KINEMATIC AND KINETIC DIFFERENCES BETWEEN SHOD AND BAREFOOT RUNNING

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ABSTRACT

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Despite the development of footwear technology, the incidence rate of running injuries remains high. The high prevalence of running-related injuries has led to a suggestion that the shoe-development approach is not working efficiently to reduce injuries. The alleged benefits of barefoot running, such as the potential for reduced risk of injury and a more economical approach, has resulted in the scientific community and media paying significant attention to barefoot running. Despite multiple differences found in kinematic and kinetic variables, there is no conclusive evidence proving or refuting the proposed advantages of barefoot running. On the other hand, there is no scientific evidence supporting the prescription of a shoe with an elevated cushioned heel and pronation control system. Therefore, the purpose of this thesis was twofold. First, to compare the kinematics and kinetics of barefoot and shod running. Second, to critically evaluate the significance of the findings with respect to the proposed advantages of barefoot running.

In total, nine healthy, habitually shod males participated in the study. Subjects were asked to perform running trials on an indoor track at 4m/s in two different conditions: barefoot and shod. A ten-camera system and five mounted force platforms were used to record marker positions and GRF data synchronously at 300 and 1500Hz, respectively. The Oxford foot model was used for kinematic modelling and kinetic variables were calculated using an inverse dynamics approach.

The results showed a few significant differences between the two running conditions, which could have clinical relevance. Barefoot condition showed significantly lower hindfoot-tibia eversion (18.9° ± 6.6°) and forefoot-tibia pronation (25.6° ± 8.0°) range of motion in comparison to shod condition (23.5° ± 5.9°, 30.1° ± 6.3°). These were concomitant with reduction in maximum knee internal rotation in barefoot (15.2° ± 14.5°) condition, in comparison to shod (22° ± 14.8°) condition Moreover, the peak knee abductor moment and knee-power absorption were significantly lower in barefoot (1.84 ± 0.72 $\mathrm{N \cdot m \cdot kg^{-1}}$, 14.1 ± 4.3 W/kg) than in shod (2.19 ± 0.62 $\mathrm{N \cdot m \cdot kg^{-1}}$, 18.1 ± 4.3 W/kg) condition. In previous studies, increased knee internal rotation has been associated to the development and prevalence of iliotibial band syndrome. Furthermore, increased knee abductor moment has been linked to the presence and progression of knee osteoarthritis, and to the development of patellofemoral pain syndrome. On the other hand, the results of multi-segment foot model angles in the current study suggest that, when compared to shod condition, the foot rolls over the phalanges more laterally in barefoot condition. An increased lateral roll-off has been shown to be a risk factor for exercise-related lower leg pain. Furthermore, when compared to shod condition, the ankle power absorption was increased in barefoot condition (11.6 ± 2.8 and 8.6 ± 1.6 W/kg in barefoot and shod condition, respectively). This could, in theory, lead to an increased Achilles tendon load in barefoot running. In conclusion, the results of the current study suggest that barefoot running could possibly be beneficial regarding injuries at the knee level, but detrimental regarding injuries at the ankle level.

Keywords: Barefoot, shod, running, kinematics, kinetics, injury

TIIVISTELMÄ

Syrjälä, J. 2016. Kinemaattiset ja kineettiset erot kengillä juoksun ja paljasjalkajuoksun välillä. Liikuntabiologian laitos, Jyväskylän yliopisto, biomekaniikan pro gradu -tutkielma, 65 s.

Juoksukenkien valmistuksessa käytetty teknologia on kehittynyt merkittävästi. Siitä huolimatta juoksusta aiheutuvien vammojen esiintyvyys on pysynyt korkeana. Juoksuun liittuven vammojen runsas esiintyvyys on johtanut päätelmään, että juoksukenkien kehitys ei ole tehokas keino juoksuun liittyvien vammojen ehkäisyssä. Paljain jaloin juoksemisen (paljasjalkajuoksu) on väitetty mahdollisesti vähentävän riskiä saada juoksemiseen yhdistettyjä vammoja ja parantavan juoksun taloudellisuutta. Nämä väitetyt hyödyt on saanut median ja tiedeyhteisön kiinnittämään huomiota paljasjalkajuoksuun. Vaikkakin paljasjalkajuoksun ja kengillä juoksun välillä on havaittu useita kinemaattisia ja kineettisiä eroja, paljasjalkajuoksun hyötyjä ei ole pystytty vakuuttavasti todistamaan. Toisaalta, modernien juoksukengien, joissa on kantavaimennus ja pronaatiotuki, ei ole myöskään tieteellisesti todettu olevan tehokkaita vammojen ehkäisyssä. Tästä syystä, tämän opinnäytetyön tarkoitus on kaksijakoinen. Ensinnä, vertailla paljasjalkajuoksun ja kengillä juoksun kinemaattisia ja kineettisiä eroja. Toiseksi, kriittisesti arvioida tulosten merkitsevyyttä suhteutettuna paljasjalkajuoksun väitettyihin hyötyihin.

Yhdeksän tavanomaisesti kengillä juoksevaa tervettä mieshenkilöä osallistui tutkimukseen. Koehenkilöitä pyydettiin juoksemaan sisäradalla 4m/s nopeudella kahdella eri tavalla: paljain jaloin (paljasjalkajuoksu) ja kengillä. Koehenkilöihin kiinnitettyjen markkereiden liikeradat ja kontaktivoimat mitattiin samanaikaisesti käyttämällä kymmenen kameran järjestelmää ja viittä radan sisäänupotettua voimalevyä. Mittaustaajuutena käytettiin 300Hz (markkereiden liikeradat) ja 1500Hz (kontaktivoimat). Kinemaattinen mallinnus tehtiin metodilla, joka jakaa nilkan ja jalkaterän useampaan segmenttiin (Oxford foot model). Kineettiset laskutoimitukset tehtiin käänteisdynamiikan avulla.

Tutkimuksessa (opinnäytetyössä) havaittiin tuloksia, jotka ovat mahdollisesti kliinisesti relevantteja. Kantapään (hindfoot) ja tibian välisen eversion liikelaajuus sekä jalkaterän (forefoot) ja Tibian välisen pronaation liikelaajuus oli huomattavasti pienempää paljasjalkajuoksussa (18.9° ± 6.6°, 25.6° ± 8.0°) verrattuna kengillä juoksuun (23.5° ± 5.9°, 30.1° ± 6.3°). Polven (tibian) maksimaalinen sisäkierto oli myös pienempää paljasjalkajuoksussa (15.2° ± 14.5°) verrattuna kengillä juoksuun (22° ± 14.8°). Paljain jaloin juostessa polvi absorpoi vähemmän energiaa (14.1 ± 4.3 W/kg) kuin kengillä juostessa (18.1 ± 4.3 W/kg). Lisäksi polven maksimaalinen abduktori momentti oli pienempi juostessa paljain jaloin (1.84 ± 0.72 $N \cdot m \cdot kg^{-1}$) kuin kengillä juostessa (2.19 ± 0.62 $N \cdot m \cdot kg^{-1}$). Aikaisemmissa tutkimuksissa, polven (tibian) suurentuneen sisäkierron on todettu olevan yhteydessä ITB (iliotibial band) syndrooman kehitykseen ja esiintyvyyteen. Lisäksi, kasvanut polven abduktori momentti on yhdistetty polven nivelrikon esiintyvyyteen ja etenemiseen, sekä patellan kiputilan kehittymiseen. Toisaalta, tämän tutkimuksen tulokset antavat viitteitä siitä, että paljain jaloin juostessa jalka kiertyy varvasluiden yli enemmän lateraalisesti kuin kengillä juostessa. Jalan lateraalisen kiertymisen yli varvasluiden on aiemmissa tutkimuksissa todettu olevan riskitekijä alaraajan (harjoitteluun yhdistettyihin) kiputiloihin. Lisäksi, tässä tutkimuksessa, nilkka absorpoi enemmän energiaa paljain jaloin juostessa (11.6 ± 2.8 W/kg) verrattuna kengillä juoksuun (8.6 ± 1.6 W/kg). Tämä voi teoriassa kasvattaa Akillesjänteeseen kohdistuvaa kuormitusta. Yhteenvetona voidaan todeta, että tämän tutkimuksen tulokset viittaavat siihen, että paljain jaloin juoksu voisi olla hyödyllistä polven kannalta, mutta vahingollista nilkan kannalta.

Asiasanat: paljasjalka, kenkä, juoksu, kinematiikka, kinetiikka, vamma

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ABBREVIATIONS

BFIS	Barefoot-inspired footwear
COP	Centre of pressure
EDL	Extensor digitorum longus
EHL	Extensor hallucis longus
ERLLP	Exercise-related lower leg pain
FDL	Flexor digitorum longus
FF	forefoot
FFS	Forefoot strike
FHL	Flexor hallucis longus
FSA	Footstrike angle
FSP	Footstike pattern
GRF	Ground reaction force
HF	Hindfoot
HX	Hallux
ITBS	Iliotibial band syndrome
MFS	Midfoot strike
OFM	Oxford foot model
PL	Peroneus longus
PIG	Plug-in gait
RFS	Rearfoot strike
STJ	Subtalar joint
TB	Tibia

CONTENTS

1 INTRODUCTION

Running is becoming an increasingly popular activity. More people are running either for fitness or performance (Hryvniak et al. 2014). According to the Running USA organization (2015 state of the sport report), running event finishers in the USA grew by 300% from 1990 to 2013, peaking at over 19 million in 2013. The increase in the running population has created a large market for running equipment, leading to an expansion of footwear technology (Hryvniak et al. 2014). Despite the development of footwear technology, the incidence rate of running injuries remains high. A systematic review of van Gent et al. (2007) reported incidence of lower-extremity running injuries ranging from 19.4% to 79.3%. The high prevalence of running-related injuries has led to a suggestion that the shoe development approach is not working efficiently to reduce injuries (Lieberman 2012). The alleged benefits of barefoot running, such as potential for reduced risk of injury and a more economical approach, has resulted in the scientific community and media to pay significant attention to barefoot running (Tam et al. 2014). The proposed benefits of barefoot running have arose, for the most part, from the evolutionary medical hypothesis. From an evolutionary perspective, humans have been walking and running barefoot for millions of years. Everyone, including athletes, ran barefoot or in minimal shoes before the modern running shoe was invented. Because it was normal for millions of years to run barefoot, the mismatch hypothesis of the evolutionary medicine states that running with big, cushioned shoes is abnormal. (Lieberman 2012.) There are moderately large numbers of published researches that indicate differences in kinematic and kinetic variables between shod and barefoot running. The differences includes: lower stride length and higher stride frequency (De Wit et al. 2000; Squadrone & Gallozzi 2009; Schütte et al. 2013; Bonacci et al. 2014), more plantarflexed foot at initial contact (De Wit et al. 2000; Bishop et al. 2006; Squadrone & Gallozzi 2009; Lieberman et al. 2010; Bonacci et al. 2013; Schütte et al. 2013), reduced peak knee extensor moment (Bonacci et al. 2013; Bonacci et al. 2014, Sinclair 2014), reduced patellofemoral force and pressure (Sinclair 2014), greater peak plantarflexor moment (Bonacci et al 2013; Sinclair 2014) and a greater Achilles tendon force load (Sinclair 2014) in barefoot running compared to shod running. Despite multiple differences found in kinematic and kinetic variables, there is no conclusive evidence proving or refuting the proposed advantages of barefoot running (Tam et al. 2014). On the other hand, a critical review by Richards et al. (2009) states that there is no scientific evidence supporting the prescription of a shoe with an elevated cushioned heel and pronation control system. The purpose of this study is twofold. First, to compare kinematics and kinetics of barefoot and shod running. Second, to critically evaluate the significance of the findings with respect to the proposed advantages of barefoot running.

This thesis is divided into two parts: the literature review (chapters 2-5) and the research part (chapters 6-10). The literature review gives readers the basic knowledge that is needed to follow the research part. Chapter 2 introduces the anatomy of the ankle and foot and describes motions in the ankle and foot joints. Chapter 3 is an overview of biomechanics of running in "normal conditions". The reader is then introduced to some of the differences between barefoot and shod running conditions reported in previous researches (chapter 4). Chapter 5 will briefly describe the key components of a comprehensive running analysis. Chapter 6 begins the research part and explains why the research was made. Chapter 7 explains the experimental protocol and the methodology used. In Chapter 8, the results of the research are reported and these results are discussed with respect to previous published researches in Chapter 9. Finally, Chapter 10 concludes what we learnt from the research and proposes some guidelines for future research.

2 ANATOMY OF THE FOOT AND ANKLE

Running biomechanics are dictated by lower limb anatomy, particularly the joints of the foot and ankle (Dugan & Bhat 2005). Therefore, before having a conversation about the biomechanics of running, it is important to understand the anatomy of the foot and ankle. This chapter focuses on ankle and feet for two reasons: 1) in this thesis, a multi-segment foot model is used, which emphasizes the need for explaining the anatomy of the foot. 2) The motions in the ankle and foot are connected with the motions of the whole lower extremity. This chapter starts with describing the bones and joints of the foot. The motions in different joins of the ankle and foot are explained next. The chapter continues by explaining the roles of arches of the foot and plantar aponeurosis during gait. Finally, the kinematic chain of the lower leg is briefly discussed.

2.1 Bones and joints of the foot and ankle

The skeleton of the foot is consisted of 26 bones (figure 1), which can be divided into the tarsus (ankle), the metatarsus (midfoot) and the digits (toes). The tarsus consists of seven bones: the talus, calcaneus, navicular, cuboid and the three cuneiform bones. (Chan and Rudins 1994; Platzer 2015, 216.)

The *talus* (ankle bone) transmits the weight of the entire body to the foot. The superior surface of the talus forms three joint surfaces (the throclea, lateral malleolar facet and medial malleolar facet), which serve for articulation with the ankle mortise (tibia and fibula). (Platzer 2015, 216.) The joint that incorporates the articulation between the surface of the tibia and fibula with the superior surface of the talus is called the *talocrural joint*, or the true ankle joint (Dugan & Bhat 2005).

The *calcaneus* is the largest tarsal bone and it is the bone where the Achilles tendon is inserted. Anteriorly, the calcaneus bears the articulation surface with the cuboid bone and forms the *calcaneocuboid joint*. On the superior surface of the calcaneus, there are three articular surfaces for the articulation with the talus, namely the anterior, middle and posterior talar articular surfaces. Corresponding articulation surfaces (the anterior, middle and posterior calcaneal facets) can be found from the inferior surface of the talus. These separate articulations function as a single joint: namely the *subtalar joint*. (Dugan & Bhat 2005; Platzer 2015, 216 & 222.)

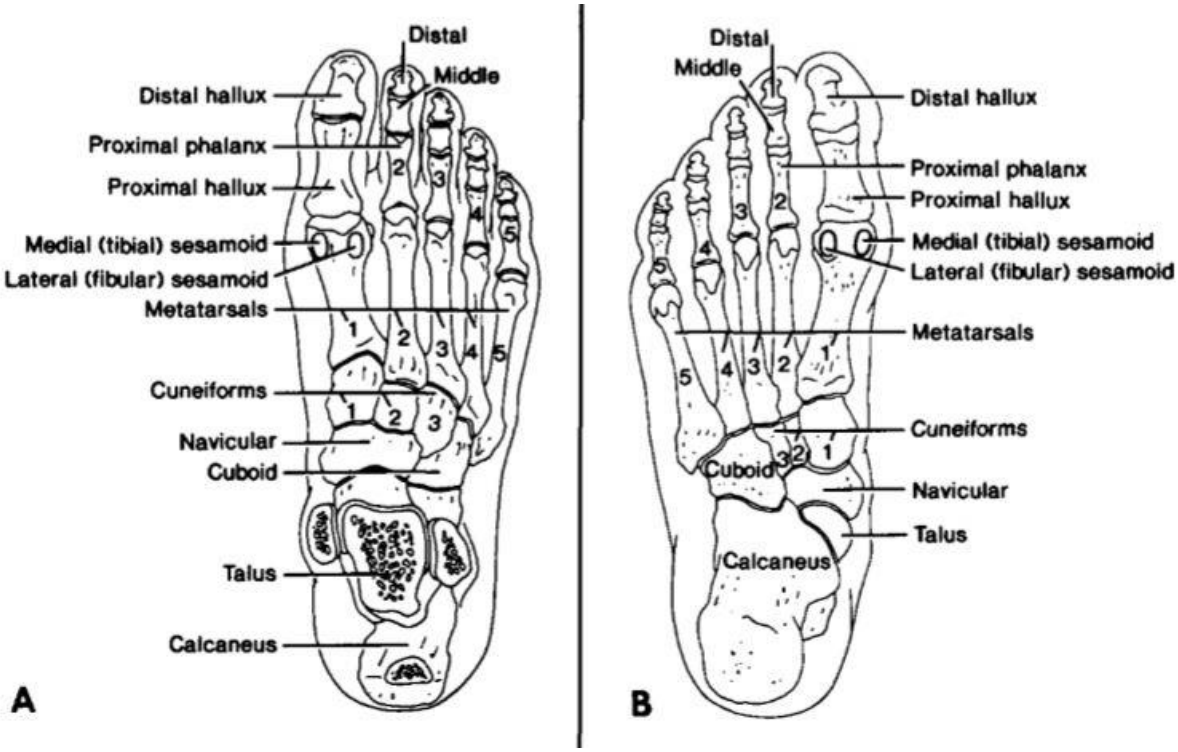

FIGURE 1. Bones of the foot and ankle. Dorsal (A) and plantar (B) view of the right foot. (Modified from: Cull 1989.)

Closely associated with the subtalar joint is the *talocalcaneonavicular joint*, which is made up of the talus, calcaneus, navicular and the plantar calcaneonavicular ligament. In addition to the joint surfaces of the talus, calcaneus and the navicular, the plantar calcaneonavicular ligament connects the sustentaculum tali of the calcaneus and plantar surface of the navicular bone, and forms the articular cavity for the head of the talus. (Chan & Rudins 1994; Platzer 2015, 224.)

The articulation between the talus and navicular is called the *talonavicular joint*. Together with the calcaneocuboid joint, they are collectively called the *transverse tarsal joint*. (Chan & Rudins 1994; Dugan & Bhat 2005.) Posteriorly, the navicular articulates with the talus and, anteriorly, with the three cuneiform bones (medial, intermediate and lateral). The three joint surfaces for the articulation with the three cuneiform bones are separated only by small crests. These three articulations between the navicular and cuneiform bones form the *cuneonavicular* joints. (Platzer 2015, 218.)

Distally, the cuneiform bones articulate with the metatarsals, so that the medial cuneiform articulates with the 1st metatarsal and, to a small extent, with the 2nd metatarsal. The intermediate cuneiform articulates with the 2nd metatarsal and the lateral cuneiform articulates with the 3rd metatarsal, while also having a small facet for the 2nd metatarsal and sometimes also a small facet for the 4th metatarsal. The cuineiform bones also articulate with each other. The *cuboid* articulates distally with the 4th and 5th metatarsals. Medially lies the joint surface for articulation with the lateral cuneiform. (Platzer 2015, 218.) The articulations between the tarsals (cuneiform and cuboid) and the metatarsals are called the *tarsometatarsal joints* (Platzer 2015, 222).

The five *metatarsals* are long, dorsally convex bones that possess a base, a shaft and a head. The 1st metatarsal is the shortest and thickest and it articulates laterally with the base of the 2nd metatarsal. The 2nd, 3rd and 4th are slimmer and on the side of their bases there are joint surfaces for articulation with each other. The 5th metatarsal articulates medially with the base of the 4th metatarsal, and it has a tuberosity on the lateral side of its base. (Platzer 2015, 220.) These articulations between the bases of the metatarsals are called the *intermetatarsal joints* (Platzer 2015, 222).

The 1st *digit* only has two phalanges, while the 2nd – 5th digits each have a proximal, middle and distal phalanx. Each phalanx has a base, a shaft and a head. (Platzer 2015, 220.) The digits articulate with the metatarsal forming the *metatarsophalangeal joints*. *The interphalangeal joints* are the articulations between phalanges. (Platzer 2015, 222.)

2.2 **Movements in the ankle and foot joints**

At first it is necessary to define three cardinal planes (figure 2), of which the description of human movement is conventionally based. These planes are the sagittal plane, the frontal plane and the transverse plane (sometimes called horizontal plane).

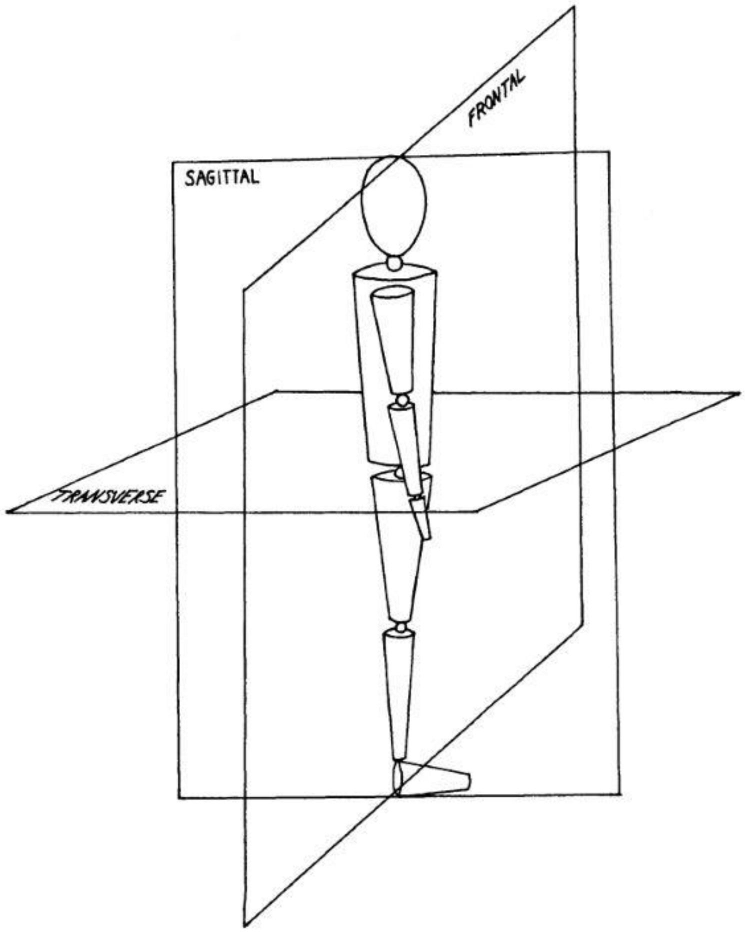

FIGURE 2. The cardinal planes of the body (Modified from: Oatis 1988).

The sagittal plane bisects the body into right and left. Regarding the foot, the terms plantarflexion and dorsiflexion are used to describe the motion in the sagittal plane. Plantarflexion is the movement when the distal aspect of the foot is angled downwards in the sagittal plane, away from the tibia, and dorsiflexion is the movement when the distal aspect is angled towards the tibia, in the sagittal plane. The frontal plane bisects the body into anterior and posterior. Inversion and eversion are the terms for the foot motions that takes place in the frontal plane. In Inversion, the medial border of the foot lifts so that the sole of the foot faces towards the other foot. Eversion is the opposite movement. The transverse plane bisects the body into superior and inferior. Adduction and abduction are the terms used for the foot motions in the transversal plane. Adduction is the movement when the distal aspect of the forefoot is angled towards the midline of the body. Abduction is the movement when the distal aspect is angled away from the midline of the body. (Wang 2012.) It is worth mentioning here that, in general, abduction and adduction refer to motions in frontal plane, but in the foot these motions occur in the transverse plane as described above. Figure 3 clarifies the motions described above.

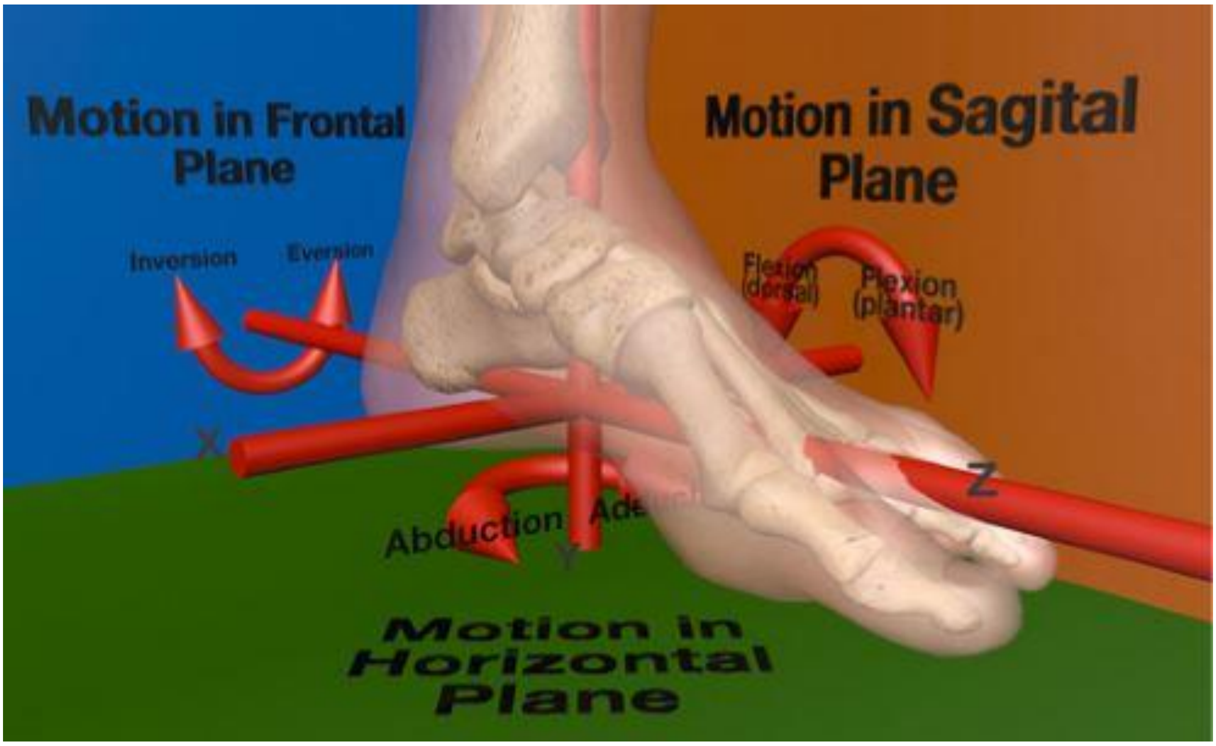

FIGURE 3. Motions of the foot in the three cardinal planes. Note that in this figure the transverse plane is called the horizontal plane. (Modified from: Snedeker et al. 2012.)

The motions of the foot and ankle are defined in terms of the cardinal planes. However, the true mechanical axes of the joints of the foot complex are not perpendicular to these cardinal planes (as will be seen later). Thus the motions of the foot and ankle occur in planes other than the cardinal planes. In reality, motion occurs in planes that pass through all three cardinal planes, and are thus known as triplanar motions. Commonly seen triplanar motions in feet are the pronation and supination (figure 4). The use of the terms pronation and supination is remarkably inconsistent and confusing in literature. For example, these terms are sometimes used as synonyms for the eversion and inversion. (Oatis 1988.) In this thesis, the most widely used terminology is used, as recommended by Oatis (1988). Therefore, the pronation is defined as dorsiflexion, eversion and abduction of the foot and supination as plantarflexion, inversion and adduction and of the foot (Oatis 1988; Chan & Rudins 1994; Dugan & Bhat 2005). The other terms defined above are also used such as described throughout the thesis, unless otherwise mentioned.

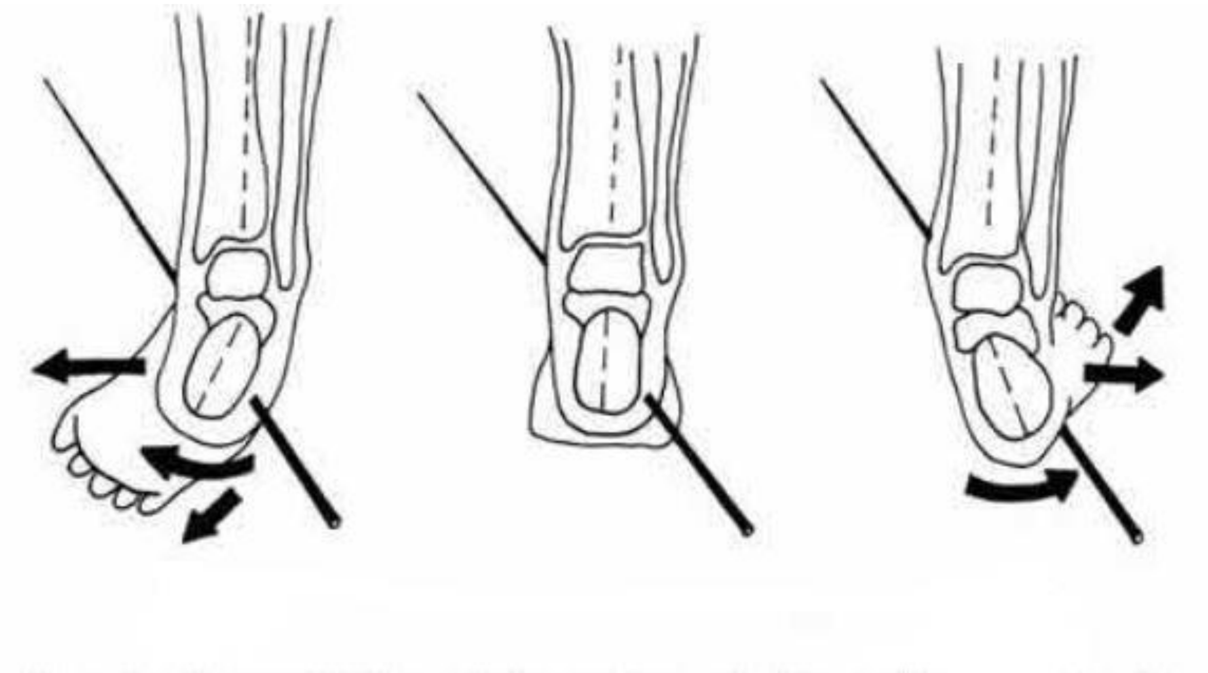

FIGURE 4. The triplanar motions of supination (on the left) and pronation (on the right) (Modified from: McPoil & Knecht 1985).

The axis of the talocrural joint travels from beneath the medial malleoli to the thickest part of the lateral malleoli. Because the lateral malleolus is located posterior to the medial malleolus, the talocrural joint axis travels primarily in the frontal plane with some posterior orientation from the medial to lateral side. Although the talocrural joint is often associated with purely plantarflexion/dorsiflexion motion, the oblique joint axis is a factor that also allows movement in the transverse plane, and to a small extent in the frontal plane. Hence the talocrural joint also contributes to the supination/pronation motion of the foot. (Chan & Rudins 1994; Dugan & Bhat 2005.)

The axis of the subtalar joint (STJ) runs downward, posteriorly in the sagittal plane and laterally in the transverse plane. In the transverse plane, the joint axis is oriented approximately 23° medial to the long axis of the foot and in the sagittal plane the joint axis is oriented, on average, 41° from the horizontal plane (figure 5). However, marked interindividual variation occurs in the orientation angles with a range of 4° to 47° and 21° to 69° in transverse and sagittal planes, respectively. (Chan & Rudins 1994; Dugan & Bhat 2005.)

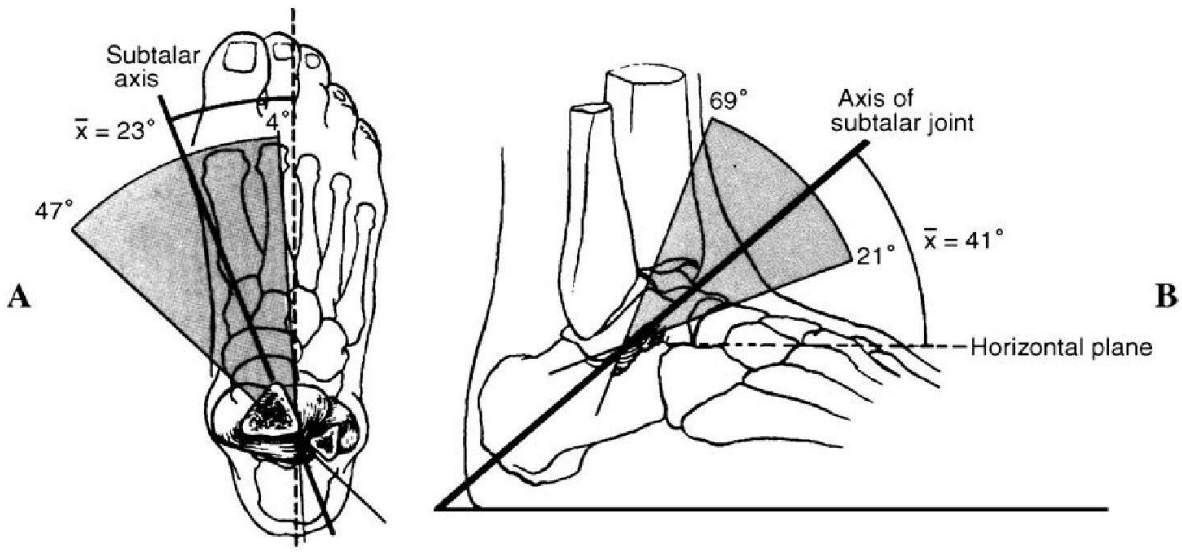

FIGURE 5. Subtalar axes in transverse plane (A) and sagittal plane (B). (Modified from: Goldberg & Hsu 1997, 139.)

The oblique hinge configuration of the subtalar joint allows the complex tripalanar motion of pronation and supination (Chan & Rudins 1994). As the joint axis becomes more horizontal, the joint contributes more to eversion and inversion than to abduction and adduction. Moreover, the closer the axis is to the sagittal plane, the less the joint contributes to plantarflexion and dorsiflexion. (Oatis 1988.) Motions about the ankle and subtalar joint axes are illustrated in figure 6.

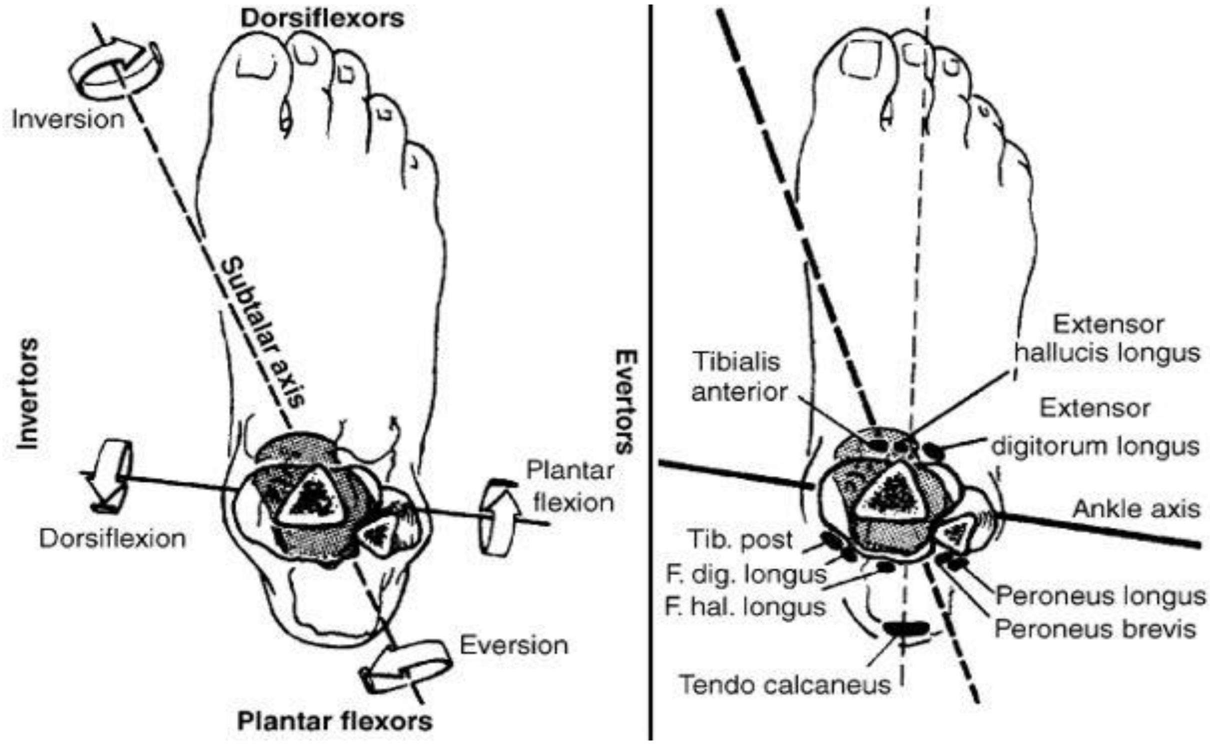

FIGURE 6. Motions about the ankle and subtalar joint axes (on the left). On the right: muscle positions in relation to the ankle and subtalar joint axes. (Modified from: Goldberg & Hsu 1997, 145.)

The transverse tarsal joint is composed of talonavicular and calcaneocuboid joints. Therefore, the transverse tarsal joint has two functional axes that describe the motions in this joint. The longitudinal axis parallels the subtalar joint axis, and the oblique axis is similar to the talocrural joint axis. The subtalar and transverse tarsal joints are closely related. With the subtalar joint pronation (calcaneus everts), the axes of the calcaneocuboid and talonavicular joints become parallel (figure 7). This parallel configuration allows increased motion in the transverse tarsal joint and thus leads to the flexible foot necessary for shock absorption in the stance phase. With supination (calcaneus inverted) the joint axes converge, which causes the transverse tarsal joint to “lock” into a rigid foot configuration necessary for foot propulsion. (Chan & Rudins 1994; Dugan & Bhat 2005.)

The tarsometatarsal (TMT) joints can be divided functionally into five rays. The motion in the first ray (medial cuneiform and 1st metatarsal) is primarily a combination of dorsiflexion and inversion, or plantarflexion and eversion. The second ray (intermediate cuneiform and 2nd metatarsal), third ray (lateral cuneiform and 3rd metatarsal) and the fourth ray (4th metatarsal alone) appear to be involved purely in dorsiflexion and plantarflexion. The fifth ray (5th metatarsal alone) allows pronation/supination motion between the 5th metatarsal and the cuboid. (Oatis 1988.) The second ray has a biomechanically important role in load transmission. The base of the second metatarsal is tightly attached into a socket formed by cuneiform bones and, therefore, forms a relatively immobile connection with the tarsus. This allows increased load to be transmitted through the second metatarsal, when load is transferred into the forefoot for toe-off. (Chan & Rudins 1994.)

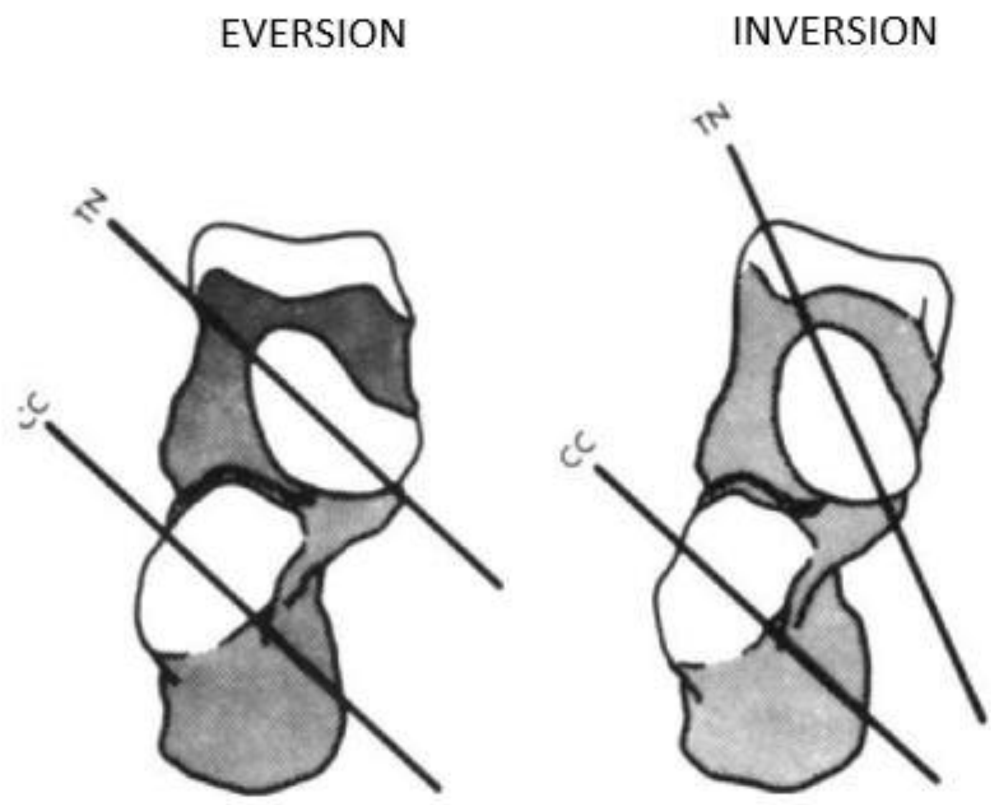

FIGURE 7. Axes of rotaton in talonavicular (TN) and calcaneocuboid (CC) joints, when hindfoot is everted (left) and inverted (right). (Modified from: American academy of orthopaedic Surgeons 1975.)

The metatarsophalangeal joints allow plantarflexion/dorsiflexion and abduction/adduction motions (Oatis 1988). An oblique axis of the metararsophalangeal joints passes from the head of the second metatarsal to that of the fifth metatarsal (figure 8). This axis, around which dorsiflexion of the toes occur, is called the metatarsal break. (Chan & Rudins 1994.)

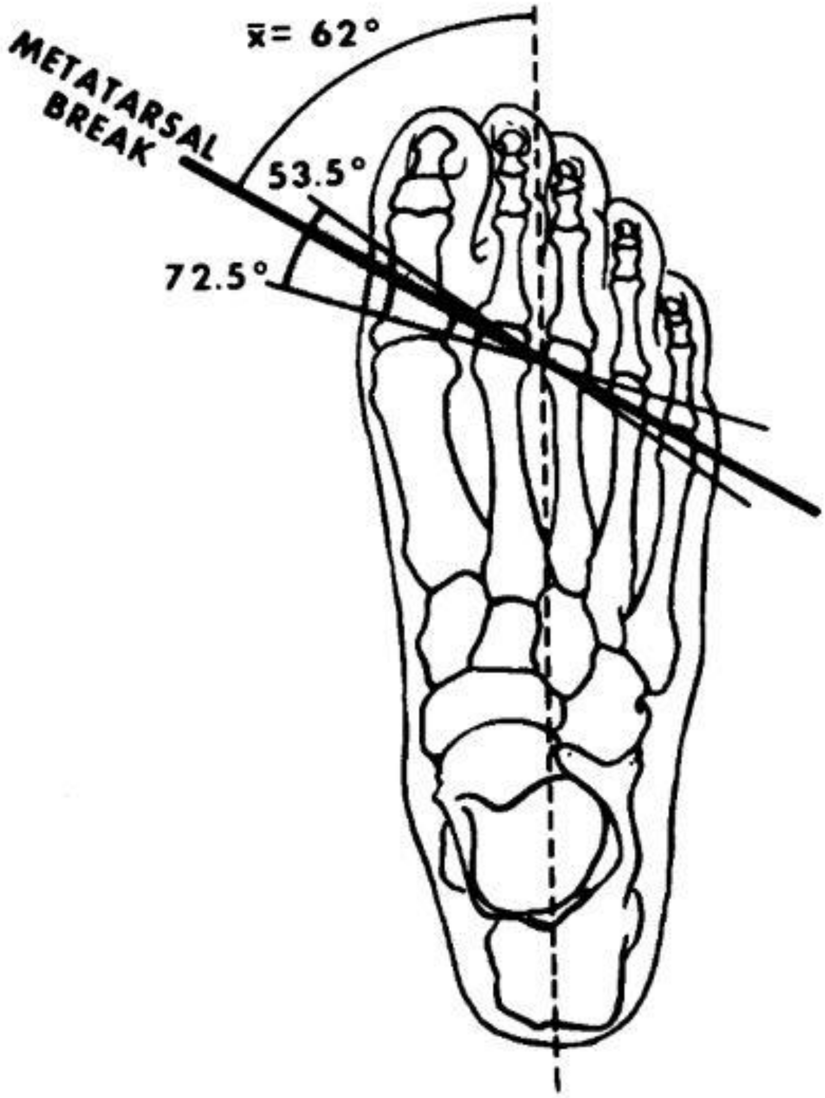

FIGURE 8. The metatarsal break (Modified from: American academy of orthopaedic Surgeons 1975).

2.3 The arches of the foot and plantar aponeurosis

It can be noticed that, posteriorly, the bones of the foot lie over each other, while in the middle and anterior region they lie side by side (Platzer 2015, 226). This organization of the bones, together with ligamentous, and to a lesser extent muscular support, produces the arches of the foot that are known as the longitudinal and transverse arches (Chan & Rudins 1994). The arches of the foot create weight-bearing points on the calcaneal tuberosity, the head of the 1st metatarsal and the head of the 5th metatarsal, thus forming a triangle-shaped supporting surface (Dugan & Bhat 2005; Platzer 2015, 228). The arches allow the foot to be mobile and protect the foot in weight bearing by redistributing pressure. Moreover, the arches permit the rigid foot configuration needed for propulsion. (Chan & Rudins 1994; Dugan & Bhat 2005.)

Of particular interest is the plantar aponeurosis, which joins the calcaneal tuberosity to the plantar surface of the digits. The tension in the transverse fibers of the aponeurosis, in the metatarsal part of the foot, supports the longitudinal and transverse arches of the foot. (Platzer 2015, 228.) The plantar aponeurosis is distally inserted to the base of the proximal phalanges, crossing the transverse tarsal and metatarsophalangeal joints. As extension occurs at the metatarsophalangeal joints, the plantar aponeurosis tightens and pulls the calcaneus and metatarsal heads towards each other. This is called the windlass mechanism (figure 9). As a result of this, the distance between the calcaneus and metatarsal heads is shortened, the height of the arch is increased and the tarsal bones are forced into locked flexion position. The tension on the plantar aponeurosis, together with muscle activity of the intrinsic muscles, increases the stability of the foot and helps to create a rigid foot for push off. (Chan & Rudins 1994; Dugan & Bhat 2005.)

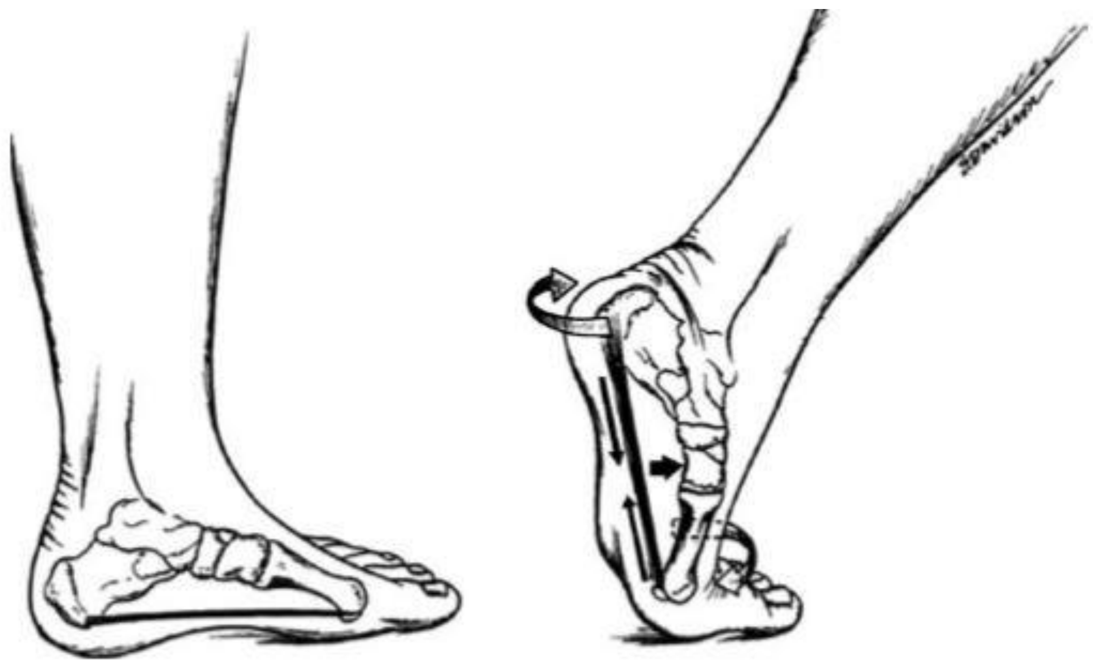

FIGURE 9. The windlass mechanism. After heel-off, metatarsophalangeal extension increases tension on the plantar fascia (converging arrows). The transverse tarsal joint is forced into flexion (arrowhead). (Geiringer 1997.)

2.4 **Kinematic chain**

In the stance phase of walking and running, when the foot is fixed (closed kinematic chain), the motion of the foot and ankle is translated proximally to the tibia, fibula and femur. Vice versa, the motion of the lower leg is translated distally to the ankle and foot. An internal rotation of the tibia causes evertion of the hindfoot when the calcaneus turns away from the midline. This motion in the subtalar joint is passed forward to the midfoot and forefoot. As a result of the eversion in the subtalar joint, the two joint axes in the transverse tarsal joint become parallel, which allows pronation and increased motion in the transverse tarsal joint. On the other hand, an external rotation of tibia causes invertion of the hindfoot when the calcaneus is brought towards the midline. With the hindfoot inversion, the joint axes converge, which results in the rigid foot configuration. Figure 10 is a simplified schema of the above described kinematic chain. (Chan & Rudins 1994; Dugan & Bhat 2005.)

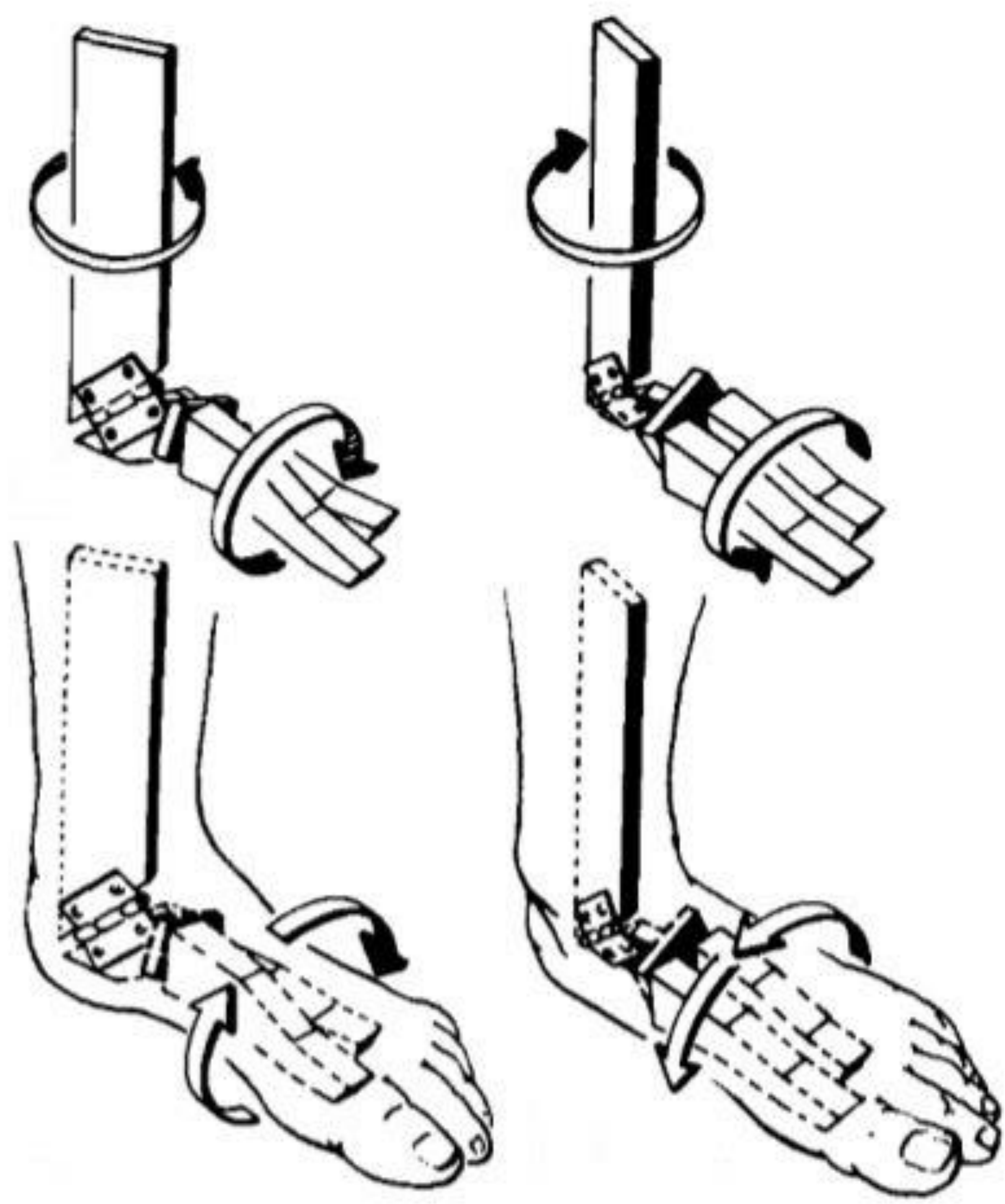

FIGURE 10. Simplified schema of the interaction between tibia, ankle and foot. On the left: outward rotation of upper stick (tibia) results in inward rotation of lower stick (calcaneus inversion), and subsequent elevation of the medial border of foot and depression of lateral border of the foot. On the right: inward rotation of upper stick (tibia) results in outward rotation of lower stick (calcaneus eversion), and subsequent depression of medial border of foot and elevation of lateral border of foot. (Modified from: American academy of orthopaedic surgeons 1975.)

3 BIOMECHANICS OF RUNNING

This chapter introduces first the basic unit of measurement in gait analysis, that is, gait cycle. Once the terminology of the gait cycle has been introduced, the chapter continues with describing the kinematics and kinetics of running in the stance phase and swing phase, respectively. There are some differences in gait cycle parameters when comparing the gait of walking, running and sprinting. For example, the initial contact during walking and running occurs with the heel. In sprinting, the initial contact occurs on the forefoot. In walking it is necessary to dorsiflex the ankle at the time of toe-off to achieve toe clearance. In contrast, this is not necessary in sprinting and running. While the pattern of motions of the knee is similar in walking, running and sprinting, the extremes of motions are very different. (Lacquaniti et al. 2012; Novacheck 1998.) The mentioned differences are only some examples of differences in gait between walking, running and sprinting. As this research centres on running, this chapter will focus on the kinematics and kinetics of running. Furthermore, the kinematics and kinetics in this chapter focus mainly on the sagittal plane, as the motions in the frontal and transversal plane are subtle compared to sagittal plane. However, the triplanar supination/pronation motions of the ankle/foot and hip motion in the frontal plane are relevant motions and will be addressed.

3.1 Gait cycle

The gait cycle starts when one foot contacts the ground and ends when the same foot contacts the ground again. The gait cycle can be divided into phases. The time point when the foot first makes contact with the ground is referred to as initial contact, which is the beginning of the stance phase. The stance phase stands for the whole period that the foot is in contact with the ground, that is, from initial contact to the point that the foot is no longer in contact with the ground. The point when the foot loses contact with the ground is called toe-off, and it marks the beginning of the swing phase. (Novacheck 1998.) In walking gait cycle, there are two periods of double support (both feet on the ground). On the contrary, there are no periods of double support in running gait cycle. In fact, both feet are airborne twice during the running gait cycle, which is referred to as the double-float phase. (Ounpuu 1990.)

During the gait cycle, alternate periods of acceleration and deceleration occur, referred to as absorption and generation periods. During the period of absorption, the body´s centre of mass falls from its peak height during the double-float phase, and the horizontal velocity of the centre of mass decelerates. The absorption period is divided by initial contact into swing-phase absorption and stance-phase absorption. Stance-phase reversal marks the point when the absorption phase ends and the generation period begins. During stance-phase generation the centre of mass is propelled upward and forward. The kinetic and potential energy increases and the limb is then propelled into a swing phase after toe-off (swing-phase generation). The next period of absorption begins at the point called swing-phase reversal. (Novacheck 1998.) Figure 11 demonstrates the gait cycle events mentioned above.

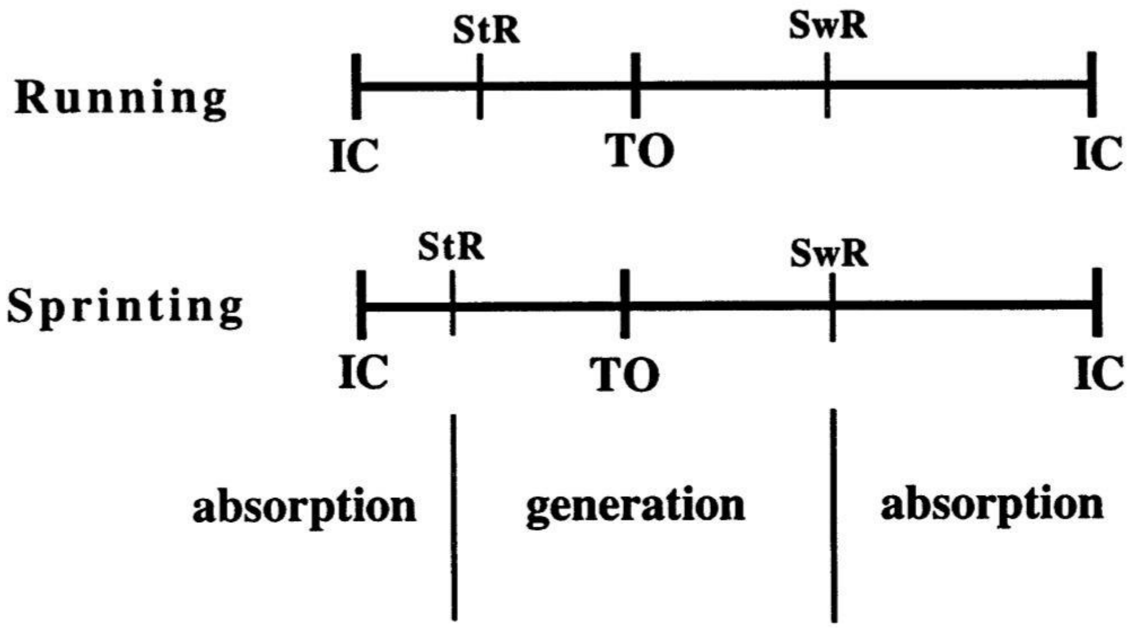

FIGURE 11. Gait cycle events. IC, initial contact; TO, toe-off; StR, Stance-phase reversal; SwR, swing-phase reversal. (Modified from: Novacheck 1998.)

Cadence, stride length and step length are terms that describe the speed and length of gait. Step length is the distance from initial contact of one foot to initial contact of the opposite foot. Stride length is the distance between successive initial contacts of the same foot. Cadence denotes the number of steps per unit time (usually steps/min). Temporal and spatial variables cannot be separated as they are generally interrelated. Velocity increase can be achieved by increasing step length, by increasing candence or by increasing both. (Ounpuu 1994.)

3.2 Kinematics and kinetics of stance phase

The stance phase starts with the initial contact. The first part of the stance phase is the absorption phase. Energy absorption is a key function of the lower extremity during the absorption stance phase. Joint motion, eccentric muscle contraction and articular cartilage compression are the factors that allow impact absorption. (Dugan & Bhat 2005.) Figure 12 is provided to help readers to assimilate the different phases of gait in terms of joint angles, moments and powers in sagittal plane.

Initial contact during running occurs with the heel. Eccentric contraction of the tibialis anterior muscles controls the lowering of the forefoot to the ground during the absorption phase. (Novacheck 1998.) Before the initial contact, hamstrings are dominant muscles and produce knee flexor moment to prepare the lower extremity for the impact, and control rapid knee extension. Following the initial contact, as the knee flexes, the quadriceps contract eccentrically, producing a knee extensor moment. Eccentric contraction of the quadriceps controls the height of the body's centre of gravity, resists excess knee flexion and works as a shock absorber. (Ounpuu 1994; Novacheck 1998.) The gastrocnemius-soleus complex contracts eccentrically to control forward movement of the tibia and to provide stability to the ankle (Dugan & Bhat 2005). The hamstrings, which act as hip extensors, are active through the stance phase as the body progresses forward on the fixed limb. Shock absorption also occurs at the hip. In the absorption period of the stance phase the hip adducts relative to the pelvis (which is relatively stationary), because the ground reaction force falls medial to the hip. Hip abductors (primarily gluteus medius) contribute to shock absorption by contracting eccentrically, therefore resisting the adductor moment caused by ground reaction force. (Ounpuu 1994; Novacheck 1998.) As forward progression continues through the middle of the stance phase, dorsiflexion reaches a maximum. Just before maximum dorsiflexion, the maximum pronation occurs. The point of maximum pronation marks the end of the stance phase's absorptive period and the beginning of the stance phase's generation period. (Dugan & Bhat 2005.)

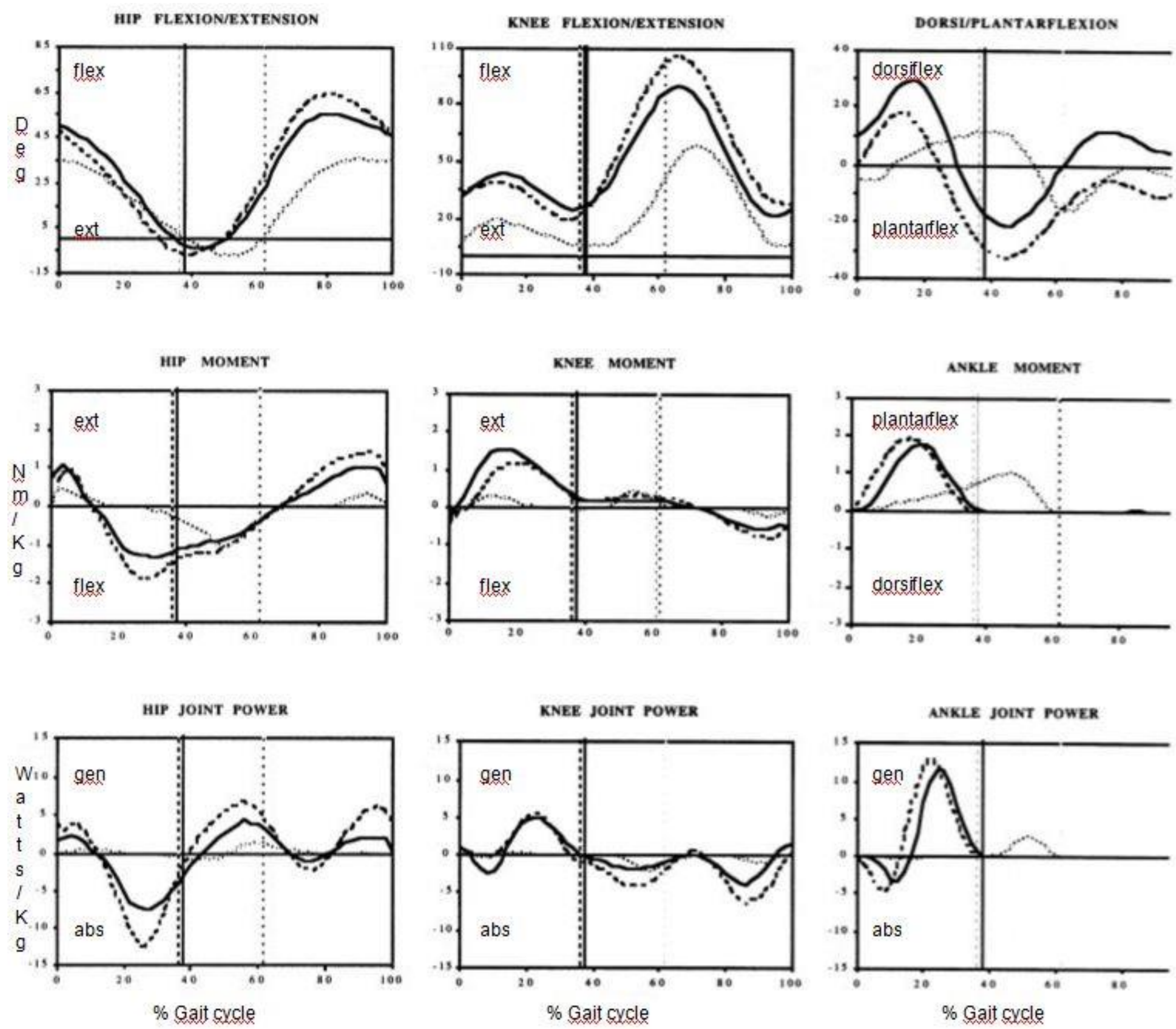

FIGURE 12. Joint angles (top), moments (middle) and powers (bottom) plotted against gait cycle. Solid line: running condition. (Modified from: Novacheck 1998.)

Ankle plantarflexion has an important role in the generation period of a stance phase. The concentric contraction of gastrocnemius-soleus complex produces plantarflexion and initiates the acceleration of the stance limb. During this point of the stance phase, the maximum ground reaction force occurs as the foot pushes off the ground and propels the body forward. Rigid foot configuration is essential for the adequate force needed to push away from the ground. Several factors allow the rigid foot configuration to occur. These factors include: STJ supination, the metatarsal break phenomenom, Plantar aponeurosis and contraction of intrinsic foot muscles. (Dugan & Bhat 2005.) In addition to ankle plantarflexion, hip and knee extension is also needed to add to the thrust of the body as it progresses into the swing phase. Hamstrings act as an active extensor of the hip and generate power through the first half of the stance phase. Following this, hip flexors become dominant, changing the hip moment to flexion moment, therefore decelerating the backward rotating thigh. Just before toe-off, rectus femoris begins to contract concentrically to maximize knee extension and starts to drive the thigh forward into swing. In addition, the gluteus medius contracts concentrically, abducting the hip and generating power. At the termination of stance phase, the tibialis anterior begins to contract, while gastrocnemius-soleus stops contracting. (Dugan & Bhat 2005; Novacheck 1998; Winter 1983.)

3.3 Kinematics and kinetics of swing phase

After the toe-off swing phase begins, the limb is swinging forward due to hip flexion caused by the concentric contraction of rectus femoris and iliopsoas. Rectus femoris also contracts eccentrically to resist knee flexion, which occurs because the line of ground reaction force (at toe-off) passes posterior to the knee joint. Tibialis antrerior is activated, therefore causing dorsiflexion of the ankle and ensuring the foot clearance. However, this is more important in walking: given the amount of knee flexion while running, foot clearance is possible without the dorsiflexion while running. After the double float phase, the opposite foot hits the ground and hip abductors are activated to stabilize the pelvis. (Ounpuu 1994; Novacheck 1998.)

Peak hip flexion occurs in the second half of the swing phase. After the peak hip flexion the hip extensors contract to extend the hip in preparation for the next initial contact. Hamstrings contract eccentrically during late swing to control momentum of the tibia and to slow down the rapidly extending knee. Gastrocnemius-soleus also starts to contract before initial contact to prepare the foot for impact. Cocontraction of tibialis anterior at the initial contact helps to create a stable foot for weight acceptance. (Ounpuu 1994; Novacheck 1998.)

4 BAREFOOT VS SHOD RUNNING

This chapter introduces some of the previously reported biomechanical differences between barefoot and shod running. The chapter starts with differences in spatiotemporal variables and continues with differences in kinematic and kinetic variables. Finally, the chapter ends with explaining how footstrike pattern (FSP) is modified during barefoot running in comparison to shod running. This chapter is not a complete literature review about the subject, but more of an introduction to the most frequent findings of the literature.

4.1 Spatiotemporal variables

There are some frequently found differences in spatiotemporal variables between barefoot running and shod running. McCallion et al. (2014) reported that stride duration was significantly greater and contact time was longer when running shod compared with running barefoot. Furthermore, running barefoot resulted in higher stride frequencies when compared with shod and minimalist shoe conditions. Bonacci et al. (2013) reported similar findings regarding stride length and stride frequency. Stride length was shorter and stride frequency was higher when running barefoot compared with other conditions (minimalist shoes and normal running shoes). Similar results have been reported in other research articles, regardless of running velocity or whether the subjects where experienced barefoot runners or not (De Wit et al. 2000; Squadrone & Gallozzi 2009; Schütte et al. 2013; Bonacci et al. 2014).

4.2 Kinematic variables

The most commonly stated difference in kinematic variables between barefoot running and shod running is more plantarflexed ankle position at footstrike while running barefoot compared to shod running. This has been reported in several research articles. (De Wit et al. 2000; Bishop et al. 2006; Squadrone & Gallozzi 2009; Bonacci et al. 2013; Schütte et al. 2013.)

In the Squadrone and Gallozzi (2009) study, subjects landed with significantly more dorsiflexed ankle when running with standard running shoes than compared to when running barefoot. However, they did not find significant differences at the knee joint. Bonacci et al. (2013) similarly reported that when running barefoot the ankle joint was less dorsiflexed at initial contact when compared with all shod conditions (minimalist, racing flat, regular shoe). During the stance phase, peak ankle dorsiflexion and adduction were reduced in barefoot and minimalist shoe conditions compared to racing flat and regular shoe conditions. Furthermore, the ankle was more plantarflexed at the toe-off in barefoot condition. Bonacci et al. (2013) did not report a difference in knee flexion angle at initial contact. However, they reported that peak knee flexion during midstance was decreased in barefoot condition compared with all shod conditions. The study by De Wit et al. (2000) also showed that in barefoot running, placement of the foot is significantly more horizontal than in the shod condition. Moreover, the initial eversion at initial contact was significantly smaller in barefoot condition.

4.3 Kinetic variables

De Wit et al. (2000) reported a significantly greater loading rate (maximal vertical loading rate between touchdown and occurrence of the first vertical impact force) during impact in barefoot running compared to shod running. They also reported that, in general, more than one impact peak was found for the barefoot condition. While De Wit et al. (2000) reported no differences in magnitude of impact peak force, Squadrone and Gallozzi (2009) reported a significantly higher magnitude of impact peak forces when running in standard running shoes compared to barefoot. Furthermore, they reported reduced peak pressures under the heel and higher pressures underneath the metatarsal heads when running barefoot compared to running with standard running shoes.

Sinclair (2014) studied effects of barefoot and barefoot-inspired footwear (BFIS) on knee and ankle loading during running. The aim was to determine whether running barefoot and BFIS caused different levels of patellofemoral force and patellofemoral pressure at the knee and Achilles tendon force at the ankle when compared to using conventional running shoes. The experimental footwear consisted of three different barefoot-inspired footwear (Inov-8 Evoskin, Nike Free 3.0 and Vibram Five Fingers), and one conventional footwear (Saucony Pro Grid Guide 2). The results showed that knee extensor moment, patellofemoral force, patellofemoral pressure and patellofemoral force load rate were all significantly greater in the conventional footwear in comparison to barefoot. Furthermore, referring to the ankle, Sinclair (2014) reported that peak plantarflexor moment, Achilles tendon force and Achilles tendon force load were all greater in barefoot compared to conventional footwear. There were also similar differences in BFIS compared to conventional footwear, but the findings were inconsistent. Sinclair (2014) concluded that running barefoot and in BFIS exhibit significant reductions in knee patellafemoral force and patellofemoral pressure. On the other hand, results showed increased forces at the Achilles tendon. Findings of the studies conducted by Bonacci et al. (2013 & 2014) are consistent with the findings of Sinclair (2014). The results showed reductions in peak knee extension moment, peak knee abduction moment, peak power generation at the knee, negative work at the knee and patellofemoral joint reaction force during barefoot running when compared to shod running. Furthermore, Bonacci et al. (2013) also found increases in peak ankle plantarflexor moment and positive work at the ankle when running barefoot compared to all shod conditions.

4.4 Footstrike patterns

During running, initial contact predominantly occurs with the heel. However, this is not always true. It has been shown that the selection of shoe (or running barefoot) and individual training backgrounds affects whether the initial contact occurs with heel, midfoot or forefoot. (Novacheck 1998.) A collision of the foot with the ground (initial contact) can occur in three ways, depending on which part of the foot makes contact with the ground first. A rearfoot strike (RFS) is the condition in which the heel lands first; a midfoot strike (MFS) means that the heel and ball of the foot land simultaneously, and forefoot strike (FFS) is a condition in which the ball of the foot lands before the heel. (Lieberman et al. 2010.) Approximately 75-80% of contemporary shod endurance runners use RFS pattern, whereas remaining runners use either FFS or MFS pattern (Hasewaga et al. 2007). Barefoot running has been associated with switching from RFS to MFS/FFS patterns (Thompson et al. 2015).

An interesting study regarding footstrike kinematics was conducted by Lieberman et al. (2010). They compared footstrike kinematics among three groups of adults who run a minimum of 20km per week: (1) habitually shod athletes from the USA; (2) athletes from the Rift Valley Province of Kenya, most of whom grew up barefoot but now wear cushioned shoes when running; and (3) US runners who grew up shod but now habitually run barefoot or in minimal footwear. Data from the study shows that habitually shod runners that grew up wearing shoes (group 1) mostly RFS when shod. These runners also predominantly RFS when running barefoot on the same hard surfaces, but adopt flatter foot placements by dorsiflexing approximately 7-10% less. Runners who grew up barefoot (group 2), most often used FFS landings in both barefoot and shod conditions. Habitually barefoot runners (group 3) predominantly RFS when running shod, but the occurrence of FFS increased markedly in barefoot compared to shod condition (37% in shod, 75% in barefoot). Lieberman et al. (2010) also reported markedly different collision forces with the ground during FFS and RFS running. Magnitudes of peak vertical force during the impact period were approximately three times lower in habitually barefoot runners who FFS than in habitually shod runners who RFS regardless of whether they ran barefoot or shod. Although barefoot running is often associated with switching to FFS pattern, it has to be remembered that barefoot is not the same as FFS. For example, in the study of Strauts et al. (2016), all of the subjects continued to adopt RFS in barefoot condition.

5 METHODOLOGY FOR RUNNING ANALYSIS

In human gait analysis we are interested in the motion itself (kinematics) and in the forces that causes the motion (kinetics). Particularly interesting aspects are the joint forces and moments. The estimation of the joint forces and moments can be achieved by using a process called inverse dynamics, which is based on the kinematic data, anthropometric parameters and force plate data (Ren et al. 2008). This chapter will briefly describe the key components (force plates, motion analysis and inverse dynamics) of a comprehensive running analysis.

5.1 Force plates

During standing, walking and running, the ground reaction force (GRF) acts on the foot. The GRF is a three-dimensional vector that has a vertical component and two shear-force components. The shear forces are generally resolved into anterior-posterior and medial-lateral directions. (Winter 2005, 96.).

The force plate is a device that can be used to measure GRF. Applied force to the force plate is measured via force transducers, which are devices that give an electrical signal proportional to the applied force. There are different types of force transducers but they all work on the same principle: the applied force causes a certain amount of strain, which unbalances the electrical characteristic of the transducer and can be translated to a signal proportional to the applied force. (Winter 2005, 95 – 96.)

Two common types of force plates are used. The first type is a flat plate that is supported by four (one in each corner) triaxial force transducers (figure 13). In this case, the total vertical force is the sum of all four force transducers, and the location of the center of pressure can be calculated by the relative vertical forces detected at each of the corner transducers. (Winter 2005, 96.)

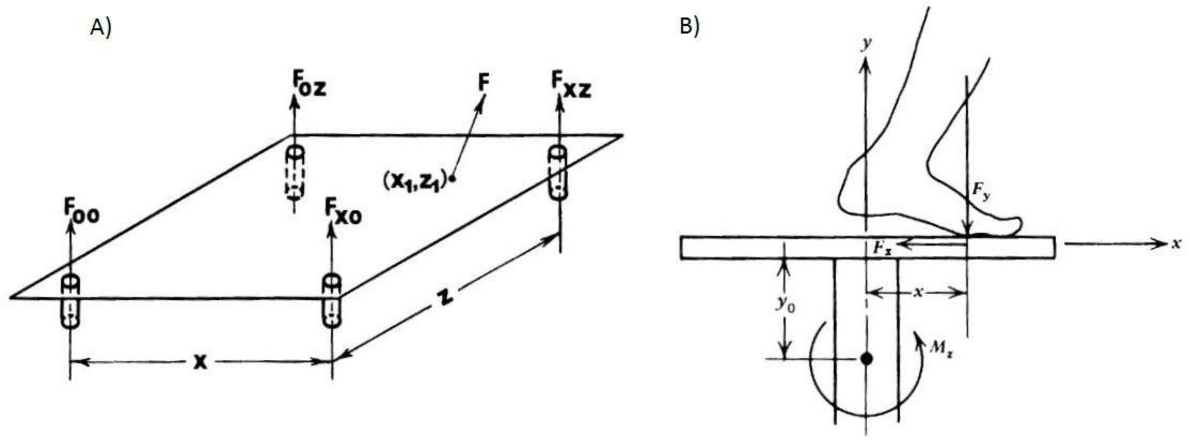

FIGURE 13. Two common types of force plates. A) Force plate with the force transducers in the corners. B) Centrally supported force plate with the forces and moments involved. (Modified from: Winter 2005, 97 – 98.)

A second type of force plate is a flat plate supported by a centrally instrumented pillar (figure 10). By adding the moments acting on the force plate, we get the following equations:

$$M_z - F_y \cdot x + F_x \cdot y_0 = 0 \quad , \quad x = \frac{F_x \cdot y_0 + M_z}{F_y}$$

Where M_z is a bending moment regarding the axis of rotation of support, and y_0 is the distance from support axis to force plate surface. Hence the location of the centre of pressure x can be calculated. (Winter 2005, 97 – 98.)

5.2 Motion analysis

The simplest form of motion analysis is the use of a camera for still photography, or a video recording device for filming to get a quantitative description of body segments in gait. To get a three-dimensional analysis of motion, more sophisticated systems have been developed. These systems use multiple cameras/detectors and active (infrared transmitters) or passive (retroreflective) markers, which are placed on body segments and imaged sequentially through a calibrated field of view. Once a full description of the motion of body segments is available, measures, such as joint angles and velocities, can be calculated. These measures of force allow for the calculation of joint moments, powers and mechanical energy. (Dugan & Bhat 2005.) The motion analysis methods used in this thesis are explained in more depth in chapter 7.

5.3 Link-segment model and inverse dynamics

Knowledge of force patterns is of particular importance for understanding the cause of movement. A direct measurement of the forces exerted by a muscle is not applicape for humans. Therefore, we need to attempt to calculate these forces indirectly by using a process called link-segment model. With a full kinematic description, accurate anthropometric data and external forces we can calculate the joint reaction forces and muscle moments (figure 14). This calculation process is called inverse solution/inverse dynamics. (Winter 2005, 86.) Inverse dynamics is an iterative solution of the body segment equation of motion, starting with measured ground reaction forces. The process calculates joint forces and moments at each successive segment, beginning with the segments that are in contact with the ground. (Ren et al. 2008.)

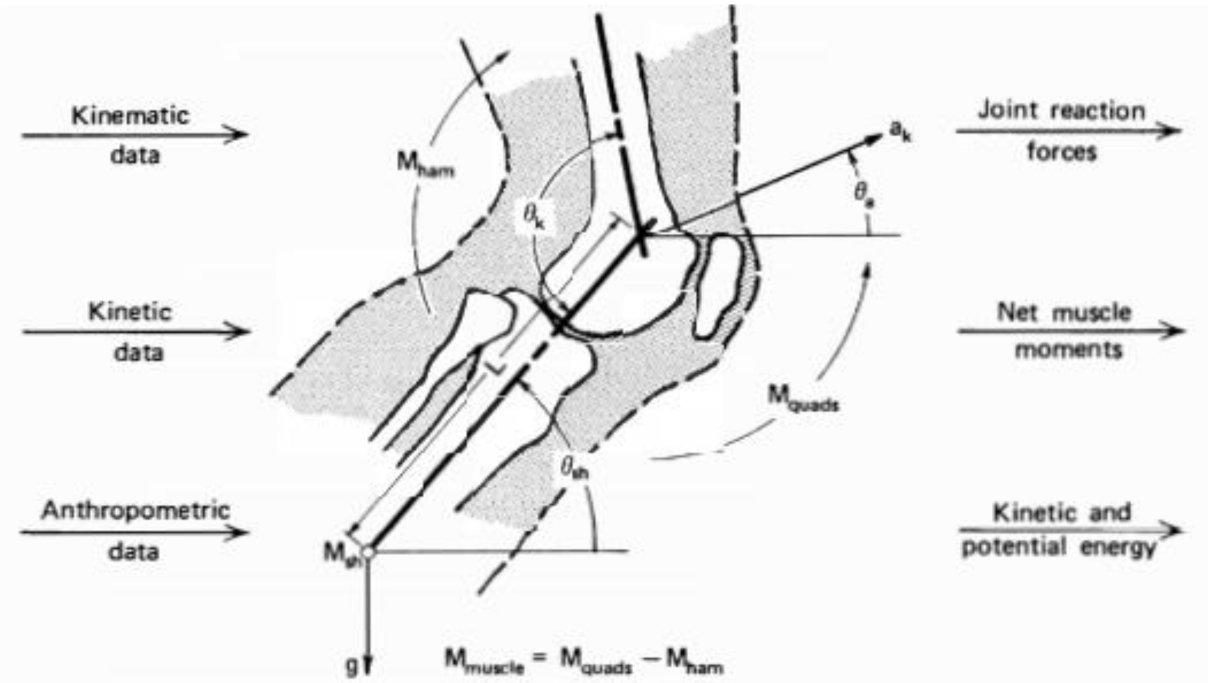

FIGURE 14. Inverse dynamics. Kinematic, kinetic and anthropometric data is used to calculate joint reaction forces, muscle moments and mechanical energy. (Winter 2005, 87.)

For the link-segment model development, accurate measures of segment masses, centre of mass, joint centre and moments of inertia are required. These anthropometric variables can be obtained from statistical tables or to some extent measured directly, but with limited accuracy. (Winter 2005, 86 – 87.). Winter (2005, 87) lists five assumptions that have to be made with respect to the link-segment model: 1) The mass of the segment is fixed and is a point mass located at the segment's centre of mass. 2) The location of the segment's center of mass remains fixed during the movement. 3) The joints are considered to be hinge (or ball and socket) joints. 4) The mass moment of inertia of the segment about its centre of mass is constant during the movement. 5) The length of each segment remains constant during the movement.

There are three forces acting on the link-segment model: gravitational forces, ground reaction or external forces and muscle/ligament forces. External forces are distributed over an area of the body, for example ground reaction forces under the area of the foot. In order to get a vector presentation of such forces, they must be considered to act at a point called the centre of pressure (COP). Muscle activity at a joint can be calculated in terms of net muscle moments. This means that calculated value represents only the net effect of agonist and antagonist muscles when co-contraction occurs. Friction effects at the joints or within the muscle, which reduces the effective muscle moment, cannot be separated from the net muscle moment value. Furthermore, it has to be noted that it is impossible to determine the contribution of the moments generated by the passive structures, such as ligaments. (Winter 2005, 88.)

6 PURPOSE OF THE STUDY

According to Nigg (2009) there are four open questions regarding the topic of barefoot running: 1) What are the biomechanical differences between shod and barefoot running? 2) What are the training effects of barefoot running? 3) What are the performance/economy advantages of barefoot running? 4) Are there association between barefoot/shod running and injuries? The biomechanical justification for barefoot running centers on the concept that the shod-to-barefoot change would result in biomechanical differences that are injury preventative (Tam et al. 2014). Advocates of barefoot running rely on the evolutionary perspective and mismatch hypothesis, which states that barefoot running is the most natural way of running and running wearing cushioned shoes is abnormal (Lieberman 2012). Furthermore, the supporters of barefoot running appeal to the lack of scientific evidence supporting the prescription of a shoe with an elevated and cushioned heel and pronation control system (Richards et al. 2009), and the high prevalence of running-related injuries (van Gent et al. 2007). On the other hand, there is no conclusive evidence supporting barefoot running as an injury preventative mode of running (Tam et al. 2014). Therefore, the purpose of this study is twofold:

1) To examine the biomechanical differences between shod and barefoot running.
2) To evaluate the implications of biomechanical differences between shod and barefoot running regarding running injuries.

The 3D- motion analysis and inverse dynamics approach was used to evaluate the kinematic and kinetic differences between shod and barefoot conditions. The multi-segment foot model was used to gain insight of the motions between the segments of the foot. To the best knowledge of the author, there is no published research evaluating biomechanical differences between shod and barefoot running, using a multi-segment foot model.

7 METHODS

7.1 Subjects

A total of nine healthy, habitually shod males, with a mean mass of 73 kg (± 6,4 kg) and height of 1,76 m (± 0,08m) underwent a 3-D running analysis. At the time of the study, the subjects were injury and symptom-free. Subjects had no previous history of any musculoskeletal problems, such as a recent injury or surgery, that could have had an effect on their running patterns. The subjects participated voluntarily and provided informed consent. The study was conducted in accordance to the Helsinki declaration.

7.2 Study protocol

Subjects were asked to perform running trials at two different conditions: barefoot and shod. Trials were performed on a 30-m indoor track at 4m/s (± 10%). The experimental setup is shown in figure 15. Two photocells were used to control the velocity. A ten-camera system (Vicon T40, Vicon) and five mounted force platforms (AMTI BP6001200; AMTI, Watertown, MA) were used to record marker positions and GRF data synchronously at 300 and 1500Hz, respectively. The force platforms were mounted in a mid-part of the runway (total length 5,7m). In each condition subjects were asked to perform as many trials as necessary to get at least five successful contacts (right foot) on a force platform. Subjects were encouraged to run with their preferred running style, regardless of running condition. After each trial feedback of the running velocity was given; if the running velocity was too high the subject was asked to lower the speed for the next trial and vice versa. No information about the objectives of the study was given in order to minimise intentional adjustments in running style.

Anthropometric measurements (height, weight, leg length) were carried out before running trials. Retroreflective markers were placed on 21 different anatomical landmarks (table 1 and figure 16) according to the Oxford foot model (Vicon, Oxford, UK). In shod condition, a modified pair of new neutral running shoes (Nike Pegasus 30) was used (figure 16). Holes were cut in the shoes to allow precisely the same location of the foot markers in both conditions.

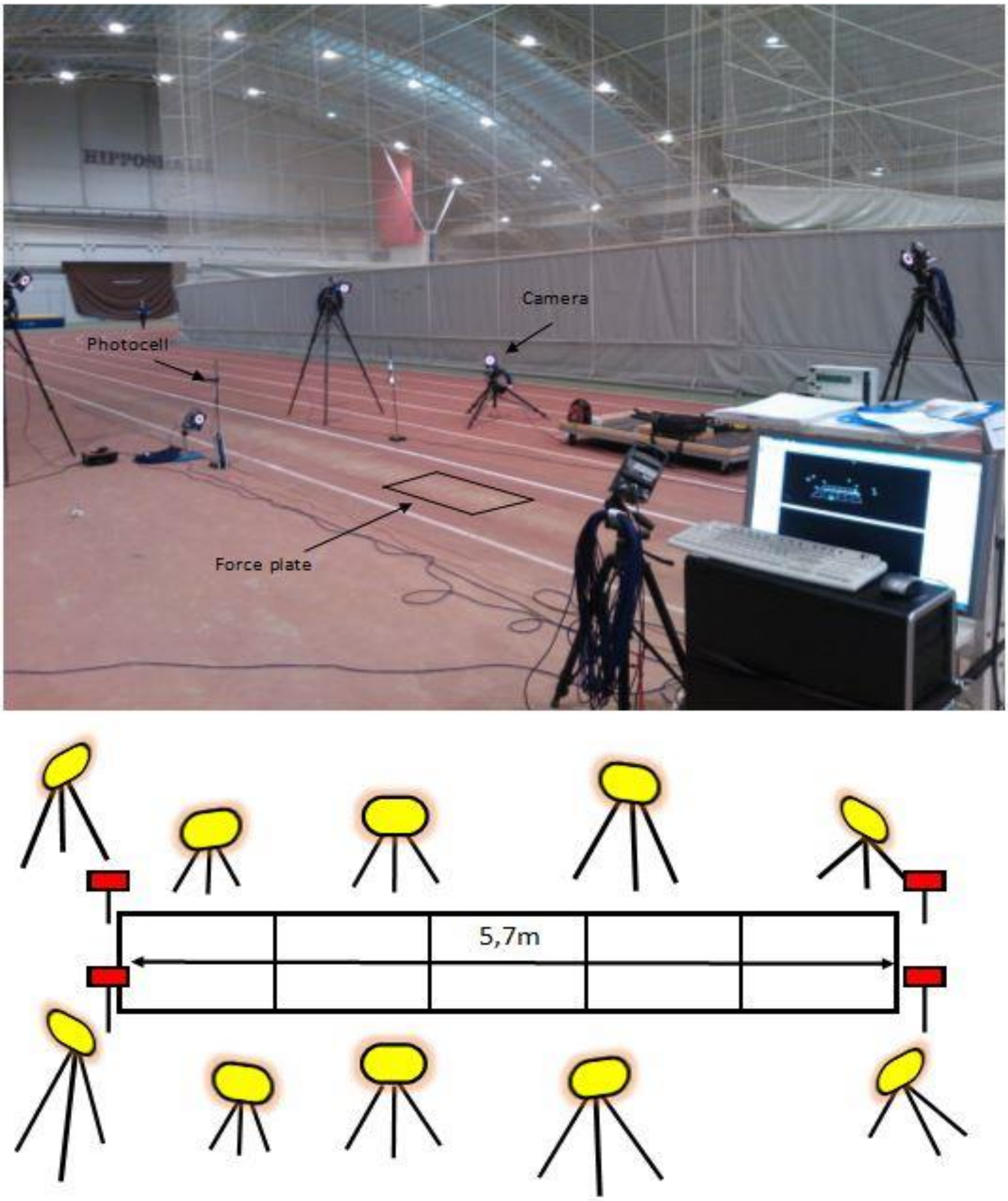

FIGURE 15. Experimental setup. Photo (above) and schematic (below).

TABLE 1. Markers and their locations.

Marker name	Position	R/B	Segment
SACR	Sacral marker: midway between the posterior superior iliac spines	B	Hip
ASI	Anterior Superior Iliac Spine	B	Hip
THI	Thigh	B	Femur
RKNE	Femoral condyle	B	Femur
RHFB	Head of Fibular	R	Tibia
RTUB	Tibial tuberosity	R	Tibia
RTIB	Tibial marker	B	Tibia
RSHN	Anterior aspect of the shin	R	Tibia
RANK	Lateral malleolus	B	Tibia
RMMA	Medial malleolus	R	Tibia
RPCA	Posterior proximal aspect of heel	R	Hindfoot
RCPG	Wand marker of posterior calcaneus aligned with transverse orientation	R	Hindfoot
RHEE	Posterior distal aspect of heel	B	Hindfoot
RSTL	Sustentaculum tali	R	Hindfoot
RLCA	Lateral calcaneus	R	Hindfoot
RP1M	Base of first metatarsal	R	Forefoot
RD1M	Head of first metatarsal	R	Forefoot
RTOE	Between second and third metatarsal heads	B	Forefoot
RD5M	Head of fifth metatarsal	R	Forefoot
RP5M	Base of fifth metatarsal	R	Forefoot
RHLX	Base of hallux	R	Hallux

Markers that are labelled red were only used in static trial and were removed for the dynamic trials. The markers that are labelled R were only placed on the right side. The markers labelled B were placed bilaterally.

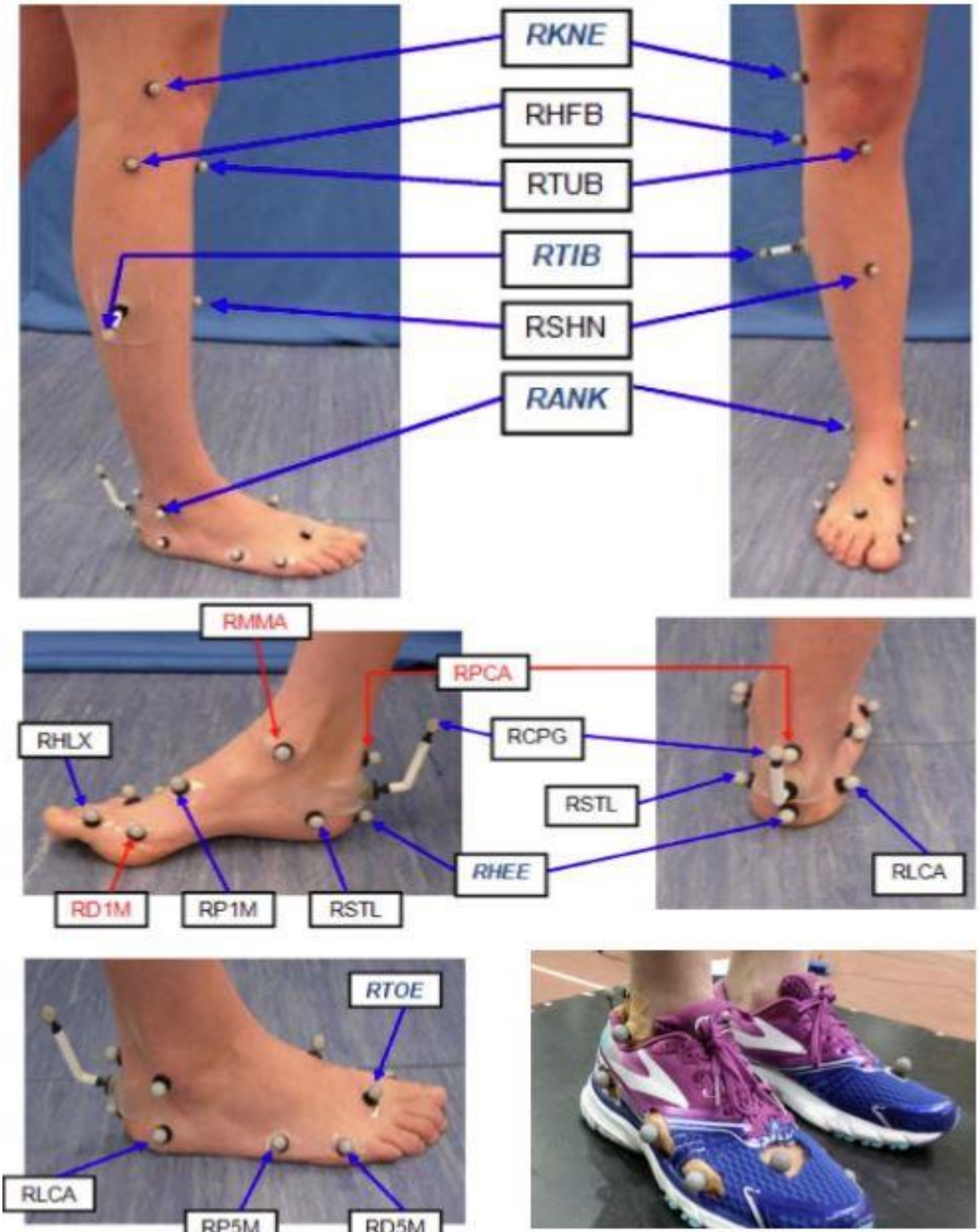

FIGURE 16. Markers of the lower leg. Lateral and frontal view (top). Medial and posterior view of the ankle and foot (middle). Lateral view of the ankle and foot (bottom-left). (modified from: Vicon 2012.) Bottom-right: one of the subjects is wearing the modified shoes.

7.3 The plug-in gait model and Oxford foot model

In this study the Vicon plug-in gait model and the Oxford foot model were used to calculate the kinematics and kinetics of running. The plug-in gait model was used to calculate ankle, knee and hip angles, moments and powers. The Oxford foot model was used to calculate interseg-ment angles between tibia, hindfoot, forefoot and hallux. The purpose of this chapter is to give a general idea of the principles of the plug-in gait (PIG) model. The Oxford foot model (OFM) is described in more detail in an attempt to aid the reader in interpreting the results.

7.3.1 Kinematic modelling

The PIG model is based on an assumption of rigid segments. The positions of the rigid segments are defined on a frame-by-frame basis. Each segment is defined by an origin in global (labora-tory) coordinates, and three orthogonal axis directions. In general, the three axis directions are defined using two directions derived from the marker data. One of these directions is taken as a dominant or principal direction, and used to directly define one of the axes in the segment. The second direction is subordinate to the first, and is used with the first direction to define a plane. The third axis of the segment is taken to be perpendicular to this plane. Then the second axis can be found that is perpendicular to both the first and third axes. (Vicon.)

The output angles for all joints are calculated from the cardan (euler) angles derived by com-paring the relative orientations of the two segments. These angles can be described as a set of rotations carried out one after the other (ordered rotation). For the relative rotations the proximal segment is "fixed" and the distal segment "moves". For example, Knee Rotation is a relative angle measured between the thigh as the proximal segment and the shank as the distal segment. (Vicon.) The definitions of the PIG model outputs can be seen in figure 17.

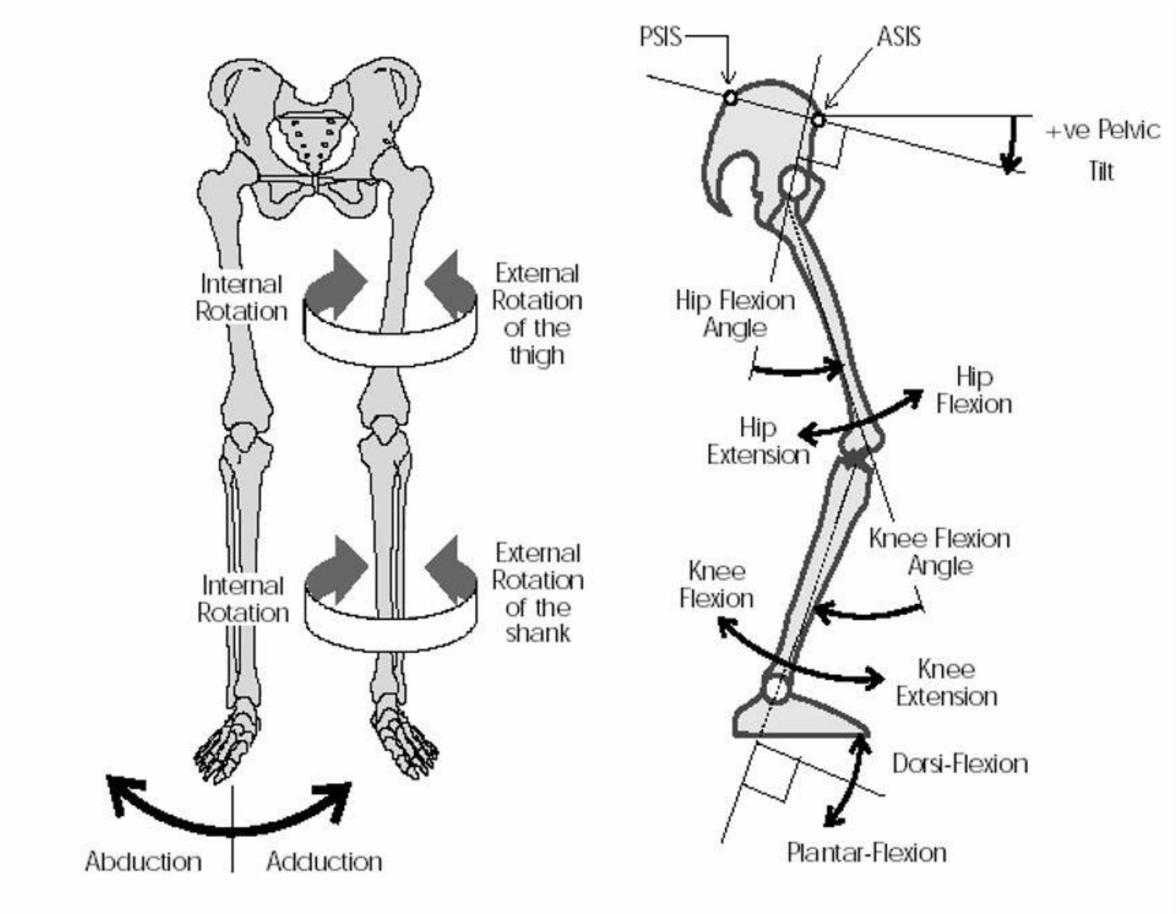

FIGURE 17. Definitions of PIG model variables (Vicon).

7.3.2 Kinetic modelling

The estimates of net joint moments are made by solving the equations of motion for the different segments of the lower limbs. Joint powers are calculated as the scalar product of moment and angular velocity. In order to accomplish this, three inputs are needed: 1) the values of all external forces applied to the limbs. 2) Distribution of mass within the limb segments. 3) Kinematics of the limb segments, including the location of the joint centres. Furthermore, two assumptions have to be made: 1) no external forces are applied, other than gravity and force plate measurements. 2) The segment masses, centres of gravity, and radii of gyration can be approximated from published tables. (Vicon.)

The kinetic modelling parts of the model assign masses and radii of gyration to the segments defined in the kinematic model. An estimate of the position of the centre of mass is required in the segment. This is defined as a point at a given proportion along a line from the distal joint centre (normally the origin of the segment) towards the proximal joint centre of a "typical" segment. The masses of each segment are calculated as a proportion of the total body mass. The principal axes' moments of inertia are calculated from (mass) normalised radii of gyration. (Vicon.)

7.3.3 The Oxford foot model

The Oxford foot model simplifies the complex anatomical structure of the foot into three segments: hindfoot (HF), forefoot (FF) and hallux (HX). These foot segments, together with the tibial segment (TB), compose the Oxford foot model (figure 18). The Oxford foot model considers the midfoot as a force-transmitting segment between the hindfoot and forefoot. (Carson et al. 2011.)

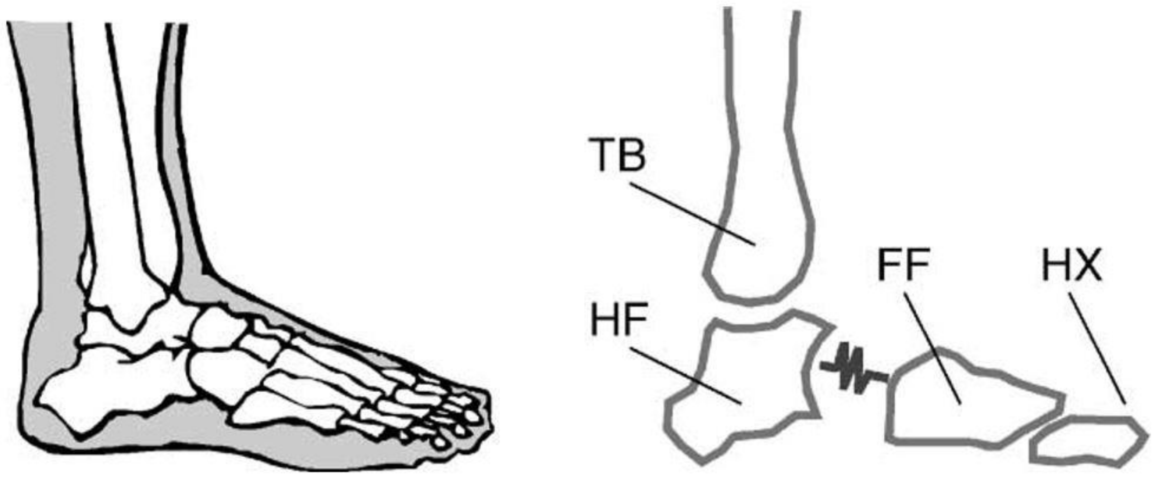

FIGURE 18. Schematic of the Oxford foot model segments. TB (tibia and fibula), HF (Calcaneus and talus), FF (five metatarsals) and HX (hallux proximal phalanx). (Carson et al. 2001.)

The *tibial segment* is composed of the tibia and fibula, and assumed to move as a single rigid body. The segment is based on the plane defined by the line from the knee joint centre to the ankle joint centre and the trans-malleolar axis. The longitudinal axis is from the ankle joint centre to the knee joint centre. The anterior axis is perpendicular to the plane defined by the longitudinal axis and the trans-malleolar axis. The transverse axis is mutually perpendicular. (Stebbins et al. 2006.)

The *hindfoot segment* is based on orientation of the mid-sagittal plane of the calcaneus in standing posture. The line that passes along the posterior surface of the calcaneus, and midway between the sustentaculum tali and lateral border of the calcaneus, defines the mid-sagittal plane. The anterior axis is from the most posterior aspect of the calcaneal tuberosity, in the plane defined above and parallel to the plantar surface of the hindfoot. The lateral axis is perpendicular to the mid-sagittal plane. (Stebbins et al. 2006.)

The *forefoot segment* is comprised of the five metatarsals, and assumed to move as a single rigid body. The segment is based on the plane defined by the centre of distal heads of the first and fifth metatarsal. The superior axis is perpendicular to this plane. The longitudinal axis is the projection of the line from the mid-point of the proximal heads of the first and fifth metatarsals to the mid-point of the distal heads of the second and third metatarsals into this plane. (Stebbins et al. 2006.)

The *hallux segment* is comprised of the proximal phalanx of the hallux and based on a longitudinal line along the proximal phalanx (Stebbins et al. 2006).

To get a more in-depth view of how different segments of the foot move in relation to each other and in relation to the tibia, intersegment angles between segment pairs can be analysed (table 2). The angular motion is examined in three anatomical planes (sagittal, frontal and transversal). The following inter-segment pairs are the most relevant: hindfoot/tibia (HF/TB), forefoot/tibia (FF/TB), forefoot/hindfoot (FF/HF) and hallux/forefoot (HX/FF).

TABLE 2. Angle, motion and axis definitions for each intersegment pair in each plane.

Inter-segment pair	angle definition	Plane	motion	motion axis
HF/TB	Hindfoot with respect to tibia	Sagittal	Plantar-/ dorsiflexion	about the transverse axis of the tibia
		Frontal	Inversion/eversion	about the longitudinal axis of the hindfoot
		Transversal	Internal/external rotation	about the common perpendicular axis
FF/TB	Forefoot with respect to tibia	Sagittal	Plantar-/ dorsiflexion	about the transverse axis of the tibia
		Frontal	Supination/ pronation	about the longitudinal axis of the forefoot
		Transversal	Abduction/ adduction	about the common perpendicular axis
FF/HF	Forefoot with respect to hindfoot	Sagittal	Plantar-/ dorsiflexion	about the transverse axis of the hindfoot
		Frontal	Supination/ pronation	about the longitudinal axis of the forefoot
		Transversal	Abduction/ adduction	about the common perpendicular axis
HX/FF	Hallux with respect to forefoot	Sagittal	Plantar-/ dorsiflexion	about the transverse axis of the forefoot

7.4 Data analysis

Marker trajectories and GRF data were low-pass filtered using a fourth-order Butterworth filter with cutoff frequencies of 12 and 50Hz, respectively. Kinetic and kinematic analyses were performed using the Oxford foot model (Vicon Nexus v.1.7, Oxford Metrics). Foot contact and toe-off events were used to calculate cadence, step length and contact time. Footstrike patterns were detected by calculating the footstrike angle (FSA), according to Altman and Davis (2012). The criterion for rearfoot strike (RFS), midfoot strike (MFS) and forefoot strike (FFS) were, FSA > 8º, -1.6º < FSA < 8º and FSA < -1.6º, respectively.

An inverse dynamics approach was used to calculate joint angles, moments and powers. Kinematic and kinetic data were time-normalised for the stance phase (0-100%) and averaged across at least three contacts to get individual mean curves and values for the parameters of interest using Polygon software (Vicon, UK). This was done for each subject and the averaged data was exported to Microsoft Excel, where it was further averaged across subjects to get condition specific mean values and mean curves for the parameters of interest. A statistical analysis was also carried out in Microsoft Excel. Differences between conditions were evaluated with a two-tailed paired t-test. P values lower than 0.05 were considered significant.

8 RESULTS

8.1 Spatiotemporal variables

Statistically significant differences between barefoot and shod condition were found in step length, stride length and cadence. Contact time did not reach a statistically significant difference. Stride and step length were shorter in barefoot condition when compared to shod condition. Cadence was higher in barefoot condition than in shod condition. The average values with standard deviation and p-values for each variable and for both conditions are listed on table 3.

TABLE 3: Means ± standard deviation for spatiotemporal variables

Variable	Barefoot	Shod	*P* value
Contact time (s)	0.21 ± 0.02	0.22 ± 0.02	0.052
Step length (m)	1.4 ± 0.03	1.45 ± 0.05	0.029*
Stride length (m)	2.79 ± 0.11	2.92 ± 0.12	0.006*
Cadence (1/min)	177 ± 8	174 ± 5	0.028*

*denotes for statistically significant difference between barefoot and shod conditions

8.2 Kinematic variables

8.2.1 Ankle angles

Besides maximum eversion angle, all the ankle angle parameters showed statistically significant differences between barefoot and shod conditions. At initial contact, the ankle was in a more plantarflexed position in barefoot than in shod condition. In barefoot condition, the ankle was in a slightly everted position at initial contact. In shod condition, however, the ankle was in a slightly inverted position at initial contact. Furthermore, the ankle was abducted in barefoot condition, while in shod condition the ankle was adducted at initial contact. Maximum dorsi-flexion was significantly higher in barefoot condition when compared to shod condition. The abduction range of motion was higher in shod condition than in barefoot condition. Table 4 shows the mean values for ankle kinematic parameters. Figure 19 (a) shows the progression of mean ankle angles in the stance phase.

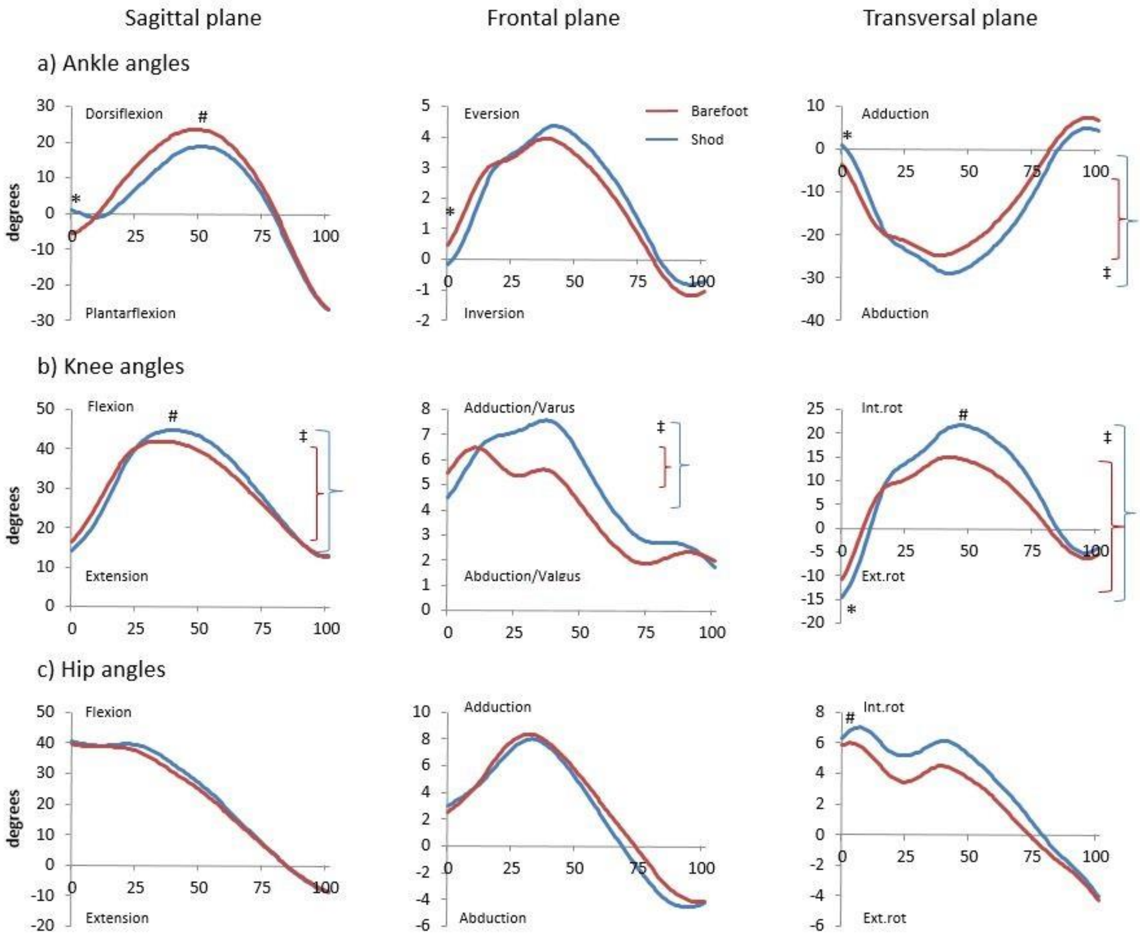

FIGURE 19. Progression of mean a) ankle angles b) knee angles c) hip angles plotted against stance phase (0-100%). * Significant difference between conditions at initial contact. # Significant difference between conditions in maximum. ǂ Significant difference between conditions in range of motion.

TABLE 4. Means ± standard deviation for ankle kinematic variables.

Parameter	**Barefoot (deg)**	**Shod (deg)**	***P* value**
Ankle			
Dorsiflexion at initial contact	-5.7 ± 7.5	1.0 ± 6.9	0.048*
Maximum dorsiflexion	23.7 ± 5.3	18.9 ± 5.6	0.0001*
Eversion at initial contact	0.5 ± 1.6	-0.18 ± 2.2	0.033*
Maximum eversion	4.0 ± 2.2	4.4 ± 2.1	0.471
Abduction at initial contact	3.5 ± 12.2	-1.1 ± 14.9	0.014*
Abduction range of motion	21.6 ± 7.1	30.3 ± 8.3	0.001*

*denotes for statistically significant difference between barefoot and shod conditions.

8.2.2 Knee and hip angles

At initial contact, regarding knee angles, the only statistically significant difference was found in the knee rotation angle, so that the knee was in a more externally rotated position in shod condition compared to barefoot condition. There was a trend of higher knee flexion angle and knee adduction ankle at initial contact during barefoot condition. However, the difference did not reach statistically significant level. Maximum knee flexion and maximum internal rotation were both higher in shod condition than in barefoot condition. The same was true for the range of motion in all planes.

Regarding hip angles, there was a higher maximum rotation and range of motion (in transversal plane) in shod condition compared to barefoot condition. All the other values at the hip were similar in both conditions. All the kinematic values of knee and hip are shown in table 5. Graphs of the progression of mean knee and hip angles are illustrated in figure 19.

TABLE 5: Means ± standard deviations for knee and hip angles in barefoot and shod conditions.

Parameter	**Barefoot (deg)**	**Shod (deg)**	***P* value**
Knee			
Flexion at initial contact	16.4 ±7.1	14.3 ± 7.7	0.068
Maximum flexion	42.3 ± 7.1	44.8 ± 7.4	0.002*
Flexion range of motion	25.9 ± 3.3	30.6 ± 3.6	0.002*
Adduction at initial contact	5.5 ± 4.4	4.5 ± 4.8	0.086
Maximum adduction	7.5 ± 6.1	9.1 ± 7.3	0.116
Adduction range of motion	2.1 ± 2.0	4.6 ± 4.0	0.014*
Internal rotation at initial contact	-10.8 ± 10.1	-14.5 ± 9.5	0.046*
Maximum internal rotation	15.2 ± 14.5	22 ± 14.8	0.001*
Internal rotation range of motion	26 ± 6.7	36.5 ± 8.1	0.001*
Hip			
Flexion at initial contact	39.7 ± 5.8	40.5 ± 7.1	0.469
Extension at toe off	8.7 ± 4.5	9.0 ± 5.2	0.709
Adduction at initial contact	2.5 ± 3.1	3.0 ± 2.4	0.499
Maximum adduction	8.5 ± 2.8	8.1 ± 3.1	0.361
Internal rotation at initial contact	5.8 ± 10.4	6.3 ± 10.6	0.637
Maximum internal rotation	6.7 ± 10.6	9.0 ± 10.7	0.031*
Internal rotation range of motion	0.9 ± 1.0	2.7 ± 2.5	0.015*

*Denotes for statistically significant difference between barefoot and shod conditions.

8.2.3 Footstrike pattern

There was a significant change in footstrike angle (FSA) when comparing barefoot and shod conditions. Mean values for footstrike angles in barefoot and shod conditions were 4.4° ± 6.5 ° and 18.2° ± 7.0, respectively. Three subjects RFS, four subjects MFS and two subjects FFS in barefoot condition. Eight subjects RFS and one subject MFS in shod condition. None of the subjects FFS in shod conditions. In total, five subjects modified their footstrike pattern from RFS to MFS, or from RFS to FFS, when condition changed from shod to barefoot. For the four subjects who did not change FSP, the trend towards lower FSA in barefoot condition is evident. Table 6 summon the FSA and FSP results.

TABLE 6. Subject specific and mean values for footstrike angle (FSA) and footstrike pattern (FSP) in barefoot and shod conditions.

Subject	Barefoot		Shod	
	FSA (deg)	FSP	FSA (deg)	FSP
1	2,7	MFS	24,0	RFS
2	-2,5	FFS	23,2	RFS
3	2,1	MFS	3,9	MFS
4	6,5	MFS	17,8	RFS
5	11,6	RFS	14,0	RFS
6	14,6	RFS	21,6	RFS
7	2,5	MFS	26,6	RFS
8	-5,7	FFS	13,6	RFS
9	8,0	RFS	18,8	RFS
Mean	4,4		18,2	
SD	6,5		7,0	

8.2.4 Foot angles

The terminology here follows the Oxford foot model convention. Please refer to table 2 for the terminology. The shapes of the intersegment angle curves were similar for both barefoot and shod conditions, except the hallux-forefoot angle curve, in which there was a notable difference in the curve shape between conditions. However, the magnitude of the intersegment angle values (at initial contact, maximum/minimum and range of motion) showed significant differences between conditions.

The hindfoot was dorsiflexed with respect to the tibia at initial contact in shod condition, as opposed to being plantarflexed with respect to the tibia at initial contact during barefoot condition. In the transversal plane, the hindfoot was externally rotated with respect to the tibia in barefoot condition, while in shod condition the hindfoot was slightly internally rotated. Compared to barefoot condition, the maximum external rotation angle was significantly greater in shod condition. In the sagittal plane, the range of motion (dorsiflexion) was greater in barefoot condition than in shod condition. The opposite was true in frontal (eversion range of motion) and transversal plane (external rotation range of motion).

Forefoot with respect to tibia angles showed significantly greater plantarflexion in barefoot condition compared to shod condition at initial contact. In fact, during shod condition, the forefoot was in a dorsiflexion with respect to the tibia in shod condition at initial contact. Maximum forefoot dorsiflexion with respect to the tibia was, however, greater in barefoot condition than in shod condition. In the transversal plane, the forefoot was more adducted with respect to the tibia at initial contact in shod condition than barefoot condition. The maximum abduction angle was greater in shod condition than in barefoot condition. In the sagittal plane, the range of motion (dorsiflexion) was significantly greater in barefoot condition than in shod condition. The opposite was true for the frontal (pronation range of motion) and transversal plane (abduction range of motion).

Regarding the forefoot, with respect to hindfoot angles, no significant differences between conditions were found at initial contact in the sagittal and frontal planes. In the transversal plane, however, the adduction angle was greater at initial contact in shod condition than in barefoot condition. In the sagittal plane, the maximum dorsiflexion angle and range of motion (dorsiflexion) was greater in barefoot than in shod condition. In the transversal plane, the minimum adduction angle was lower in barefoot than in shod condition.

The hallux, with respect to forefoot angle curve, is notably different in the two conditions. Maximum dorsiflexion and the dorsiflexion range of motion is significantly greater in shod condition than barefoot condition. Mean hindfoot-tibia, forefoot-tibia, forefoot-hindfoot and hallux-forefoot angle values are listed in tables 7, 8, 9 and 10, respectively. Mean inter-segment angle curves, plotted against stance phase in both conditions, can be seen in figure 20.

TABLE 7. Mean hindfoot with respect to tibia angles.

Hindfoot - Tibia angle	**Barefoot** (deg)	**Shod** (deg)	***P* value**
Sagittal plane			
Dorsiflexion at initial contact	-4.0 ±7.7	3.3 ± 7.0	0.028*
Maximum dorsiflexion	19.6 ± 7.2	21.0 ± 6.8	0.216
Dorsiflexion range of motion	23.6 ± 5.1	17.7 ± 4.8	0.036*
Frontal plane			
Eversion at initial contact	-5.1 ± 4.7	-7.2 ± 5.3	0.197
Maximum eversion	13.8 ± 7.6	16.3 ± 7.9	0.123
Eversion range of motion	18.9 ± 6.6	23.5 ± 5.9	0.009*
Transversal plane			
External rotation at initial contact	3.9 ± 13.3	-0.5 ± 16.7	0.043*
Maximum external rotation	24.8 ±17.5	29.5 ± 15.9	0.005*
External rotation range of motion	20.9 ± 8.6	29.9 ± 8.6	0.002*

*Denotes for statistically significant difference between barefoot and shod conditions.

TABLE 8. Mean forefoot with respect to tibia angles.

Forefoot - Tibia angle	**Barefoot** (deg)	**Shod** (deg)	***P* value**
Sagittal plane			
Dorsiflexion at initial contact	-2.3 ± 7.4	6.8 ± 7.7	0.015*
Maximum dorsiflexion	29.4 ± 6.6	23.3 ± 7.6	0.001*
Dorsiflexion range of motion	31.8 ± 7.0	16.4 ± 7.1	0.003*
Frontal plane			
Pronation at initial contact	-5.4 ± 5.7	-7.6 ± 4.6	0.136
Maximum pronation	20.2 ± 9.5	22.5 ± 9.5	0.097
Pronation range of motion	25.6 ± 8.0	30.1 ± 6.3	0.032*
Transversal plane			
Abduction at initial contact	-8.2 ± 13.8	-13.7 ± 16.1	0.005*
Maximum abduction	12.5 ± 16.6	17.3 ± 15.4	0.002*
Abduction range of motion	20.6 ± 7.0	30.9 ± 8.6	0.0004*

*Denotes for statistically significant difference between barefoot and shod conditions.

TABLE 9. Mean forefoot with respect to hindfoot angles.

Forefoot - Hindfoot angle	**Barefoot (deg)**	**Shod (deg)**	***P* value**
Sagittal plane			
Dorsiflexion at initial contact	0.6 ± 7.0	1.6 ± 5.0	0.410
Maximum dorsiflexion	12.5 ± 7.8	6.0 ± 5.3	0.003*
Dorsiflexion range of motion	11.9 ± 4.8	4.4 ± 2.3	0.001*
Frontal plane			
Pronation at initial contact	-1.0 ± 2.4	-0.5 ± 4.2	0.751
Maximum pronation	4.0 ± 5.1	6.3 ± 4.6	0.144
Pronation range of motion	5.0 ± 3.0	6.8 ± 1.7	0.136
Transversal plane			
Adduction at initial contact	11.9 ± 4.2	13.4 ± 4.7	0.021*
Minimum adduction	8.1 ± 3.9	10.4 ± 4.3	0.00004*
Abduction range of motion	3.7 ± 1.3	3.0 ± 1.3	0.081

*Denotes for statistically significant difference between barefoot and shod conditions.

TABLE 10. Mean hallux with respect to forefoot angles.

Hallux - Forefoot angle (sagittal plane)	**Barefoot (deg)**	**Shod (deg)**	***P* value**
Dorsiflexion at initial contact	5.7 ± 6.3	3.6 ± 7.8	0.483
Maximum dorsiflexion	6.6 ± 7.0	15.8 ± 6.3	0.008*
Dorsiflexion range of motion	1.0 ± 1.6	12.3 ± 2.8	0.000007*

*Denotes for statistically significant difference between barefoot and shod conditions.

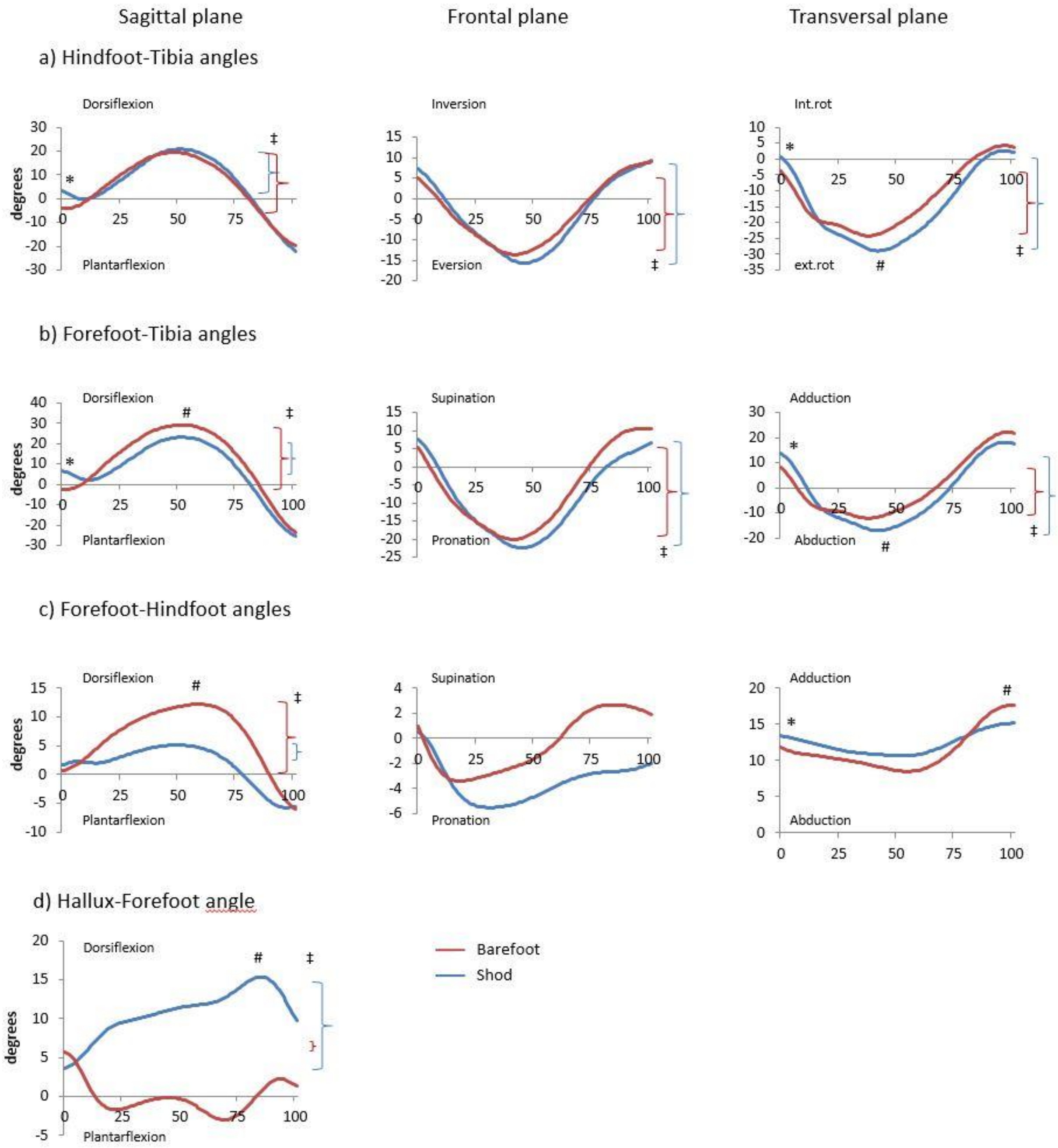

FIGURE 20. Mean intersegment angle curves plotted against stance phase (0-100%). * Significant difference between conditions at initial contact. # Significant difference between conditions in maximum. ‡ Significant difference between conditions in range of motion.

8.3 **Kinetic variables**

8.3.1 Ground reaction forces

No significant differences were found in ground reaction forces in vertical and medio-lateral directions. However, horizontal braking force was significantly higher in barefoot condition than in shod condition. There was a trend of higher propulsion force in barefoot condition but the difference did not reach statistically significant level. Mean values of maximum ground reaction forces are presented in table 11. Ground reaction force progression curves are presented in figure 21.

TABLE 11. Means of maximum ground reaction forces in both conditions.

Parameter	**Barefoot (%BW)**	**Shod (%BW)**	***P* value**
Horizontal braking force	46.7 ± 5.7	39.6 ± 2.9	0.008*
Horizontal propulsion force	39.4 ± 5.3	36.4 ± 3.7	0.098
Lateral horizontal force	11.6 ± 8.9	10.7 ± 7.9	0.589
Medial horizontal force	14.3 ± 5.0	12.7 ± 7.0	0.287
Vertical	272.6 ± 13.6	276.8 ± 16.0	0.408

*Denotes for statistically significant difference between barefoot and shod conditions.

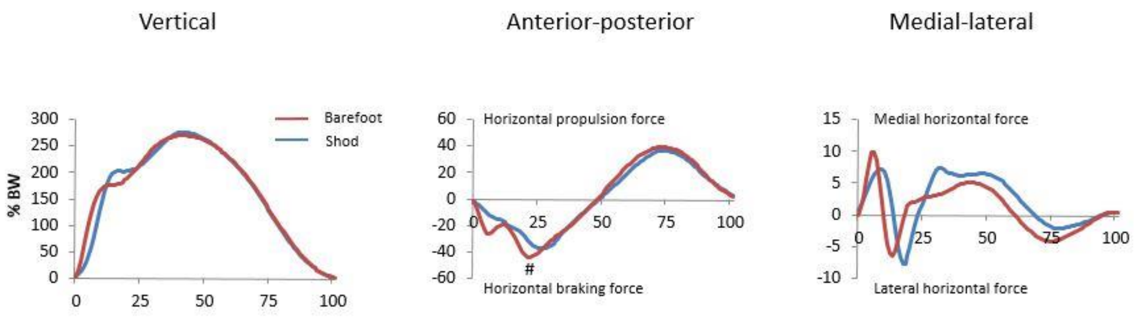

FIGURE 21. Mean ground reaction force curves. # Significant difference between conditions in maximum.

8.3.2 Ankle, knee and hip moments

The joint moment progression curve shapes are similar in both conditions. The only statistically significant difference in joint moments between the conditions was found at the knee abductor moment. The maximum knee abductor moment was significantly higher in shod condition than in barefoot condition. The internal joint moments can be found in table 12, and the joint moment progression curves, plotted against the stance phase, is presented in figure 22.

TABLE 12. The mean maximum values of internal joint moments in both conditions.

Parameter	Barefoot ($N \cdot m \cdot kg^{-1}$)	Shod ($N \cdot m \cdot kg^{-1}$)	*P* value
Ankle			
Plantarflexor moment	3.40 ± 0.38	3.27 ± 0.34	0.177
Dorsiflexor moment	0.16 ± 0.12	0.24 ± 0.13	0.166
Invertor moment	0.25 ± 0.11	0.24 ± 0.14	0.866
Maximum adductor moment	0.58 ± 0.35	0.56 ± 0.34	0.480
Knee			
Extensor moment	2.29 ± 0.71	2.38 ± 0.88	0.524
Abductor moment	1.84 ± 0.72	2.19 ± 0.62	0.005*
External rotator moment	0.20 ± 0.06	0.23 ± 0.09	0.321
Hip			
Extensor moment	2.24 ± 0.43	2.25 ± 0.33	0.953
Flexor moment	0.79 ± 0.30	0.89 ± 0.38	0.181
Abductor moment	2.13 ± 0.40	1.97 ± 0.41	0.085
External rotator moment	0.52 ± 0.14	0.58 ± 0.08	0.129

*Denotes for statistically significant difference between barefoot and shod conditions.

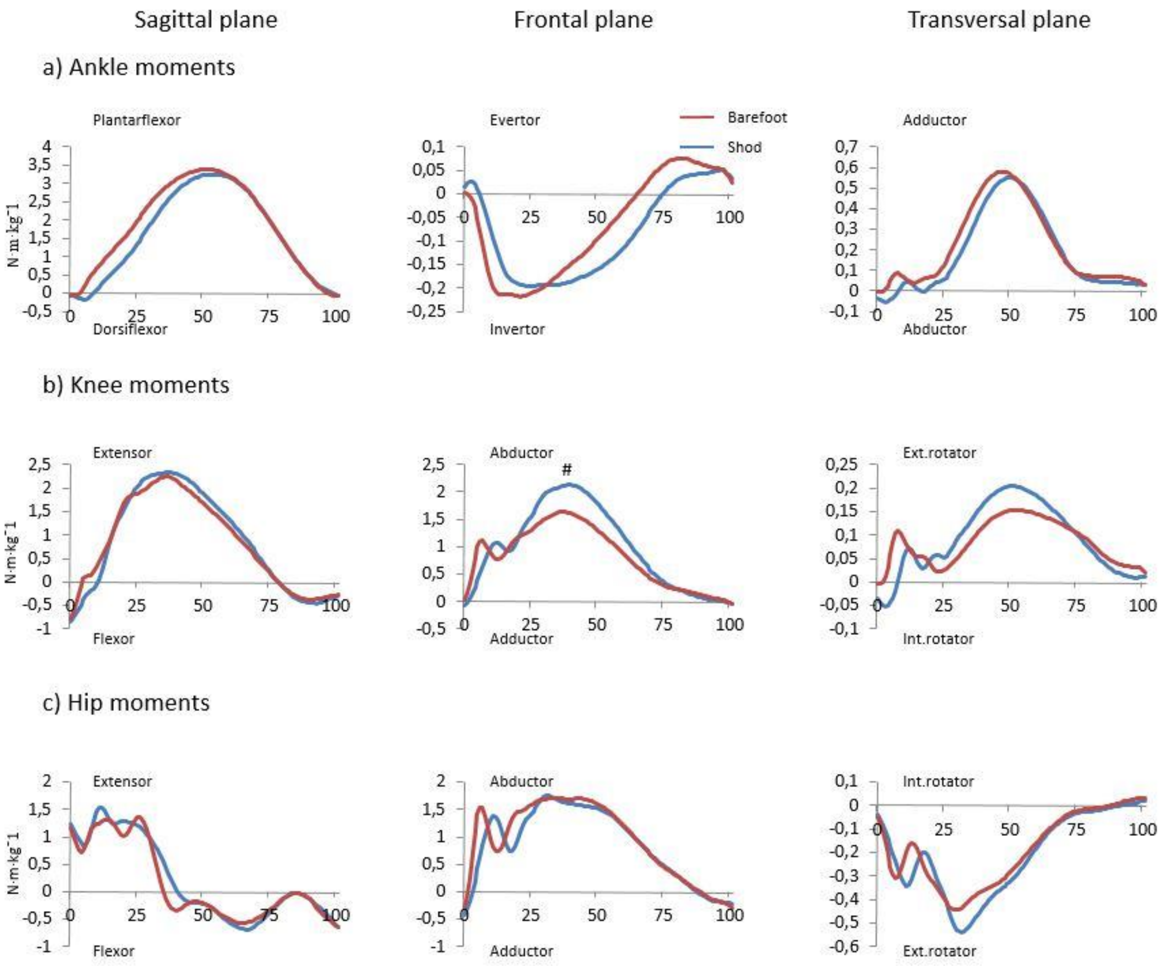

FIGURE 22. Mean internal joint moment curves plotted against the stance phase (0-100%) in both conditions. # Significant difference between conditions in maximum.

8.3.3 Powers

The maximum ankle generation and ankle absorption powers were both significantly greater in barefoot condition than shod condition. Regarding knee power, the opposite was true. The knee generation and absorption powers were both greater in shod condition than in barefoot condition. No statistically significant differences between the conditions were found in hip power. The ankle, knee and hip power are presented in table 13. Mean power progression curves are shown in figure 23.

TABLE. 13. Mean maximum ankle, knee and hip powers.

Power	**Barefoot** (W/kg)	**Shod** (W/kg)	***P* value**
Ankle generation	19.1 ± 3.4	16.1 ± 2.8	0.002*
Ankle absorbtion	11.6 ± 2.8	8.6 ± 1.6	0.011*
Knee generation	8.0 ± 2.6	9.7 ± 3.2	0.042*
Knee absorbtion	14.1 ± 4.3	18.1 ± 4.3	0.012*
Hip generation	4.1 ± 1.7	3.5 ± 1.5	0.345
Hip absorbtion	4.4 ± 1.6	4.9 ± 1.5	0.099

*Denotes for statistically significant difference between barefoot and shod conditions.

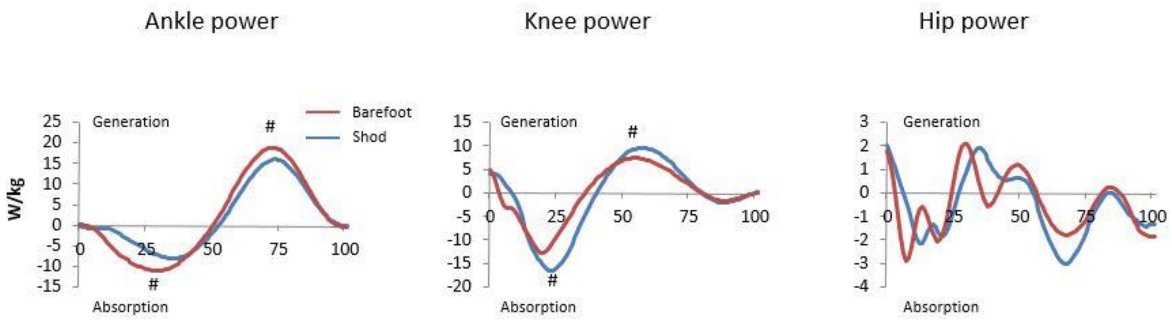

FIGURE 23. Mean ankle, knee and hip power curves plotted against stance phase (0-100%). # Significant difference between conditions in maximum.

9 DISCUSSION

The results of spatiotemporal variables showed an increased cadence, as well as shorter step and stride lengths when running in barefoot condition than shod condition. These results are in accordance with several previous studies (De Wit et al. 2000; Squadrone & Gallozzi 2009; Schütte et al. 2013; Bonacci et al. 2014; McCallion et al. 2014). The results suggest that differences in spatiotemporal variables can occur within a single running session in habitually shod males.

It has been reported in previous studies that barefoot running facilitates a more horizontal foot (more plantarflexed ankle angle) at initial contact (De Wit et al. 2000; Bishop et al. 2006; Squadrone & Gallozzi 2009; Lieberman et al.2010 Bonacci et al. 2013). In the current study, the ankle was significantly more plantarflexed in barefoot condition than in shod condition. In barefoot condition there was also a trend of greater angles of knee flexion at initial contact than in shod condition. Because the hip angle at initial contact was similar in both conditions, the lower footstrike angles (FSA) observed in barefoot condition can primarily be attributed to the more plantarflexed ankle and, to some extent, the more flexed knee. All of the nine subjects in the current study had lower FSA in barefoot than in shod condition. The decrease in FSA resulted in a footstrike pattern (FSP) alteration in five of the subjects as three subjects changed the FSP from rearfoot strike to midfoot strike and two subjects changed the FSP from rearfoot strike to forefoot strike.

There are two suggested explanations for the change in FSP when changing from shod running to barefoot running. De Wit et al. (2000) showed significantly larger loading rates during impact in barefoot condition than in shod condition running, but no difference in impact peak or peak ground reaction force between conditions. They also found that pressure underneath the heel was correlated negatively with the sole angle at initial contact. Referring to these findings, De Wit et al. (2000) proposed that runners adopt a flatter foot placement in an attempt to reduce local pressure underneath the heel. The other side of the coin is that barefoot running (De Wit et al. 2000; Squadrone & Gallozzi 2009) and FFS patterns (Nunns et al. 2013) have been found to increase plantar pressure in the forefoot region. This has been suggested to increase the risk of metatarsal stress fractures (Nunns et al. 2013). There are some case studies which suggest that the transition to minimal footwear can lead to metatarsal stress fractures (Giuliani et al. 2011; Ridge et al. 2013). Another proposed explanation for flatter foot placement is an attempt to reduce impact peak magnitudes and loading rates (Lieberman et al. 2010; Thompson et al. 2015). Contrary to the findings of De Wit et al. (2000), reduced impact peak forces (Squadrone & Gallozzi 2009; Lieberman et al. 2010; Thompson et al. 2015) and reduced loading rate (Lieberman et al. 2010; Shih et al. 2013) in barefoot running compared to shod running have been reported. In the studies of Squadrone & Gallozzi (2009) and Lieberman et al. (2010) the reduced impact peak forces and loading rate were related not only to barefoot condition, but to forefoot strike pattern, indicating that the key factor is the FSP and not the running condition. The study of Kulmala et al. (2013) supports this finding. They reported that impact peak force and average vertical loading rate were both reduced in FFS compared to RFS runners. Lieberman (2012) claims that reduced impact peak and loading rate are a result of greater compliance of the lower leg in FFS pattern. The greater compliance is achieved by landing with a more plantarflexed ankle and flexed knee, as well as greater ankle dorsiflexion and knee flexion during the period of impact, allowing the lower extremity to dampen forces more effectively. However, FSP may not be the only factor involved in reducing impact forces, as Cheung (2013) noted a reduced loading rate irrespective of the FSP adopted. In the current study, impact peak force and impact loading rate were not analysed and the peak ground vertical force was similar between conditions. From visual examination of figure 21, it seems that in the current study, impact peak force is reduced in barefoot condition but occurs earlier, that is, the loading rate is increased in barefoot condition. In the current study, only two subjects used a full forefoot strike pattern in barefoot condition. Therefore, it might be that alterations in the FSP patterns between conditions were sufficient for reducing impact peak, but insufficient for reducing loading rate. However, this is highly speculative since the impact peak and loading force were not analysed. The results of different studies examining the impact peak and loading rate are equivocal. There are studies that observe no difference between barefoot and shod running, and there are studies that observe higher loading rates and impact peaks in barefoot compared to shod and vice versa (Warne 2014). Furthermore, while there is some evidence that links higher impact peaks and loading rates to injuries (Hreljac 2004), one of the leading authors (Nigg 2001) has questioned the association between impact forces and injury.

Sinclair (2015) reported evidence that conventional running shoes may interfere with normal foot functioning and reduce the extent of the work output of the foot muscles, leading to weakness in foot musculature. In relation to shod running, the average and peak forces of the flexor digitorum longus (FDL), flexor hallucis longus (FHL) and peroneus longus (PL) were all significantly larger when running barefoot. Contrarily, the average and peak forces of extensor digitorum longus (EDL) and extensor hallucis longus (EHL) were significantly larger in shod condition than in barefoot condition. Although muscle forces were not analysed in the current study, the results of the current study indirectly support the findings of Sinclair (2015). In barefoot condition the ankle exhibits more plantarflexion than in shod condition. This can be seen in ankle angles (figure 19 a) and in hindfoot/forefoot-tibia angles (figures 20 a and b). This is reflected in the plantarflexor moment, as there is a trend towards a larger plantarflexor moment in barefoot condition. However, the difference did not reach a significant level. The power value at the ankle indicates higher demands of the foot musculature, as the absorption and generation powers were both greater in barefoot condition.

The weakness in the plantar foot musculature may lead to a lowering of the medial longitudinal arch, which would reduce the potential for elastic energy storage, and promote higher levels of foot eversion (Sinclair 2015). An examination of the hindfoot-tibia (figure 20 a) and forefoot-tibia (figure 20 b) intersegment angles in the frontal plane reveals that, when compared to shod condition, the range of motion was reduced in barefoot condition. However, in the current study, subjects were habitually shod runners with rearfoot strike (in shod condition). Therefore, even if true, the potential strength benefits of barefoot running cannot explain the reduced eversion/pronation motion in barefoot condition. The reason might lie in proprioception. It is proposed that cushioned high-heeled running shoes limit proprioception (Lieberman et al. 2010). If that were the case, enhanced proprioception would recruit small muscles of the foot, which are suited for sensing changes in foot abduction/adduction and inversion/eversion (Nigg 2009). Whether or not barefoot running enhances foot muscle strength, the reduced eversion/pronation motion in barefoot condition observed in the current study might have clinical significance. The current study showed reduced tibial rotation in barefoot condition compared to shod condition, as knee maximum internal rotation and internal rotation range of motion were both greater in shod condition. Furthermore, maximum external rotation and the external rotation range of motion of the hindfoot, with respect to the tibia, and the maximum abduction and abduction range of motion of the forefoot, with respect to the tibia, were both reduced in barefoot condition. These changes can be seen as a result of the decreased eversion/pronation motion of the foot in barefoot condition. It is also worth mentioning that the reduced knee internal rotation in barefoot condition can be independent of footstrike pattern, as Boyer and Derrick (2015) found that peak knee internal rotation was not affected by footstrike pattern. Increased knee internal rotation at the baseline was associated with later development of iliotibial band syndrome (ITBS) in the prospective study of Noehren et al. (2007). Furthermore, the review by Aderem and Louw (2015) concluded that female shod runners with ITBS appear to have increased peak knee internal rotation when compared to healthy runners. Therefore, the reduced peak knee internal rotation observed in the current study might have clinical significance. It must be noted, however, that the studies of Noehren et al. (2007) and Aderem and Louw (2015) had females as subjects. The significance of the reduced peak knee internal rotation observed in the current study, with respect to injury prevention, is therefore questionable as the subjects were males.

An interesting difference between conditions was observed in the intersegment angle of hallux, with respect to the forefoot. In barefoot condition and in relation to the forefoot, the hallux went to plantarflexion early in the stance phase and turned into slight dorsiflexion late in the stance phase. On the contrary, in shod condition the hallux was in dorsiflexion in relation to the forefoot during the whole stance phase. To the best knowledge of the author, this is the first time that this difference in the hallux-forefoot angle between barefoot and shod conditions is reported. A speculative explanation for the marked difference in the hallux-forefoot angle between conditions could be drawn by analysing the hindfoot-tibia, forefoot-tibia and forefoot-hindfoot angle curves (figures 20 a, b and c). It can be seen, from the hindfoot-tibia and forefoot-tibia curves, that in barefoot condition the curve travels above the curve in shod condition, from the point of approximately 20% of the stance phase. This means that at any point after 20% of the stance phase the external rotation/abduction of the hindfoot/forefoot, in relation to the tibia, is greater in shod condition. Moreover, the forefoot-hindfoot curve in the frontal plane reveals that at any point after approximately 20 % of the stance phase, the forefoot is in a much less pronated position with respect to the hindfoot in barefoot condition than in shod condition. The combination of these above- mentioned differences suggest that the foot rolls more over the lateral side of the phalanges in barefoot condition and more over the hallux in shod condition. This would be seen as more lateral centre of pressure during the late stance phase. However, pressure measurements were not carried out in the current study. The prospective study of Willems et al. (2006) observed that increased lateral roll-off was a risk factor for exercise-related lower leg pain (ERLLP). This result proposes that barefoot running with more lateral roll-off would be detrimental. However, there was no barefoot condition in the study of Willems et al. (2006) and no information was given about footstrike patterns of the subjects. Therefore, interpretation of the results of the current study with respect to the study of Willems et al. (2006) must be done with caution.

In the current study, an increased peak knee adduction angle was observed in shod condition compared to barefoot condition. A higher knee abductor moment observed in shod condition is most likely a result of the higher knee adduction, as it increases the length of the GRF leaver arm at the knee in the frontal plane. A higher knee abductor moment has been linked to the presence (Baliunas et al. 2002) and progression (Miyazaki et al. 2002) of knee osteoarthritis. Furthermore, the prospective study of Stefanyshyn et al. (1999) showed that a higher knee abductor moment at baseline was related to the development of patellofemoral pain syndrome. The lower knee abductor moment observed in barefoot condition might, therefore, have clinical relevance.

There were no differences in other moment values (except for the peak knee abductor). This is contrary to previous studies, in which it was found that the knee extensor moment was lower (Bonacci et al. 2013; Bonacci et al. 2014; Sinclair 2014) and the ankle plantarflexor moment was higher (Bonacci et al. 2013; Sinclair 2014) in barefoot condition compared to shod condition. Furthermore, the same studies have found reductions in the patellofemoral joint stress (Bonacci et al. 2014) and patellofemoral joint load, and an increase in the Achilles tendon load (Sinclair 2014) in barefoot condition compared to shod condition. Although patellofemoral and Achilles tendon forces were not analysed in the current study, the results obtained from joint power calculations supports the findings of the studies of Bonacci et al. (2014) and Sinclair (2014). It has been speculated that the reduced patellofemoral force in barefoot condition could be a result of increased energy absorption at the ankle joint, which reduces the amount of energy absorption at the knee joint (Sinclair 2014). The increased ankle joint absorption and reduced knee joint absorption found in the current study supports this speculation. Kulmala et al. (2013)

studied the effect of FSP on joint loadings. They found, in comparison to rearfoot strikers, reduced patellofemoral stress and increased Achilles tendon loading in forefoot strikers. Therefore, it could be possible that the reductions in patellofemoral stress and increase in Achilles tendon loading might be due to forefoot strike pattern. Interestingly, the reduction in patellofemoral stress was found without concomitant reduction in the knee extensor moment. The lack of difference in knee moments in the current study does not, therefore, exclude the possibility that there would have been reductions in patellofemoral stress if it had been analysed. It must be noted that these changes in patellofemoral stress and Achilles tendon loading have been observed when studying the kinetics/kinematics of averaged stance phases. Although barefoot running decreased patellofemoral stress in these studies, the concomitant increase in stride frequency might offset the potential benefits. The same applies for the greater Achilles tendon load seen in barefoot condition. The conclusion regarding the above discussion is that barefoot running could reduce patellofemoral stress and therefore decrease the risk of knee injuries. On the other hand, increased ankle involvement in shock absorption and Achilles tendon loading could predispose to ankle and foot injuries.

The current study has limitations that warrant consideration when interpreting the results. A relatively low number of subjects participated in the study. Furthermore, the kinematic and kinetic data were averaged across over 3-5 stance phases. The running pattern observed during this limited number of gait cycles might not accurately represent the running pattern that the subjects obtain when running in "normal" conditions. The modified shoes allowed the exact same positioning of the markers in both conditions. However, the holes that were made to the shoes might compromise the structural integrity of the shoes and therefore change the properties of the shoes. It has been shown that the largest hole that does not disrupt shoe integrity is an oval of 1,7cm x 2,5cm (Shultz & Jenkyn 2012). The size of the hole in this study was not measured, it is therefore possible that some of the holes were big enough to disrupt the integrity of the shoe. The results of the Oxford foot model angles have found to be the least repeatable in the transverse plane (Stebbins et al. 2006). Therefore, the results of the transverse plane must be interpreted with caution.

10 CONCLUSION

The results of the current study showed increased cadence and shorter step and stride lengths when running in barefoot condition compared to shod condition. FSA was significantly reduced in barefoot condition and led to changes in FSP from RFS to MFS (3 subjects), and from RFS to FFS (2 subjects) when condition changed from shod to barefoot.

The peak vertical ground reaction force was similar between conditions. It seems that impact peak force was reduced, but the impact loading rate was increased in barefoot condition compared to shod condition. However, these variables were not analysed. The significance of impact peak force and loading rate as an injury risk factor remains equivocal.

At the knee joint level, few variables indicate that barefoot running might have the potential for injury prevention at the knee joint. Firstly, a significant reduction in eversion and pronation range of motion was found in barefoot condition in comparison to shod condition. This led to reduced internal knee rotation in barefoot condition. The reduced knee internal rotation, together with reduced peak abductor moment in barefoot condition compared to shod condition, could have clinical relevance. Increased knee internal rotation has been associated with the development (Noehren et al. 2007) and prevalence (Aderem & Louw 2015) of iliotibial band syndrome. Furthermore, increased knee abductor moment has been linked to the presence (Baliunas et al. 2002) and progression (Miyazaki et al. 2002) of knee osteoarthritis and to the development of patellofemoral pain syndrome (Stefanyshyn et al. 1999). Secondly, reduced knee power absorption in barefoot condition compared to shod condition could be associated with reduced patellofemoral joint stress (Bonacci et al. 2014) and patellofemoral joint load (Sinclair 2014) observed in previous studies. However, the patellofemoral joint stress was not analysed in the current study.

In comparison to knee joint level, the results at the ankle joint level suggest that barefoot running might be hazardous regarding risk of injury to the ankle and foot. The power absorption at the ankle was increased in barefoot condition compared to shod condition. This might be associated with a previously found increased Achilles tendon load (Sinclair 2014) in barefoot condition. However, Achilles tendon load was not analysed in the current study. There was also an interesting difference between conditions in the hallux-forefoot angle. The maximum dorsiflexion and dorsiflexion range of motion was significantly higher in shod condition than in barefoot condition. This finding, together with the results obtained from the hindfoot-tibia, forefoot-tibia and hindfoot-forefoot angles, suggest that the foot rolls over the metatarsophalangeal joints more laterally. This could be detrimental, as it has been reported that increased lateral roll-off was a risk factor for exercise-related lower leg pain (Willems et al. 2006).

In conclusion, the results of the current study suggest that barefoot running could be beneficial regarding injuries at the knee level, but detrimental regarding injuries at the ankle level. Only two subjects obtained a forefoot strike pattern in barefoot condition. Therefore, these results suggest that some of the differences in kinetic/kinematics between barefoot and shod conditions can occur without change to forefoot striking. The variability in age, gender and research methodology between the studies makes it difficult to compare different studies and might lessen the strength of the evidence of the results of the current study. In the future, more studies are needed that would differentiate the effect of barefoot condition and footstrike pattern on kinetic and kinematic variables. Moreover, there is need for a standard method for evaluating footstrike pattern. More prospective studies are needed to clarify which kinematic/kinetic variables are risk factors for running-related injuries.

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