefoot and shod runners. *Gait & Posture*, *35*(2), 298–300. doi:10.1016/j.gaitpost.2011.09.104

Altman, A. R., & Davis, I. S. (2016). Prospective comparison of running injuries between shod and barefoot runners. *British Journal of Sports Medicine*, *50*, 476–480. doi:10.1136/bjsports-2014-094482

Andriacchi, T. P., Alexander, E. J., Toney, M. K., Dyrby, C., & Sum, J. (1998). A point cluster method for in vivo motion analysis: applied to a study of knee kinematics. *Journal of biomechanical engineering*, *120*(6), 743–749.

Areblad, M., Nigg, B. M., Ekstrand, J., Olsson, K. O., & Ekström, H. (1990). Three-dimensional measurement of rearfoot motion during running. *Journal of biomechanics*, *23*(9), 933–940.

Bates, B. T., Osternig, L. R., Mason, B., & James, S. L. (1978). Lower extremity function during the support phase of running. *Biomechanics VI-b*, 30–39.

Bobbert, M. F., Schamhardt, H. C., & Nigg, B. M. (1991). CALCULATION OF VERTICAL GROUND REACTION FORCE ESTIMATES DURING RUNNING FROM POSITIONAL DATA. *Journal of Biomechanics*, *24*(12), 1095–1105.

Bonacci, J., Saunders, P. U., Hicks, A., Bonacci, J., Saunders, P. U., Hicks, A., Rantalainen, T., et al. (2013). Running in a minimalist and lightweight shoe is not the same as running barefoot: a biomechanical study. *British Journal of Sports Medicine*, *47*(6), 387–392. doi:10.1136/bjsports-2012-091837

Burkett, L. N., Kohrt, W. M., & Buchbinder, R. (1985). Effects of shoes and foot orthotics on VO2 and selected frontal plane knee kinematics. *Medicine and science in sports and exercise*, *17*(1), 158–163.

Cappozzo, A. (1991). Three-dimensional analysis of human walking: Experimental methods and associated artifacts. *Human Movement Science*, *10*(5), 589–602.

Cappozzo, A., Catani, F., Leardini, A., Benedetti, M. G., & Della Croce, U. (1996). Position and orientation in space of bones during movement: experimental artefacts. *Clinical biomechanics*, *11*(2), 90–100.

Caravaggi, P., Matias, A. B., Taddei, U. T., Ortolani, M., Leardini, A., & Sacco, I. C. (2019). Reliability of medial-longitudinal-arch measures for skin-markers based kinematic analysis. *Journal of biomechanics*, *88*, 180-185.

Catlin, M. J., & Dressendorfer, R. H. (1979). Effect of shoe weight on the energy-cost of running. *Medicine and science in sports and exercise* (Vol. 11, p. 80). WILLIAMS & WILKINS 351 WEST CAMDEN ST, BALTIMORE, MD 21201-2436.

Cauthon, D. J., Langer, P., & Coniglione, T. C. (2013). Minimalist shoe injuries: Three case reports. *The Foot*, *23*(2-3), 100–103. doi:10.1016/j.foot.2013.03.001

Cavanagh, P. R., Williams, K. R., & Clarke, T. E. (1981). COMPARISON OF GROUND REACTION FORCES DURING WALKING BAREFOOT AND IN SHORES. *Unknown Host Publication Title*. Univ Park Press.

Chambon, N., Delattre, N., Gueguen, N., Berton, E., & Rao, G. (2014). Is midsole thickness a key parameter for the running pattern? *Gait & Posture*, *40*(1), 58–63. doi:10.1016/j.gaitpost.2014.02.005

Chen, T. L., Wong, D. W., Wang, Y., Lin, J., & Zhang, M. (2019). Foot arch deformation and plantar fascia loading during running with rearfoot strike and forefoot strike : A dynamic finite element analysis. *Journal of Biomechanics*, *83*, 260–272. doi:10.1016/j.jbiomech.2018.12.007

Cheung, R. T.-H. (2009). *Investigation of Running Shoe Design on the Foot Kinematics, and Muscle Recruitment Pattern in People with Overpronation Problem*.

Clark, T. E., Frederick, E. C., & Hamill, C. L. (1983). The effects of shoe design parameters on rearfoot control during running. *Med Sci Sports Exerc*, *15*(5), 376–381.

Collier, R. (2011). The rise of barefoot running. *CMAJ*, *183*(1), 37–38. doi:10.1503/cmaj.109-3745

Curran, S., & Tozer, J. (2010). Do Vibram Fivefingers ® really mimic barefoot conditions ? A study examining walking efficiency. *Journal of Foot and Ankle Research*, *3*(1). doi:10.1186/1757-1146-3-S1-O9

De Wit, B., De Clercq, D., & Aerts, P. (2000). Biomechanical analysis of the stance phase during barefoot and shod running. *Journal of biomechanics*, *33*(3), 269–278.

Decker, M. J., Torry, M. R., Wyland, D. J., Sterett, W. I., & Steadman, J. R. (2003). Gender differences in lower extremity kinematics , kinetics and energy absorption during landing. *Clinical Biomechanics*, *18*, 662–669. doi:10.1016/S0268-0033(03)00090-1

Dickinson, J. A., Cook, S. D., & Leinhardt, T. M. (1985). The measurement of shock waves following heel strike while running. *Journal of biomechanics*, *18*(6), 415–422.

Divert, C, Mornieux, G., Baur, H., Mayer, F., & Belli, A. (2005). Mechanical Comparison of Barefoot and Shod Running. *Int J Sports Med*, *26*(7), 593–598. doi:10.1055/s-2004-821327

Divert, C, Mornieux, G., Freychat, P., Baly, L., Mayer, F., & Belli, A. (2008). Barefoot-Shod Running Differences: Shoe or Mass Effect? *Int J Sports Med*, *29*(6), 512–518. doi:10.1055/s-2007-989233

Divert, Caroline, Baur, H., Mornieux, G., Mayer, F., & Belli, A. (2005). Stiffness Adaptations in Shod Running. *Journal of Applied Biomechanics*, *21*, 311–321.

Dugan, S. A., & Bhat, K. P. (2005). Biomechanics and Analysis of Running Gait. *Phys Med Rehabil Clin N Am*, *16*(3), 603–621. doi:10.1016/j.pmr.2005.02.007

Firminger, C. R., Fung, A., Loundagin, L. L., & Edwards, W. B. (2017). Effects of footwear and stride length on metatarsal strains and failure in running. *Clinical Biomechanics*, *49*, 8-15.

Fong, A., Sinclair, P. J., Hiller, C., Wegener, C., & Smith, R. M. (2013). Impact attenuation during weight bearing activities in barefoot vs . shod conditions : A systematic review. *Gait & Posture*, *38*(2), 175–186. doi:10.1016/j.gaitpost.2012.11.017

Frederick, E. C. (1984). Physiological and ergonomics factors in running shoe design. *Applied Ergonomics*, *15*(4), 281–287. doi:https://doi.org/10.1016/0003-6870(84)90199-6

Frey, C. (2000). Foot Health and Shoewear for Women. *Clinical Orthopaedics and Related Research (1976-2007)*, *372*. Retrieved from https://journals.lww.com/corr/Fulltext/2000/03000/Foot_Health_and_Shoewear_for_Women_.5.aspx

Fukano, M., & Fukubayashi, T. (2009). Motion characteristics of the medial and lateral longitudinal arch during landing. *Eur J Appl Physiol*, *105*, 387–392. doi:10.1007/s00421-008-0915-3

Gantz, A. M., & Derrick, T. R. (2018). Kinematics and metabolic cost of running on an irregular treadmill surface.*Journal of sports sciences*, *36*(10), 1103-1110.

Goss, D. L., & Gross, M. T. (2012). A Review of Mechanics and Injury Trends Among Various Running Styles. *US Army Med Dep J*, 62–71.

Hamill, J., & Gruber, A. H. (2017). Is changing footstrike pattern beneficial to runners ? *Journal of Sport and Health Science*, *6*(2), 146–153. doi:10.1016/j.jshs.2017.02.004

Hashish, R., Samarawickrame, S. D., Sigward, S., Azen, S. P., & Salem, G. J. (2018). Lower-limb dynamics and clinical outcomes for habitually shod runners who transition to barefoot running. *Physical Therapy in Sport*, *29*, 93-100.

Hong, Y., Wang, L., Li, J. X., & Zhou, J. H. (2011). Changes in running mechanics using conventional shoelace versus elastic shoe cover. *Journal of Sports Sciences*, *29*(4), 373–379. doi:10.1080/02640414.2010.534805

Hreljac, A. (2004). Impact and Overuse Injuries in Runners. *Med Sci Sports Exerc*, *36*(5), 845–849. doi:10.1249/01.MSS.0000126803.66636.DD

Hreljac, A., Marshall, R. N., & Hume, P. A. (2000). Evaluation of lower extremity overuse injury potential in runners. *Med Sci Sports Exerc*, *32*(9), 1635–1641.

James, S. L., Bates, B. T., & Osternig, L. R. (1978). Injuries to runners. *Am J Sports Med*, *6*(2), 40–50.

Jenkyn, T R, & Nicol, A. C. (2007). A multi-segment kinematic model of the foot with a novel definition of forefoot motion for use in clinical gait analysis during walking. *Journal of Biomechanics*, *40*, 3271–3278. doi:10.1016/j.jbiomech.2007.04.008

Jenkyn, T R, Shultz, R., Giffin, J. R., & Birmingham, T. B. (2010). A comparison of subtalar joint motion during anticipated medial cutting turns and level walking using a multi-segment foot model. *Gait & Posture*, *31*, 153–158. doi:10.1016/j.gaitpost.2009.09.016

Jenkyn, Thomas R, & Anas, K. (2009). Foot Segment Kinematics During Normal Walking Using a Multisegment Model of the Foot and Ankle Complex. *J Biomech Eng*, *131*(3). doi:10.1115/1.2907750

Jungers, William L. "Biomechanics: Barefoot running strikes back." *Nature* 463, no. 7280 (2010): 433.

Kindred, J., Trubey, C., & Simons, S. M. (2011). Foot Injuries in Runners. *Curr Sports Med Rep*, *10*(5), 249–254.

Kurup, H. V., Clark, C. I. M., & Dega, R. K. (2012). Footwear and orthopaedics. *Foot and Ankle Surgery*, *18*(2), 79–83. doi:10.1016/j.fas.2011.03.012

Leardini, A., & Caravaggi, P. (2017). Kinematic Foot Models for Instrumented Gait Analysis. *Handbook of Human Motion*, 1-24.

Leardini, A., Chiari, L., Croce, U. D., & Cappozzo, A. (2005). Human movement analysis using stereophotogrammetry Part 3. Soft tissue artifact assessment and compensation. *Gait & Posture*, *21*, 212–225. doi:10.1016/j.gaitpost.2004.05.002

Lieberman, D. E., Castillo, E. R., Otarola-Castillo, E., Sang, M. K., Sigei, T. K., Ojiambo, R., ... & Pitsiladis, Y. (2015). Variation in foot strike patterns among habitually barefoot and shod runners in Kenya. *PLoS One*, *10*(7), e0131354.

Lieberman, D. E., Venkadesan, M., Werbel, W. A., Daoud, A. I., D'Andrea, S., Davis, I. S., Mang'Eni, R. O., et al. (2010). Foot strike patterns and collision forces in habitually barefoot versus shod runners. *Nature*, *463*, 531–535. doi:10.1038/nature08723

Lovett, R.A. (2014, August 29). *Daniel Lieberman, 10 Years After "Born to Run"*. Retrieved from https://www.runnersworld.com/advanced/a20811582/daniel-lieberman-10-years-after-born-to-run/

Macera, C. A., Pate, R. R., Powell, K. E., Jackson, K. L., Kendrick, J. S., & Craven, T. E. (1989). Predicting Lower-Extremity Injuries Among Habitual Runners. *Arch Intern Med*, *149*, 2565–2568.

Mann, R. A., Baxter, D. E., & Lutter, L. D. (1981). Running Symposium. *Foot & Ankle*, *1*(4), 190–224. doi:10.1177/107110078100100402

Mann, R., & Inman, V. T. (1964). Phasic Activity of Intrinsic Muscles of the Foot. *JBJS*, *46*(3). Retrieved from https://journals.lww.com/jbjsjournal/Fulltext/1964/46030/Phasic_Activity_of_Intrinsic_Muscles_of_the_Foot.1.aspx

Martin, E. A., & McFerran, T. A. (2014). *A dictionary of nursing*. Oxford University Press.

Maslen, B. A., & Ackland, T. R. (1994). Radiographic study of skin displacement errors in the foot and ankle during standing. *Clinical Biomechanics*, *9*(5), 291–296. doi:https://doi.org/10.1016/0268-0033(94)90041-8

McCarthy, C., Fleming, N., Donne, B., & Blanksby, B. (2014). 12 Weeks of Simulated Barefoot Running Changes Foot-Strike Patterns in Female Runners. *Int J Sports Med*, *35*(5), 443–450.

Mcpoil, T. G. (2000). Athletic Footwear : Design , Performance and Selection Issues. *Journal of Science and Medicine in Sport*, *3*(3), 260–267.

Mei-Dan, O., Kahn, G., Zeev, A., Rubin, A., Constantini, N., Even, A., Nyska, M., et al. (2005). The Medial Longitudinal Arch as a Possible Risk Factor for Ankle Sprains: A Prospective Study in 83 Female Infantry Recruits. *Foot Ankle Int*, *26*(2), 180–183.

Messier, S. P., Martin, D. F., Mihalko, S. L., Ip, E., Devita, P., Cannon, D. W., Love, M., et al. (2018). A 2-Year Prospective Cohort Study of Overuse Running Injuries The Runners and Injury Longitudinal Study (TRAILS). *The American Journal of Sports Medicine*, *46*(9), 2211–2221. doi:10.1177/0363546518773755

Milner, C. E., & Brindle, R. A. (2016). Reliability and minimal detectable difference in multisegment foot kinematics during shod walking and running. *Gait & posture*, *43*, 192-197.

Milner, C. E., Ferber, R., Pollard, C. D., Hamill, J., & Davis, I. S. (2006). Biomechanical Factors Associated with Tibial Stress Fracture in Female Runners. *Med Sci Sports Exerc*, *38*(2), 323–328. doi:10.1249/01.mss.0000183477.75808.92

Moody, D., Hunter, I., Ridge, S., & Myrer, J. W. (2018). Comparison of Varying Heel to Toe Differences and Cushion to Barefoot Running in Novice Minimalist Runners. *International Journal of Exercise Science*, *11*(1), 13–19.

Morales-Orcajo, E., Becerro, R., Vallejo, D. B., Losa, M., Bayod, J., Barbosa, E., & Casas, D. L. (2018). Engineering Foot internal stress distribution during impact in barefoot running as function of the strike pattern.

Computer Methods in Biomechanics and Biomedical Engineering, *21*(7), 471–478. doi:10.1080/10255842.2018.1480760

Morio, C., Lake, M. J., Gueguen, N., Rao, G., & Baly, L. (2009). The influence of footwear on foot motion during walking and running. *J Biomechanics*, *42*, 2081–2088. doi:10.1016/j.jbiomech.2009.06.015

Murphy, K., Curry, E. J., & Matzkin, E. G. (2013). Barefoot Running : Does It Prevent Injuries? *Sports Medicine*, *43*(11), 1131–1138. doi:10.1007/s40279-013-0093-2

Nachbauer, W., & Nigg, B. M. (1992). Effects of arch height of the foot on ground reaction forces in running. *Med Sci Sports Exerc*, *24*(11), 1264–1269.

Nigg, B. (2009). Biomechanical considerations on barefoot movement and barefoot shoe concepts. *Footwear Science*, *1*(2), 73–79. doi:10.1080/19424280903204036

Nigg, B., Bahlsen, A. H., Denoth, J., Luethi, S. M., & Stacoff, A. (1986). *Factors influencing kinetic and kinematic variables in running*. *Biomechanics of Running Shoes* (pp. 139–159).

Nigg, B. M. (2001). The Role of Impact Forces and Foot Pronation: A New Paradigm. *Clin J Sports Med*, *11*, 2–9.

Nordin, M., & Frankel, V. H. (2001). *Basic biomechanics of the musculoskeletal system*. Lippincott Williams & Wilkins.

Nunns, M., House, C., Fallowfield, J., Allsopp, A., & Dixon, S. (2013). Biomechanical characteristics of barefoot footstrike modalities. *Journal of Biomechanics*, *46*(15), 2603–2610. doi:10.1016/j.jbiomech.2013.08.009

Oae, K., Takao, M., Uchio, Y., & Ochi, M. (2010). Evaluation of anterior talofibular ligament injury with stress radiography , ultrasonography and MR imaging. *Skeletal Radiol*, *39*, 41–47. doi:10.1007/s00256-009-0767-x

Perl, D. P., Daoud, A. I., & Lieberman, D. E. (2012). Effects of Footwear and Strike Type on Running Economy. *Med Sci Sports Exerc*, *44*(7), 1335–1343. doi:10.1249/MSS.0b013e318247989e

Perry, J., Thorofare, K. S., & Davids, J. R. (1992). Gait Analysis: Normal and Pathological Function. *Journal of Pediatric Orthopaedics*, *12*(6). Retrieved from https://journals.lww.com/pedorthopaedics/Fulltext/1992/11000/Gait_Analysis__Normal_and_Pathological_Function.23.aspx

Pohl, M. B., & Buckley, J. G. (2008). Changes in foot and shank coupling due to alterations in foot strike pattern during running. *Clinical Biomechanics*, *23*(3), 334–341.

Pohl, M. B., Hamill, J., & Davis, I. S. (2009). Biomechanical and Anatomic Factors Associated With a History of Plantar Fasciitis in Female Runners. *Clin J Sports Med*, *19*(5), 372–276.

Rao, U. B., & Joseph, B. (1992). The influence of footwear on the prevalence of flat foot. A survey of 2300 children. *Bone & Joint Journal*, *74*(4), 525–527.

Reinschmidt, C., Van Den Bogert, A. J., Lundberg, A., Nigg, B. M., Murphy, N., Stacoff, A., & Stano, A. (1997). Tibiofemoral and tibiocalcaneal motion during walking: external vs. skeletal markers. *Gait & Posture*, *6*(2), 98–109.

Richards, C. E., Magin, P. J., & Callister, R. (2008). Is your prescription of distance running shoes evidence-based? *British Journal of Sports Medicine*, *43*, 159–162. doi:10.1136/bjsm.2008.046680

Robbins, S. E., & Hanna, A. M. (1987). Running-related injury prevention through barefoot adaptations. *Med Sci Sports Exerc*, *19*(2), 148–156.

Rodgers, M. M. (1988). Dynamic Biomechanics of the Normal Foot and Ankle During Walking and Running. *Phys Ther*, *68*, 1822–1830.

Rossi, W. A. (1999). Why shoes make “normal” gait impossible.

Ryan, M., Elashi, M., Newsham-West, R., & Taunton, J. (2013). Examining injury risk and pain perception in runners using minimalist footwear. *British Journal of Sports Medicine*, *48*, 1257–1262. doi:10.1136/bjsports-2012-092061

Saltzman, C. L., Nawoczenski, D. A., & Talbot, K. D. (1995). Measurement of the Medial Longitudinal Arch. *Phys Med Rehabil*, *76*, 45–49.

Shih, Y., Lin, K. L., & Shiang, T. Y. (2013). Is the foot striking pattern more important than barefoot or shod conditions in running?. *Gait & posture*, 38(3), 490-494.

Shorten, M. R. (2000). RUNNING SHOE DESIGN: PROTECTION AND PERFORMANCE. *Marathon Medicine*, 159–169.

Shroyer, J. F., Etheredge, C. E., & Weimar, W. H. (2011). Effect of minimalist footwear on medial arch height. *Medicine & Science in Sports & Exercise*, *43*(5), 313.

Shultz, R., & Jenkyn, T. (2012). Determining the maximum diameter for holes in the shoe without compromising shoe integrity when using a multi-segment foot model. *Medical Engineering and Physics*, *34*(1), 118–122. doi:10.1016/j.medengphy.2011.06.017

Sinclair, J. (2015). Barefoot and shod running: their effects on foot muscle kinetics. *The Foot and Ankle Online Journal*, *8*(2). doi:10.3827/faoj.2015.0802.0002

Sinclair, J., Butters, B., & Stainton, P. (2018). Acute Effects of Barefoot And Minimalist Footwear on Medial Tibiofemoral Compartment Loading During Running : A Statistical Parametric Mapping Approach by. *Journal of Human Kinetics*, *65*, 35–44. doi:10.2478/hukin-2018-0037

Soares, T., Oliveira, C. F. D., Pizzuto, F., Garganta, R. M., Vila-boas, J. P., Casimiro, M., Socorro, T., et al. (2018). Acute kinematics changes in marathon runners using different footwear. *Journal of Sports Sciences*, *36*(7), 766–770. doi:10.1080/02640414.2017.1340657

Squadrone, R., & Gallozzi, C. (2009). Biomechanical and physiological comparison of barefoot and two shod conditions in experienced barefoot runners. *Journal of Sports Medicine and Physical Fitness*, *49*(1), 6–13.

Stackhouse, C. L., Davis, I. M., & Hamill, J. (2004). Orthotic intervention in forefoot and rearfoot strike running patterns. *Clinical Biomechanics*, *19*, 64–70. doi:10.1016/j.clinbiomech.2003.09.002

Stacoff, A., Kaelin, X., Stuessi, E., & Segesser, B. (1989). The torsion of the foot in running. *International Journal of Sport Biomechanics*, *5*(4), 375–389.

Stacoff, A., Nigg, B. M., Reinschmidt, C., van denBogert, A. J., & Lundberg, A. (2000). Tibiocalcaneal kinematics of barefoot versus shod running. *J Biomechanics*, *33*, 1387–1395.

Takabayashi, T., Edama, M., Yokoyama, E., & Kanaya, C. (2018). Quantifying coordination among the rearfoot , midfoot , and forefoot segments during running. *Sports Biomechanics*, *17*(1), 18–32. doi:10.1080/14763141.2016.1271447

Taunton, J. E., Ryan, M. B., Clement, D. B., McKenzie, D. C., Lloyd-Smith, D. R., & D, Z. B. (2002). A retrospective case-control analysis of 2002 running injuries. *British Journal of Sports Medicine*, *36*, 95–101. doi:10.1136/bjsm.36.2.95

Tucker, R. (2014). The Vibram lawsuit, barefoot running and science perspective. *The Science of Sport*. Retrieved from https://sportsscientists.com/2014/05/vibram-lawsuit-barefoot-running-common-sense/

Van Mechelen, W. (1992). Running injuries. *Sports Medicine*, *14*(5), 320–335.

Waetjen, L., Parker, M., & Wilken, J. M. (2012). The effects of altering initial ground contact in the running gait of an individual with transtibial amputation. *Prosthetics and Orthotics International*, *36*(3), 356–360. doi:10.1177/0309364611433353

Wapner, K. L. (2000). Pes cavus. *Foot and ankle disorders*, *2*, 919.

Warburton, M. (2001). Barefoot Running. *Sportscience*, *5*(3).

Warne, J. P., & Gruber, A. H. (2017). Transitioning to Minimal Footwear : a Systematic Review of Methods and Future Clinical Recommendations. *Sports Medicine*, *3*(33). doi:10.1186/s40798-017-0096-x

Wikler, S. J. (1961). *Take Off Your Shoes and Walk: Steps to Better Foot Health*. Devin-Adair Company.

Willy, R. W., & Davis, I. S. (2013). Kinematic and kinetic comparison of running in standard and minimalist shoes.

Wolf, S., Simon, J., Patikas, D., Schuster, W., Armbrust, P., & Doderlein, L. (2008). Foot motion in children shoes — A comparison of barefoot walking with shod walking in conventional and flexible shoes. *Gait & Posture*, *27*, 51–59. doi:10.1016/j.gaitpost.2007.01.005

Zhang, X., Paquette, M. R., & Zhang, S. (2013). A comparison of gait biomechanics of flip-flops , sandals , barefoot and shoes. *Journal of Foot and Ankle Research*, *6*(45). doi:10.1186/1757-1146-6-45

van Gent, R. N., Siem, D., van Middelkoop, M., van Os, A. G., Bierma-Zeinstra, S. M. A., & Koes, B. W. (2007). Incidence and determinants of lower extremity running injuries in long distance runners: a systematic review. *British Journal of Sports Medicine*, *41*(8), 469–480. doi:10.1136/bjsm.2006.033548

Appendix A

Letter of information (Participant):
Barefoot running and barefoot-mimicking biomechanics study

Principal Investigator: Dr. Tom Jenkyn
Department of Mechanical and Materials Engineering, Western University
Study site: WOBL Lab, Room 1215, 3M, Western University

INTRODUCTION

You are invited to participate in a research study looking at how barefoot-mimicking running shoes impact running kinematics and performance. The purpose of this letter is to provide you with the information you require to make an informed decision on whether you would like to participate in this research.

This study is important as running related injuries are often associated with excessive motion of the segments of the foot. Despite advances in running shoe technology and design, the rates of repetitive injury are unchanged [1]. Barefoot running is a relative new technology and the science has become a popular topic in recent years with the introduction of shoes such as the Nike Free and VFF as some of the popular barefoot running simulating shoes.

Barefoot running and barefoot running shoes aim to mimic the shape of the foot, mimic the movement of the foot during barefoot running or mimic the feeling of barefoot running [2]. It is thought that designing shoe types for specific types of feet will help decrease running injuries caused by excessive motion of the foot. A better understanding of how the foot moves and shoe's motion control systems will decrease these injuries.

This project is conducted by a Western University MSc candidate (Lisa Oikawa) under the supervision of Dr. Tom Jenkyn.

SUMMARY EXPLANATION OF RESEARCH
The aim of this study is to gain insight into barefoot running and barefoot running mimicking shoes. There will be 50 habitual barefoot adult runners that will perform various running tasks. To observe the movements we will use an optical motion capture system that uses 10 cameras and a series of markers that attach to land marks on your body.

DESCRIPTION OF THE RESEARCH PROCEDURES
The trials will involve recording body and shoe or foot kinematics while you perform a series of brief runs. Things that you need to bring to the testing session are your own shorts and a t-shirt. Reflective markers will be placed with hypoallergenic tape and will not interfere with your movements. There will also be markers located on the shoes. You will

be performing a series of 4 basic runs in an indoor facility. These short runs will be recorded using video and motion capture cameras. The video will assist in the analysis of barefoot running and barefoot running mimicking shoe kinematics. This experiment will require one session approximately 30 minutes in duration.

MARKER PLACEMENT

For each gait analysis, each subject will wear the modified Helen Hayes marker set (standard for motion analysis on humans) consisting of 22 passive reflective markers, which will allow the motion analysis system to record their movements. The markers are attached with double sided adhesive discs over the following landmarks: left and right shoulder, right scapula, left and right elbow, left and right wrist, left and right hips, sacrum, left and right knee, left and right lateral ankle, left and right toe of shoe, left and right heel of shoe, left and right thigh, and left and right shank wand. The feet are tracked with an additional 'Multi-segmented Foot Model' marker set. Clusters of three markers are placed on the hallux, first metatarsal, fifth metatarsal, and the heel. A single marker is placed on the navicular. These markers are attached with the same adhesive as all other markers.

PUBLICATION OF RESULTS

If you wish, sample data will be reviewed with you at the end of the session. You will have the opportunity to consent today to receive a copy of the final results of this study. If you indicate that you would like to receive the final results of this study, then we will send you this data via e-mail. Results of this research will be published in scientific journals and presented at conferences.

BENEFITS

You will not be compensated for your time. Your participation will help us obtain knowledge that may minimize future running injury through modifying running shoe and barefoot running mimicking shoe design.

RISK/HARMS

There is a minimal risk of injury when performing these experiments. There will be no obstacle or uneven terrain, and the number of repetitions is limited; therefore, the risk of injury is below that of an outdoor run.

LOCATION

The testing will take place in the Wolf Orthopaedic Biomechanics Lab (WOBL) at Western University

PARTICIPATION

Your participation is strictly voluntary. You may refuse to participate, refuse to answer any questions or withdraw from the study at any time with no effect on your athletic, academic or employment status. If you are participating in another study at this time, please inform the researchers right away to determine if it is appropriate for you to participate in this study.

CONFIDENTIALITY

We hope to learn more about running biomechanics and begin to understand how barefoot mimicking equipment affects performance. You will not directly benefit from this study, but may receive insight on how shoes and running barefoot affect your performance. Kinematic and video data will be stored on an external hard drive and locked in a secure filing cabinet in the Joint Biomechanics Laboratory at Western University. You will not be identified in these files. Paper files containing your identity will also be stored in a locked filing cabinet. These data will be retained for five years so that we can use it for future analyses as well as for illustrations in scientific meetings, scientific manuscripts and potential teaching opportunities. We will obscure your face in these images in order to protect your confidentiality. Representatives of the Western University Health Sciences Research Ethics Board may contact you or require access to your study-related records to monitor the conduct of the research. There are two copies of this consent form; one which the researcher keeps, and one that you keep.

The purpose of collecting your full name and partial date of birth is to give identifiers for future work. This study is the first of many observing barefoot running mimicking biomechanics; therefore, we request consent to use your data in future studies and to contact you for future research. If you choose to not give permission to use your data for future studies or to be contacted for future studies your data and personal information will be disposed of accordingly. Your decision on this matter has no effect on your athletic, academic or employment status

NO WAIVER OF RIGHTS

You do not waive any legal rights by signing the consent form.

CONTACT PERSON

If you have any questions or concerns about the study or about being a subject, you should contact the principal investigator, Dr. Tom Jenkyn, Department of Mechanical and Materials Engineering, Western University, _______. If you have any questions about your rights as a research participant or the conduct of the study you may contact the Office of Research Ethics ___________, email _________.

This letter is for you to keep.
You will be given a copy of this letter of information and consent form once it has been signed.

Consent Form: Barefoot running and barefoot-mimicking biomechanics study
I have read the letter of information, have had the nature of the study explained to me and I agree to participate. All questions have been answered to my satisfaction.

Participant's Signature: ____

______________________ ________________
Printed name Date:

Do you wish to receive a copy of the results of this study?
☐ No ☐ Yes

If Yes, please provide your e-mail address: ____________________________

Do you consent to the use of video cameras during the testing session?
☐ No ☐ Yes

Do you consent to using your data for future research projects?
☐ No ☐ Yes

Can we contact you to participate in future studies?
☐ No ☐ Yes

If Yes, you may change your mind and withdraw your data at a future time by contacting Dr. Tom Jenkyn at the above address.

____

Do you consent to us using de-identified images from the video for scientific presentations, scientific manuscripts or for purposes of teaching?
☐ No ☐ Yes

Person Obtaining Informed Consent:

Signature: ____

______________________ ________________
Printed name Date:

Appendix B

Research Participant Recruitment

THE WOLF ORTHOPAEDIC BIOMECHANICS LABORATORY IS LOOKING FOR PARTICIPANTS!

THE STUDY:
Looks at how the bones of 'normal' feet interact during barefoot running and within barefoot-mimicking running shoes. Measures of the change in bone motion in the foot will be done by comparing the results from 'optical motion capture' using the multi-segment foot model.

WHAT YOU HAVE TO DO:
You will have to come in for half an hour where we will stick reflective markers on several locations on your foot and body with double-sided tape. You will be asked to stand still and run a series of indoor trials.

WHO TO CONTACT:
Send an email to ________

Appendix C

<u>Telephone Script for Recruitment</u>

Hello, may I please speak with {insert the name of the potential participant here}.
If the potential participant is not home ask if there is a better time to call. Do not leave a message as it may be a confidential matter you are calling about that may not be apparent to you
If they are home, continue with the conversation

Hi, {insert the name of the potential participant here} this is Lisa Oikawa calling from the Wolf Orthopaedic Biomechanics Laboratory at Western University.
I am calling today to ask if you are interested in a research study that we are conducting.

The study is being conducted by Dr. Thomas Jenkyn and will look at how the bones in the foot interact both during barefoot running and while wearing barefoot-mimicking running shoes. This project is designed to accurately measure the change in bone motion in the foot by comparing the results from 'optical motion capture' using the multi-segment foot model. You will be required to come in for half an hour where we will stick reflective markers on several locations on your foot and body with double-sided tape. You will be asked to stand still and run a series of indoor trials. Would you be interested in hearing more about this study?

If no, thank them for their time and say good-bye
If yes, continue to explain the study details to them based on the letter of information
I am now going to read you the letter of information over the phone {Clearly read the letter of information the participant over the phone}
Do you have any questions?
{Answer any questions they may have}
Do you agree to participate in this study?
**If yes, continue with the study*
**If no, thank them for their time and say good-bye*

Appendix D

Summary of kinematic outcomes

Author	**Model**	**Running Conditions**	**Variable Studied & Definition**	**Outcome**
Bates, et al. (1978)	Stop-action projector *2 dimensional	Barefoot running & Nike shoes on treadmill	-Pronation: eversion of the calcaneus relative to shank -Supination: inversion of calcaneus relative to shank	-Pronation BF: began sooner and ended later (% stance phase) & time in pronation higher, max pronation occurs later -RF strike: upon contact heel is supinated, between heel strike and 20% stance phase the foot pronates until 85% -Max pronation occurs at 35-40% stance phase
De Wit, et al. (2000)	Video tape & skin mounted markers (body, calcaneus, shank, Achilles, 5th) *2 dimensional	Barefoot & Shod	-Sagittal plane: extension & flexion of calcaneus relative to shank -Frontal plane: posterior shank relative to Achilles and calcaneus	-BF: significantly smaller initial eversion at impact, higher step frequency, flatter foot at touchdown (higher plantar flexion) -No differences seen at toe off -Sagittal plane: consistent -Frontal plane: not consistent (overall; hindfoot less inverted)
Lieberman, et al. (2010)	Infrared camera system with reflective skin mounted markers *3 dimensional	Barefoot & Shod	-GRF and impact transients -Strike type	-Habitually shod adopt a flatter foot (dorsiflex 7-10 degrees less) -BF: MLA stretches passively during first half of stance -Shod: MLA stretches passively later in stance
Morio, et al. (2009)	Two segment foot model using 17 reflective markers	Barefoot & Shod (rearfoot strikers)	-Forefoot (metatarsals) to rearfoot (calcaneus) relative motion -Frontal plane: PF/DF	-BF: more eversion of the forefoot and it occurred faster than shod, high variability between subjects at push-off, sig. higher add -Sandals: range of eversion reduced by 20% and 60% in

			-Sagittal plane: inv/ev -Transverse plane: add/abd -Forefoot spreading (metatarsals)	late stance, forefoot inversion during pushoff, eversion is reduced -Sole of shoes constrain inv/ev and adduction -Shoes constrain forefoot spreading & pronation during toe off -Sandals constrain add/abd and do not follow natural motion
Wolf et al. (2008)				-Footwear constrains forefoot spreading, foot pronation during toe off -Forefoot to rearfoot add/abd may be dependent on flexibility of the upper covering of the shoe
Nachbauer & Nigg, (1992)	Video analysis of arch height with reflective markers	Shod	-MLA height -GRF	-Flattening values between 0.1 and 0.8 cm show evidence that foot structures change during stance phase in running -Rigidity of the foot due to low flattening could improve propulsion of the foot -Should be explored multi-segmentally
Nunns, et al. (2013)	Active markers; greater trochanter, L&M epicondyle, shank, gastroc, L malleolus, posterior calcaneus (2), M cuneiform, 3rd metatarsal, 5th MP joint	Habitually shod runners running barefoot	-Barefoot foot strike modality and pressure patterns	-Ground contact time is sig. shorter with a forefoot or toe strike -FFS & TS: ankle is plantarflexed at contact -HF & MF: ankle is dorsiflexed at contact -No differences in rearfoot kinematics -Runners did not adapt to barefoot running and simply used their typical running style
Perl, et al. (2012)	8 camera Oqus system: infrared markers; 1st met, navicular,	Habitually barefoot or minimalist runners running BF	-Arch strain	-FFS: flattened arch from contact to midstance and heightened slightly before toe off -RFS: arch was slightly

	M calcaneus, Achilles, L malleolus, L & M femoral epicondyle, fibula, greater trochanter			higher at impact and flattened until midstance and heightened slightly at toe off -Minimalist shoes may allow more elastic energy storage in the MLA by having greater movement
Pohl & Buckley, (2008)	Seven ProReflex cameras; 17 reflective markers- forefoot, rearfoot and shank	Habitually shod runners	-BF running with either a toe strike, forefoot strike or rearfoot strike	-Rearfoot strike: rearfoot eve, shank internal rotation, forefoot dorsiflexion, forefoot abduction sig. lower -RF eve/inv and forefoot abd/add are highly coupled -RF running: more inverted rearfoot during forefoot strike; magnitude HF-12°, FF-17.9° -RF running: eversion; magnitude HF-11.1° and FF-9.1° -FF: abd and add coupled with RF inv/eve (particularly in HF strike). Less forefoot abd during early stance -FF transverse plane motions are coupled with rearfoot frontal plane
Stackhouse, et al. (2004)	Retro-reflective markers; greater trochanters, lumbo-sacral joint, L & M femoral condyles, M & L malleoli, M & L borders of 1st & 5th, forefoot	Shod running	-Forefoot and rearfoot shod running	-RF running: more inverted rearfoot during forefoot strike; magnitude HF-13.7°, FF-16.4° -RF running: eversion; magnitude HF-10.5° and FF-8.8° -Tibialis posterior is responsible for control of inv/ev so orthotics may help prevent injuries
Robbins & Hanna, (1987)	Weight bearing platform & x-ray	BF running	MLA	-Shortening of the arch can only be attributed to activation of the intrinsic muscles of the arch -Skeletal muscular conditioning is a progressive change occurring over 2-3 months

				-Ability of the arch to change allows for dampening of impact and therefore decreases injury rates
Squadrone & Gallozzi, (2009)	Instrumented treadmill with preferred foot strike technique & video camera. Circular markers were used; acromion, greater trochanter, lat condyle, fibula, lat malleolus, calcaneus, 5th MP *2 dimensional	Habitually barefoot runners running in VFF and barefoot	-Foot angle relative to the ground -Ankle joint angle -Plantar flexion angle	-When given 10 days -Sig. more ROM with VFF (total) -Values and results of VFF were closer aligned to shod running -Lower extremity kinematics did not change

Appendix E

Physical Activity Readiness Questionnaire - PAR-Q (revised 2002)

PAR-Q & YOU

(A Questionnaire for People Aged 15 to 69)

Regular physical activity is fun and healthy, and increasingly more people are starting to become more active every day. Being more active is very safe for most people. However, some people should check with their doctor before they start becoming much more physically active.

If you are planning to become much more physically active than you are now, start by answering the seven questions in the box below. If you are between the ages of 15 and 69, the PAR-Q will tell you if you should check with your doctor before you start. If you are over 69 years of age, and you are not used to being very active, check with your doctor.

Common sense is your best guide when you answer these questions. Please read the questions carefully and answer each one honestly: check YES or NO.

YES	NO		
☐	☐	1.	**Has your doctor ever said that you have a heart condition and that you should only do physical activity recommended by a doctor?**
☐	☐	2.	**Do you feel pain in your chest when you do physical activity?**
☐	☐	3.	**In the past month, have you had chest pain when you were not doing physical activity?**
☐	☐	4.	**Do you lose your balance because of dizziness or do you ever lose consciousness?**
☐	☐	5.	**Do you have a bone or joint problem (for example, back, knee or hip) that could be made worse by a change in your physical activity?**
☐	☐	6.	**Is your doctor currently prescribing drugs (for example, water pills) for your blood pressure or heart condition?**
☐	☐	7.	**Do you know of any other reason why you should not do physical activity?**

If you answered

YES to one or more questions

Talk with your doctor by phone or in person BEFORE you start becoming much more physically active or BEFORE you have a fitness appraisal. Tell your doctor about the PAR-Q and which questions you answered YES.

- You may be able to do any activity you want — as long as you start slowly and build up gradually. Or, you may need to restrict your activities to those which are safe for you. Talk with your doctor about the kinds of activities you wish to participate in and follow his/her advice.
- Find out which community programs are safe and helpful for you.

NO to all questions

If you answered NO honestly to all PAR-Q questions, you can be reasonably sure that you can:

- start becoming much more physically active – begin slowly and build up gradually. This is the safest and easiest way to go.
- take part in a fitness appraisal – this is an excellent way to determine your basic fitness so that you can plan the best way for you to live actively. It is also highly recommended that you have your blood pressure evaluated. If your reading is over 144/94, talk with your doctor before you start becoming much more physically active.

DELAY BECOMING MUCH MORE ACTIVE:

- if you are not feeling well because of a temporary illness such as a cold or a fever – wait until you feel better; or
- if you are or may be pregnant – talk to your doctor before you start becoming more active.

PLEASE NOTE: If your health changes so that you then answer YES to any of the above questions, tell your fitness or health professional. Ask whether you should change your physical activity plan.

Informed Use of the PAR-Q: The Canadian Society for Exercise Physiology, Health Canada, and their agents assume no liability for persons who undertake physical activity, and if in doubt after completing this questionnaire, consult your doctor prior to physical activity.

No changes permitted. You are encouraged to photocopy the PAR-Q but only if you use the entire form.

NOTE: If the PAR-Q is being given to a person before he or she participates in a physical activity program or a fitness appraisal, this section may be used for legal or administrative purposes.

"I have read, understood and completed this questionnaire. Any questions I had were answered to my full satisfaction."

NAME ____

SIGNATURE ____ DATE ____________________

SIGNATURE OF PARENT ____ WITNESS ____________________
or GUARDIAN (for participants under the age of majority)

Note: This physical activity clearance is valid for a maximum of 12 months from the date it is completed and becomes invalid if your condition changes so that you would answer YES to any of the seven questions.

Supported by: Health Canada Santé Canada

continued on other side...

Appendix F

LO

LISA OIKAWA

SUMMARY OF SKILLS

- 5 years+ of professional working experience in a dynamic customer service environment
- 3 years of working experience in a C2B sales environment
- 2 years of working experience in a B2B sales environment
- 5 years + of working experience in CRM systems such as Salesforce and Microsoft Dynamics
- Highly proficient in the Microsoft suite
- Highly professional and experienced in communicating with consultants, industry professionals, the public community as well as managing partnerships and relationships

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WORKING EXPERIENCE

AEROTEK **May 2017 - Present**
Account Manager **June 2018 – Present**

- Responsible for business development in a B2B manner, gaining and maintaining client relationships within the professional services and engineering consulting field
- High focus on customer service, working within the 1/24 rule, getting back to every inquiry within the 1 hour mark and responding with a solution within 24 hours
- Responsible for managing a team of up to 5 people, developing, leading, training and performance managing work
- Responsible for multitasking and maintaining a high level of data integrity within Salesforce, updating account information, contact information and meeting notes
- Responsible for keeping up with KPIs, such as making over 75+ outbound calls, meeting with a minimum of 14 new and prospective clients per week
- Responsible for dealing with tough customer inquiries and quick decision making

Recruiter **May 2017 – May 2018**

- Responsible for creating relationships within the professional services, engineering and architecture community
- High focus on customer service, working within the 1/24 rule, getting back to every inquiry within the 1 hour mark and responding with a solution within 24 hours
- Responsible for keeping up with KPIs, such as making over 100+ contacts, 25+ phone screens or intake calls, 5+ meetings per week and updating all information
- Responsible for multitasking and maintaining a high level of data integrity within an internal CRM system
- Responsible for dealing with tough public inquiries and quick decision making

Kinetic Konnection **July 2014 – May 2017**
Orthopaedic Bracing Sales Consultant

- Responsible for working in a C2B environment, providing customer service, dealing with customer inquiries, maintaining and creating relationships with suppliers, vendors and members of the community
- Consulting clients based on their injuries, educating them on exercises, and providing the best solution possible whether that be an orthopaedic off the shelf bracing product or something custom
- Responsible for meeting sales objectives and reconversion sales by repeat customers
- Responsible for dealing with 20+ consultations daily, responding in a professional and courteous manner and then ensured customer satisfaction was maintained and exceeded through a follow up
- Providing injury support to the community, including handling referrals from doctors, physiotherapists and a variety of referrals dealing with community based programs such as ODSP, Ontario Works, WSIB and CAS
- Spearheaded the design of an office training manual for training in Microsoft Dynamics, to create clear and consistent use of client portfolios, set follow up tasks and ensure data integrity
- Responsible for developing targets and criteria to measure client follow up effectiveness and success of reconversion programs
- Providing support to sales representatives, accountant (invoicing and accounts receivable), responding to general inquiries about the business
- Developing charts, graphs and information manuals for internal and external customers

University of Western Ontario **September 2012 – April 2014**
Teaching Assistant

- Responsible for holding office hours to offer support to students outside of the classroom
- Responsible for responding to email inquiries about concepts, assignments and exams
- Leading a class of up to 30 students at a time
- Responsible for teaching Excel tutorials for data analysis, graphing and statistical analysis
- Design and implement interactive and educational tutorial sessions
- Responsible for grading assignments/tests and day to day queries about class lectures, tutorials and assignments

lululemon athletica **September 2012 – January 2013**
Educator

- Responsible for handling 20+ customer service situations daily, responding in a professional and courteous manner
- Providing and fostering a positive and inclusive workplace and community through community based programs and activities
- Educating and supporting customers on product and community events

Queen's University **September 2010 – April 2012**
Student Athletic Therapist

- Responsible for injury management of the Men's Varsity Football team and Women's Soccer team throughout their training camps through to the end of the season
- Supported Queen's University Women's Varsity Soccer Team to a CIS National Championship by following up on the status of athletes' conditions and progress with Head Athletic Therapists and followed through with suggested treatment plans
- Responsible for supporting the athletic therapy clinic on campus, providing customer service, treatment and report writing

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EDUCATION

- University of Western Ontario (MSc Biomechanics) Completion August 2019
- Queen's University (BScH SSP KIN Honours) 2012