BIOMECHANICS OF THE PLANTAR FASCIA IN RUNNING AND THE IMPLICATION FOR PLANTAR FASCIITIS

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Chen Linwei

ABSTRACT

Running is an inexpensive form of physical exercises and has been gaining popularity in the past decades. However, the growth of the healthy lifestyle is accompanied by an increased risk of running-related injuries, among which plantar fasciitis is one of the preponderances. With the numerous efforts in studying the biomechanics of plantar fascia, researchers intended to minimize the risk of the problems by optimizing exercise paradigms or utilizing taping treatments. Nevertheless, challenges in assessing the loading environment and internal conditions of the plantar fascia impede the understanding of its interaction with these extrinsic factors.

Hereinafter, the overall objective of this study was to investigate the internal biomechanical feature and mechanical environment of the plantar fascia that provides more insights for both scientists and runners. The scope of the study included the investigation of 1) biomechanical property of plantar fascia and 2) plantar fascia loading upon different foot strike techniques in running, and 3) the offloading effect of different taping techniques, which were achieved by ultrasound elastography, locomotion analysis, musculoskeletal modeling, and finite element analysis.

In the first study, thirty-five recreational runners using different foot strike techniques received the ultrasound elastographic measurements. Forefoot strikers exhibited reduced plantar fascia

elasticity compared to rearfoot strikers, which could potentially relate to the pathomechanics of plantar fasciitis. Immediately after, the second study aimed at reconstructing a finite element model of a typical runner and examining the differences in plantar fascia loading between forefoot strike running and rearfoot strike running. The model simulation was driven by data of the locomotion analysis and musculoskeletal modeling of the participant, following the procedures of a mesh convergence test and model validation. The predictions showed that, compared to rearfoot strike, forefoot strike increased the tensile force on the plantar fascia and depressed the medial longitudinal arch. The findings further supported the additional risk of plantar fascia injury contributed by forefoot strike.

Using the same model, in the third study, we evaluated two tapping techniques (Fascia taping and Low-Dye taping) and their capabilities to offload the plantar fascia during running. The prediction showed that Fascia taping was more effective in reducing the strain of the fascia band and increasing the arch height, whilst the effects of Low-Dye taping seemed unapparent.

In conclusion, we investigated different biomechanical features of the plantar fascia, from intrinsic, extrinsic factor to assistive device (taping). The significance of this study lay in its potential to provide quantitative evidence suggesting risk factors of plantar fascia problems by integrating ultrasonography, gait experiments and computer simulations, otherwise difficult to be assessed using in-vivo experiments alone. Moreover, we overcame some technical challenges in simulating highly dynamic scenarios with deep tissue deformation using a smooth particle hydrodynamics (SPH) technique, which was traditionally used for fluid or astrophysical simulation. Future studies shall consider a detailed material constitutive model of the plantar fascia considering the yielding and micro-torn responses. In addition, a prospective, controlled study can be conducted to confirm the causal relationship between the risk factors and plantar fasciitis.

PUBLICATIONS

Publications arising from the thesis

Journal Paper

Chen, T.L.-W., Wong, D.W.-C., Peng, Y., Zhang, M., 2020. Prediction on the strain alteration of plantar fascia upon Fascia taping and Low-Dye taping during running. Journal of Orthopaedic Translation 20, 113–121.

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Wong D. W. -C., Wang, Y., Yang, M., **Chen, T.L.-W**., Leung, A.K.-L., Zhang, M. Metatarsal stress after subtalar arthroereisis. Proceedings of The 8th World Association for Chinese Biomedical Engineers (WACBE) World Congress on Bioengineering, Hong Kong, SP2.1-4, 30 Jul – 2 Aug 2017.

Wang, Y., Wong, D.W.-C., Tan, Q., **Chen, T.L.-W.,** Zhang, M. Biomechanical evaluation of the design of the total ankle arthroplasty. Proceedings of The 8th World Association for Chinese Biomedical Engineers (WACBE) World Congress on Bioengineering, Hong Kong, C2-1, 30 Jul – 2 Aug 2017. *[Young Investigator Award]*

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LIST OF ABBREVIATIONS

AHI Arch Height Index

BW Body-weight

EMG Electromyography

GRF Ground Reaction Force

ICC Intraclass Correlation Coefficients

MLA Medial Longitudinal Arch

MRI Magnetic Resonance Image

MTP Metatarsophalangeal

SPH Smooth Particle Hydrodynamics

SWV Shear Wave Velocity

CHAPTER 1 INTRODUCTION

1.1. Background

Humans are natural runners. The ability of running is a part of the ancient gene that is tied to the evolutionary process of the human race (Heinrich, 2002). For either proposed reasons—pursuing predators on the open savannahs or scavenging carcasses of dead animals, the primitive sense of survival helps to shape the most distinguishable human anatomical features—spring-like leg tendons, propel-facilitate foot structure, flexible shoulder, and vast sweat glands—that make humans uniquely excel at endurance running (Bramble and Lieberman, 2004). Running confers the advantages of persistence hunting to humans to survive the natural selection before weapons were created and possibly transformed human to its modern form.

In 490 BC, Pheidippides ran from the town of Marathon to Athens to bring news that the Greek army defeated the Persian army and dropped dead on arrival (Tsiotos, 2001). Centuries later, in the late 1890s, the first Olympic marathon was held in Athens and the oldest marathon of the world that is still organized today was founded in Boston (Lucas, 1976). Since then, as the participation extends across continentals, races, and genders, marathon running has changed from an elite sports competition to a mass recreational leisure event. Nowadays running is one of the most popular forms of exercise. The number of regular runners in the United States had reached 55.9 million in 2017 (Schwellnus and Derman, 2014). The biggest marathon event—the New York City marathon sees over 50,000 runners competing every year (Vitti et al., 2019).

Along with this rising popularity of running, the number of running-related injuries also increase. The predominant site of these injuries is the lower limb. Depending on the nature of the studies, the annual incidence rate of lower extremity running injury was reported to range from 19.4% to 79.3% (van Gent et al., 2007) or 6.8 to 59 injuries per 1000 hours of exposure to running (Lopes et al., 2012). Injury not only interferes with runners' training routine and affects their performance, it also incurs a large amount of cost as a result of healthcare expenses. In this regard, prevention of running-related injuries is important and should draw attention from either runners or scientists.

Among many established training concepts that intend to reduce the injury risk, foot strike pattern modification has recently sparked many interests in the runner community. Particularly after the publication of "Born to run" (McDougall, 2010) in 2009 reheated the topic of "natural running" (Abshire, 2010; Gouttebarge and Boschman, 2013), debates on the ideal type of foot strike have come to a fierce point. Foot strike pattern defines which part of the foot contacts the ground first at touchdown in running gait. Benefiting from the modern footwear technology, most of the shod runners these days are habitual rearfoot strikers (de Almeida et al., 2015). People do not notice that they land on the heel due to the massive comfort provided by the shoes' midsole cushion (Lieberman et al., 2010). However, as the landing style could influence force transmission within the leg and thus relate to the injury profile, there is a resurgence of argument on how runners should hit the ground for the best interest of running healthy (Hamill and Gruber, 2012). The basic theory is that, compared to rearfoot strike, forefoot strike better harnesses the shock-absorbing properties of foot soft tissues, thereby reducing the impact shock when the runners crash into the ground and softening the loading to the lower limb joints (Shih et al., 2013; Williams et al., 2012).

Though forefoot strike possesses many evidence-based merits in terms of injury prevention, there are still various counterarguments regarding its possible drawbacks. It was found that, after the transition from rearfoot strike to forefoot strike, the overall lower limb force was redistributed among the segments (Fuller et al., 2016). More specifically, forefoot strike may offload the knee joint at the expense of increasing the loading on the ankle and foot (Fuller et al., 2016). Excessive mechanical burdens can be associated with the pathologies of many foot injuries, such as plantar fasciitis.

Plantar fasciitis is a foot problem primarily resulting from overstraining on the fascia band (Wearing et al., 2006b). Overstraining can cause micro-tears of the fibral tissue, trigger a series of pathologies, including local inflammation and collagen degeneration (Walden et al., 2016). These histological changes in combination can induce pain symptom and lead to plantar fasciitis. Plantar fasciitis accounts for 11% to 15% of all foot complaints that require clinical counseling (Buchbinder, 2004). In America, the number of plantar fasciitis patients is 2 million per year (Irving et al., 2006). Compared to ordinary people, regular runners are more vulnerable to plantar fasciitis. About 10% of the runner population suffers from plantar fasciitis on an annual basis (Wearing et al., 2006b).

Based on the above statements, some scholars speculate that runners using forefoot strike are more likely to fall victim to plantar fasciitis, in spite of that this concern is mostly theoretical. Forefoot strike running oftentimes features a plantarflexed ankle at initial contact, consequently the ground reaction force (GRF) shifts anterior to the ankle center and acts on the forefoot (Stacoff et al., 2000). During this period, a larger Achilles tension is generated to lift the heel off the ground and counteract the ankle dorsiflexion moment caused by the GRF (Landreneau et al., 2014). Together with the body force applied on the ankle joint, a three-point-bending motion was formed to compress the longitudinal foot arch (Perl et al., 2012). To prevent arch flattening under this loading scene, the plantar foot connective tissues, e.g. the plantar fascia, can be increasingly strained and thus prone to plantar fasciitis (McKeon et al., 2015).

To support this theory and link it to the epidemiology, researchers need to establish an evidentiary framework from both experimental and clinical domains. However, there is currently a scarcity of studies addressing the differences between rearfoot strike and forefoot strike with the focuses on the biomechanics of the plantar fascia and its correlation to plantar fasciitis. There are many confounding factors that can influence the injury risk of the plantar fascia and they are difficult to be completely accounted for in research. Another handicap is that, noninvasive assessment of loading on deep foot tissues are not easily feasible, nor is real-time measurements of the fascial strains during running gait. Information of the fascial strain is imperative to the understanding of the pathology, prognosis, and treatment effects for plantar fasciitis.

Enthusiastic runners usually adhere to their running regime for either a competition purpose or personal wellbeing. It can be frustrating for them to abstain from training due to injury. Runners having plantar fasciitis can be particularly upset because the pain on the heel could progress from step to step and the symptoms commonly recur in bouts spanning their entire career (Thomas, 2015). Therefore, a treatment method can be easily acceptable to this injured group if it is convenient to access and also wearable during exercise. Fortunately, therapeutic taping is one of those methods.

Taping is widely used to treat plantar fasciitis for its alleged ability to correct the biomechanical faults of the foot and take the strains off the plantar fascia (Podolsky and Kalichman, 2015). Reducing tensions on the fascial band can improve tissue repairment and facilitate recovery from the injury (Tsai et al., 2010). In spite of the variances in taping modality and treatment course, taping is consistently showed to produce positive effects on general gait kinematics, patients' feedback on pain relief, and function restore (van de Water and Speksnijder, 2010). However, these variables do not necessarily reflect a correction to the loading status of the plantar fascia because the causes of symptomatic changes could be multi-factorial. While the functionality of taping should be primarily justified by how much it can offload the plantar fascia, more evidence regarding the fascial strains in the taped foot during running is needed to substantiate the application value of taping treatment.

1.2. Formulation of research question

For runners habitually using rearfoot strike, modifying their foot strike patterns sometimes could be a mix of hardship and risks because forefoot strike running can increasingly engage the plantar foot muscles, ligaments, and the plantar fascia to cope with the mechanical burdens on the foot (Kelly et al., 2018, 2014). Transition to forefoot strike should be built on solid foot strength for better protection of the soft tissues (Kelly et al., 2018). However, this precondition is not always satisfied in runners who change their landing styles too quickly, in which way the plantar fascia is more likely to be overstrained and injured (Tan et al., 2008). Clinicians have raised the concerns of how the plantar fascia may response mechanically to a premature foot strike pattern modification (Daoud et al., 2012; Lieberman et al., 2010), while very few studies so far have compared the differences between rearfoot strikers and forefoot strikers with focuses on the plantar fascia.

The injury risk of plantar fasciitis involved in forefoot strike running is closely related to the loading status of the plantar fascia (Perl et al., 2012). The plantar fascia being increasingly strained by forefoot strike running is a conjecture deduced from a physical model, which describes a scenario that the three-point-bending motion on foot arch can further stretch the plantar foot tissues at the early stance. In spite of the difficulties of measuring loading on the deep foot tissues, the statement is currently supported by very limited evidence.

By far, many studies have examined the effects of taping treatments on plantar fasciitis. Though the reported outcomes are generally positive (Radford et al., 2006a; Salvioli et al., 2017; van de Water and Speksnijder, 2010), the taping modalities adopted by existing studies are many and they normally vary in structures, which makes the research results less generalizable to the whole population. Besides, current measures of the taping treatment effects are mainly patient-rated. Symptom improvement does not necessarily reflect the changes in fascial strain, which is the key factor influencing the risks of plantar fasciitis. More studies should be conducted to quantify how much different taping treatments can offload the plantar fascia.

1.3. Objectives of the study

- To evaluate and compare the mechanical property of plantar fascia in runners using rearfoot strike and forefoot strike techniques.
- To compare the plantar fascia loading between rearfoot strike running and forefoot strike running.
- To investigate the offload effects of taping on the plantar fascia during running.

1.4. Outline of the thesis

Chapter 1 serves as a general overview of the running event, injury of the plantar fascia among runners, and how it can relate to the running styles and taping treatments from a mechanical point of view. The chapter also summarizes the current status of research addressing the relevant points. The last section of this chapter formulates a problem statement that proposes the overall value of the project.

Chapter 2 is the literature review. The first four sections review the fundamental anatomies and functions of the foot and the plantar fascia, the pathomechanics of plantar fasciitis, and the status of running as a general public sports event. The remaining two sections are critical reviews on existing studies investigating the effects of different foot strike techniques and taping methods on the outcomes of running biomechanics, including general gait kinematics, GRF, pedobarography, joint loading, muscle activities, patient's feedback, and scores on functional recovery. The evaluation of these reviews comes up with the state of the art of research, reveals the research gaps, and navigates to the initiative of the up-coming works in the study.

Chapter 3 is a brief method overview. The chapter displays the details of the major measurement tools and simulation platforms used in the study, including the ultrasound shear wave elastography, infare-based motion capture analysis, musculoskeletal modeling, and finite element modeling.

Chapter 4, 5, and 6 present the three separate studies that focus on the biomechanics of the plantar fascia in different experimental conditions that correspond to the research goals. **Chapter 4** examines the shear wave elasticity of the plantar fascia in recreational runners and compares the results between rearfoot strikers and forefoot strikers. **Chapter 5 and Chapter 6** are studies of computational simulations that quantify the tensile force/strain on the plantar fascia during running in different foot strike and taping treatment conditions respectively.

Chapter 7 discusses the features and limitations of the study, draws an overall conclusion to the project, and provides suggestions for possible future research.

CHAPTER 2 LITERATURE REVIEW

2.1. Foot

The human foot is one of the most complex structures of the musculoskeletal system that is constituted of over one hundred parts, including the bones, muscle, tendons, and ligaments. Human foot is designed to sustain body weight and support diverse movements such as walking, jogging, running, jumping, etc. To accomplish a locomotion task, components of the foot coordinate in a fine-tuned form under the central nerve control. Any movements thus-generated are based on a balanced amount of flexibility and stability. Due to this complexity, human foot can be easily vulnerable to injuries. Understanding the anatomy and biomechanics of the foot can shed a light on its injury mechanism and the associated risk factors.

2.1.1. Anatomy

Bones

The bony structures of the foot can be normally divided into three sections: the forefoot, midfoot, and hindfoot (Figure 1).

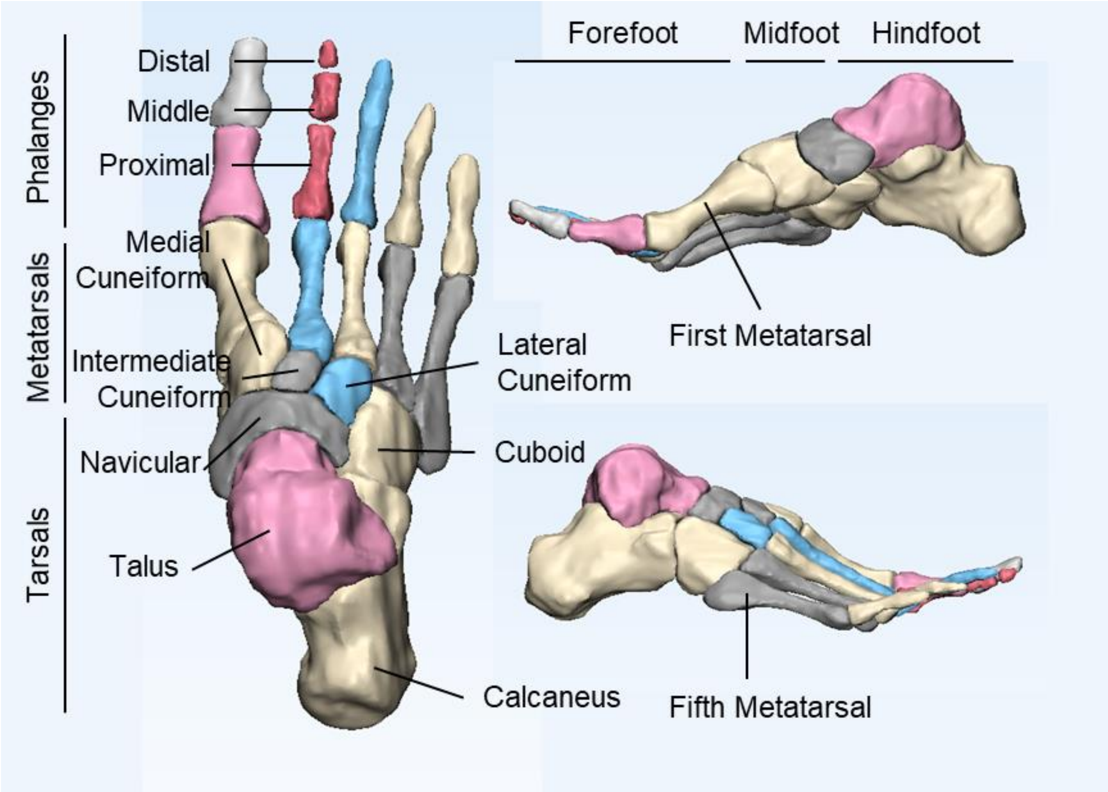


Figure 1. Bony structure of the foot.

The forefoot is the very front part of the foot that includes fourteen phalanges (two for the big toe and three per each of the other four toes) and five metatarsals. The first metatarsal bone is the shortest and thickest of the five metatarsals. It plays the primary role in loading bearing during locomotion (Kernozek et al., 2016). It also provides attachment for several tendons. The second, third, and fourth metatarsal bones are the most stable of the metatarsals. They are well-protected and have only minor tendon attachments. In addition to the phalanges and metatarsals, the forefoot contains two small, oval-shaped sesamoid bones. They usually sit beneath the head of the first metatarsal, on the plantar surface or underside of the foot, and are held in place by tendons and ligaments. The forefoot meets the midfoot at the five tarsometatarsal joints.

The midfoot is the portion of the foot that sits between the hindfoot and forefoot. The midtarsal joint (Chopart joint) joins the hindfoot to the midfoot. The tarsometatarsal joints join the midfoot to the forefoot. The midfoot section is comprised of five irregularly shaped bones that were connected by short ligaments and only allow a small amount of relative movement (Rüedi and Foundation, 2007). They are the medial cuneiform, intermediate cuneiform, lateral cuneiform, navicular, and cuboid. Together they form the arch of the foot, which plays a major role in impact absorption and weight support during movements.

The hindfoot is the most posterior portion of the foot. It begins immediately below the ankle joint and ends at the level of the Chopart joint. Hindfoot section contains two large bones—the talus and the calcaneus. The calcaneus is the largest foot bone that forms the heel. The talus rests on top of the calcaneus and forms the pivoting joint of the ankle: the subtalar joint. The two long bones of the lower leg, the tibia and fibula, are connected to the top of the talus to form the ankle. The calcaneus is cushioned underneath by a layer of fat (Sarrafian, 1993).

Joints

A joint is formed at the junction between two or more bones. The foot contains a total of thirty joints, amongst them three are the most important for rearfoot: the talocrural, subtalar, and transverse tarsal joints. These three joints confer a large degree of sagittal motion and a certain amount of coronal/transversal motion to the heel.

The talocrural articulation, also known as the ankle joint, a synovial joint located in the lower limb that connects the lower limb to proximal talus. Bound together by tibiofibular ligaments, the tibia and fibula form a bracket shaped socket called a mortise at their distal ends. The body of the talus fits tightly into the mortise and constructs a strong bony complex (Bonnel et al., 1998). Talocrural joint is a hinge type joint that permits dorsiflexion of 0–16.5 degrees (Baggett and Young, 1993) and plantarflexion of 0–50 degrees on the ankle (Brockett and Chapman, 2016). The degree of motion could vary upon the weight-bearing condition

(Baggett and Young, 1993). The proximal surface of the talus is wedge-shaped—broader anteriorly and narrower posteriorly. As a matter of fact, the talocrural articulation is more stable when dorsiflexed and has its bony constraints reduced when plantarflexed, wherein the ankle ligaments are more susceptible to strain and injury (Brockett and Chapman, 2016).

The subtalar joint is a posterior foot joint formed between the talus and the calcaneus. The joint features three articulating facets between the two bones and allows inversion/eversion of the foot. The transverse tarsal joint, namely the midtarsal joint or the Chopart's joint, combines two joints. One is the talocalcaneonavicular joint formed between the talus, the calcaneus, and the navicular bones, the other one is the calcaneocuboid joint formed between the front of the calcaneus and the posterior surface of the cuboid bone (Kutaish et al., 2017). The movement which takes place in this joint contributes, in a small part, to the rearfoot inversion/eversion, by means of rotations between the hindfoot and midfoot sections.

Most of the midfoot joints are small joints formed by the five midfoot bones. They only allow for minor motions and contribute, in part, to foot arch deformation (Gray, 2011).

The midfoot section and forefoot section are connected by the tarsometatarsal joints, which are formed between the tarsal bones and the bases of the metatarsals. The metatarsals also articulate each other on their bases to the intermetatarsal joints. Tarsometatarsal joint is also named Lisfranc joint after 19th-century surgeon and gynecologist, Jacques Lisfranc (Gray, 2011). Movements permitted between the tarsal and metatarsal bones are limited to slight gliding of the bones upon each other.

The big toe has one interphalangeal joint and each of the other four toes have two: the proximal interphalangeal joint in the middle of the toe, and the distal phalangeal joint. Each toe connects to the corresponding metatarsal with the metatarsophalangeal joint at its base. All interphalangeal joints are ginglymoid (hinge) joints that flex and extend in the sagittal plane. These movements are more extensive between the first and second phalanges than between the second and third. For interphalangeal joints, the amount of flexion is very considerable, but extension is limited by the plantar and collateral ligaments (Gray, 2011).

Muscles

The muscles that control the movements of the foot mostly originate in the lower leg and are attached the bones in the foot through tendons. The foot muscles can be divided into two major categories based on where their bellies locate: the extrinsic foot muscles and intrinsic foot muscles.

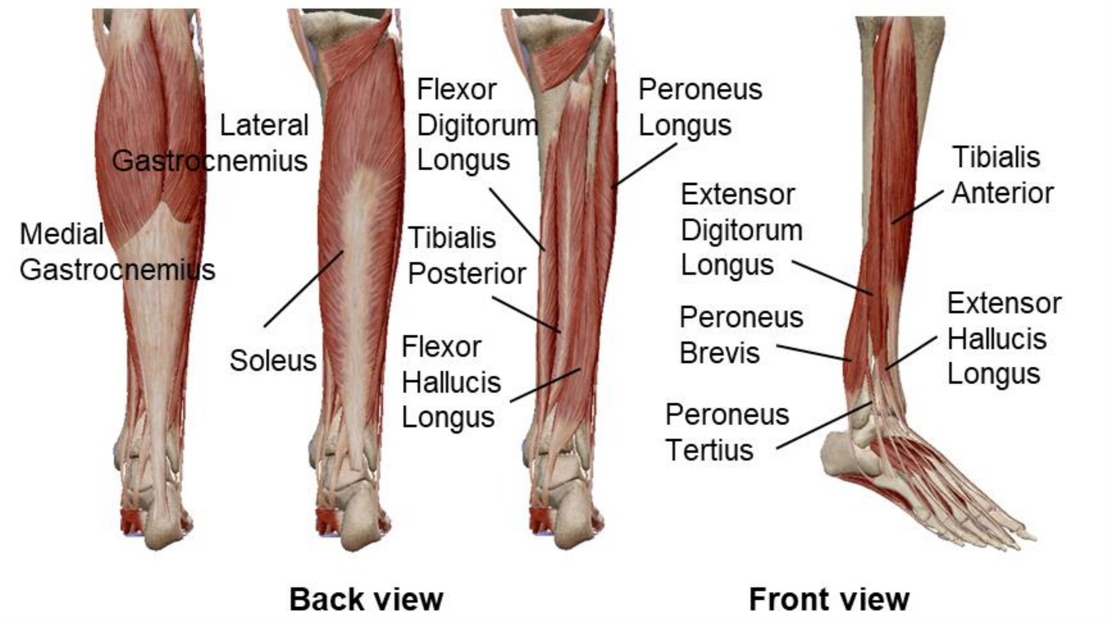


Figure 2. The extrinsic foot muscles.

There are eleven extrinsic foot muscles arising from the anterior, posterior, and lateral compartments of the leg (Figure 2). With the large cross-sectional areas and moment arms to the ankle joint complex, the extrinsic foot muscles are mainly in charge of producing gross joint motion such as ankle dorsiflexion/plantarflexion (Gray, 2011). Extrinsic foot muscles normally have long tendons that insert to the foot and interconnect with the ligamentous network, which confers the ability of adjusting foot structural rigidity and regulating energy cycle to the muscles (McKeon et al., 2015).

In the anterior compartment, there are the anterior tibialis, extensor hallucis longus, extensor digitorum longus, and peroneus tertius. The anterior tibialis is the most medial muscle of the anterior compartment of the leg. It is responsible for dorsiflexing and inverting the foot. The extensor hallucis longus and extensor digitorum longus can extend the phalanges of the first ray and the other four toes respectively. The peroneus tertius can evert the rearfoot and provide week dorsiflexion to the ankle.

In the lateral compartment, there are the peroneus longus and peroneus brevis. The two muscles together can plantarflex the ankle. The peroneus longus also everts the foot through its oblique tendon across the sole. The triceps surae, also known as the calf muscle, forms the major part of the posterior compartment of the leg—the two-headed gastrocnemius and the soleus. These muscles both insert into the calcaneus by joining in the Achilles tendon, and. They are one of the most powerful leg muscles with the primary function of plantarflexing the ankle. The muscle group is heavily involved in static balancing, ambulation, and power jumping. The posterior compartment also includes the posterior tibialis, flexor hallucis longus, and flexor digitorum longus. Their actions are ankle plantarflexion/inversion, first ray flexion, and toes flexion respectively.

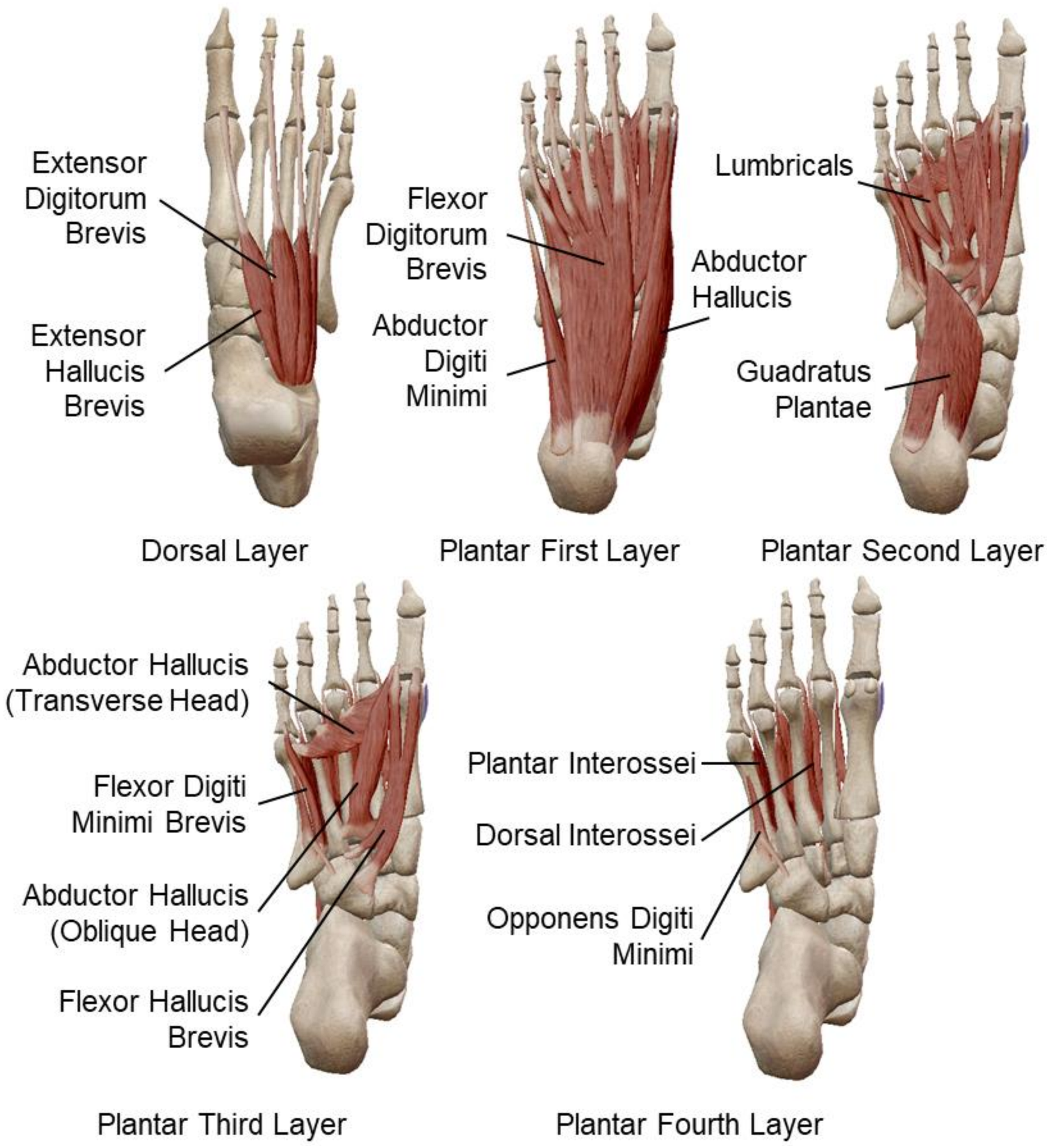


Figure 3. The intrinsic foot muscles.

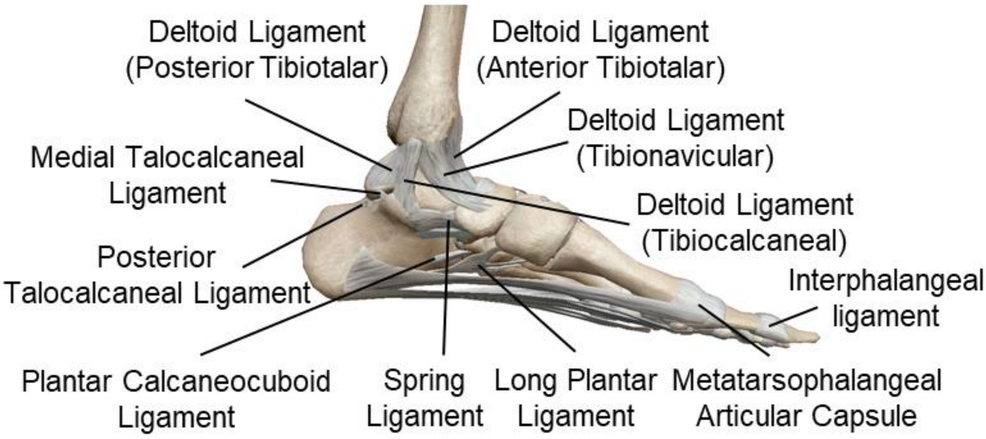
The intrinsic muscles are located within the foot, they are the stabilizers of the foot arch and responsible for the fine motor actions of the foot segments. These muscles generally have small moment arms and small cross-sectional areas (Schünke et al., 2010).

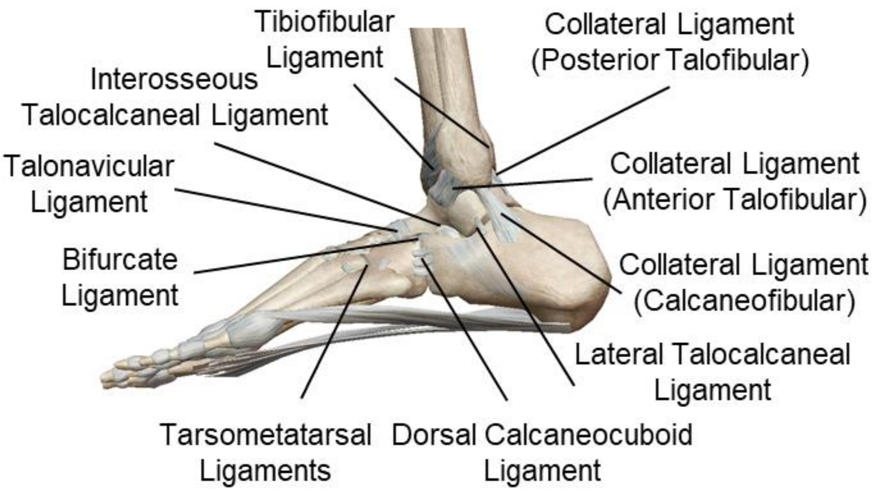
Most of the intrinsic foot muscles located in the sole of the foot while only two located in the dorsal foot: the extensor digitorum brevis and the extensor hallucis brevis (Figure 3). These two muscles are mainly responsible for assisting some of the extrinsic foot muscles for extending the five phalangeal.

Plantar intrinsic foot muscles consist of four layers of muscles deep to the plantar aponeurosis (Figure 3). Muscles of the first two layers align with the medial and lateral longitudinal arches of the foot whereas the deeper layers configure more with the anterior and posterior transverse arches (McKeon et al., 2015). They act collectively to stabilize the foot arch and control the movement of the digits. In the first layer, there are the abductor hallucis, flexor digitorum brevis, and abductor digiti minimi. Flexor digitorum brevis is the large central muscle located immediately below the plantar aponeurosis. It flexes the lateral four digits and is flanked by abductor hallucis and abductor digiti minimi. The second layer contains two muscles: the quadratus plantae and lumbricals. The quadratus plantae assists the flexor digitorum longus in flexing the lateral four toes. The lumbricals can flex the metatarsophalangeal joints while also extend the interphalangeal joints. In the third layer, the oblique head of adductor hallucis joins the muscle's transversal head on the lateral side of the big toe. The muscle assists in forming the transverse foot arch and can adduct the great toe. Medially to the adductor hallucis are the two heads of flexor hallucis brevis, deep to the tendon of flexor hallucis longus. The considerably smaller flexor digiti minimi brevis is located on the lateral side of foot. The two flexors flex the corresponding toes at the metatarsophalangeal joints. In the fourth layer. the dorsal and plantar interossei are located between and below the metatarsal bones and act as antagonists in terms of digits adduction/abduction.

Tendons and Ligaments

The most notable foot tendon is the Achilles tendon, which is a tough band of fibrous tissue running from the calf muscle to the heel. The gastrocnemius and soleus join into one band of tissue at their distal ends to form the Achilles tendon. The Achilles tendon then inserts into the posterior calcaneus. The Achilles tendon is the largest and strongest tendon in the body. When the calf muscles contract, the Achilles tendon pulls on the heel. This movement facilitates running, jumping, climbing stairs, and standing on toes. Other important tendons in the foot include the tibialis posterior (posterior tibial tendon), which attaches the calf muscle to the bones on the inside of the foot and supports the arch of the foot, and the tibialis anterior (anterior tibial tendon) which runs from the outer tibia to the first metatarsal and surfaces of the median cuneiform tarsal, which allows for toe dorsiflexion.





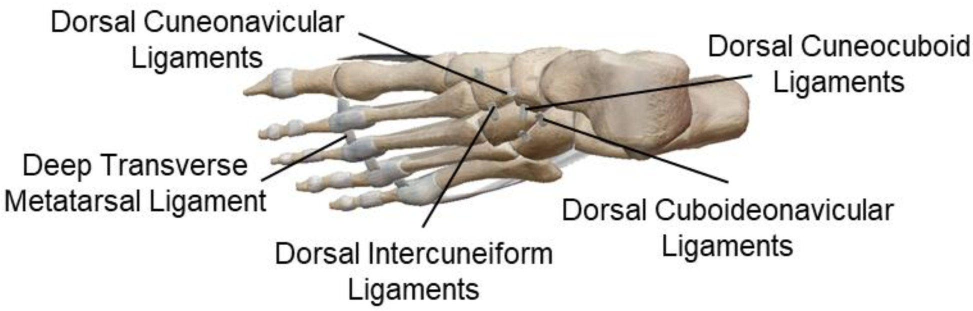


Figure 4. Ligaments of the foot.

Ligaments are soft tissues made of collagen for enhancing bone-to-bone connections. On the lateral side of the ankle there are three major ligaments: the anterior talofibular ligament, posterior talofibular ligament, and calcaneal fibular ligament (Figure 4). The three ligaments primarily serve to restrain subtalar inversion and limit the talar movements in various ankle plantarflexion/dorsiflexion angles. The deltoid ligament and calcaneonavicular ligament are the two major ligaments on the medial side of the ankle. The deltoid ligament is a strong, flat, triangular band composed of three independent ligaments. It attaches to the apex, anterior, and posterior borders of the medial malleolus and restraint the subtalar eversion. Together with the ligaments on the lateral ankle, they are the major ankle joint stabilizers. The spring ligament, namely the plantar calcaneonavicular ligament, is a broad and thick band with three constituent ligaments that connect the anterior calcaneus to the plantar surface of the navicular. The ligament complex maintains the medial longitudinal foot arch and bears most of the body weight in a normally functioning foot. Other major foot ligaments include the long plantar ligament, which goes under the foot and connects the calcaneus with the cuboid bone, and the plantar calcaneocuboid, which stretches from the calcaneus to the cuboid bone.

The plantar fascia is another major ligamental tissue in the foot. It is a thick connective tissue that runs from the plantar calcaneus or heel bone to the metatarsal heads at the base of the toes. The plantar fascia is the longest foot ligament that adds to the strength of the foot arch and assists in energy absorption during locomotion.

2.1.2. Function during running

The human foot is an important anatomical part of the body with one major functionality: assist ambulation. Ambulation can be manifested in many forms such as walking, jogging, running, etc. During ambulation, the body is propelled forward, backward, or in multiple directions. The procedure can invoke a number of low limb locomotors and cause movements of many foot segments, whereof the combined efforts are to damp the impact shock in contact to the ground and produce sufficient power to propel the body.

Though the nature of ambulation is uniquely complicated, it can usually be characterized by repeats of a representative gait cycle. A typical gait cycle contains two bipedal phases—the stance phase and the swing phase. The foot remains in contact with the ground during the stance phase and is aerial during the swing phase. A gait cycle starts from the foot strike to the ground and ends at the next strike. For a normal walking gait, the stance phase usually accounts for approximately 60% of the step cycle and the swing phase accounts for the rest 40% (Figure 5). As the speed of gait increases from walking (around 1.32 m/s) to running (around 4.77 m/s), the duration of the stance phase decreases from 0.62 to 0.14 seconds (Mann et al., 1986). Compared to swing phase, stance phase receives much research interests because during this period, the foot experiences a drastic change of loading in contact to the ground, regulates large force transmission, and contributes to the whole-body momentum. Depending on the functions that the foot fulfills, stance phase can be subdivided into three distinct periods: initial contact, midstance, and toe-off (Figure 5).

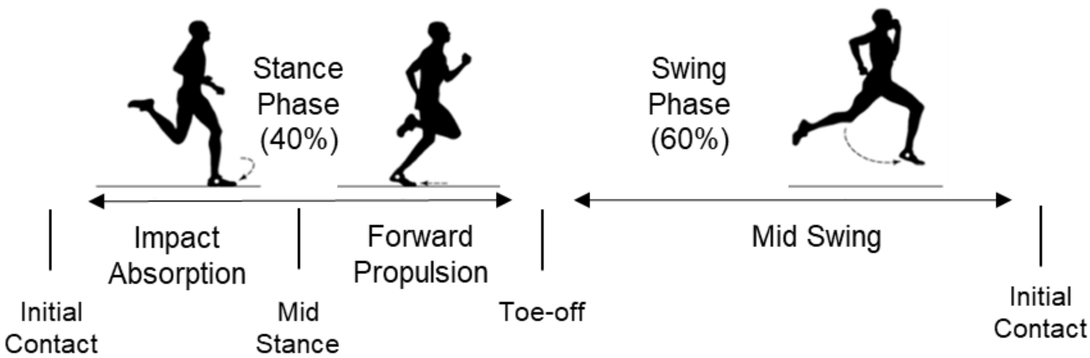


Figure 5. The running gait cycle.

Initial contact refers to the moment in which the foot firstly touches the ground and denotes the beginning of the stance phase. Approximately 80% of runners land on the heel while some land on the midfoot or forefoot (de Almeida et al., 2015). At initial contact the foot functions to both absorb shock as well as rapidly secure a stable lower limb position for bodyweight acceptance. From initial contact to early stance, the rearfoot moves from a supinated position to a pronated position (foot rolling inward), the foot arch gradually collapses, and the hip/knee joint flexes (Hageman et al., 2011; Lee et al., 2010). In combination with the work done by muscle contraction, these adjustments help in attenuating the impacts of foot strike and converting a part of the kinetic energy into elastic energy stored in the soft tissues. The plantar foot also makes full contact to the terrain surface during this period.

Midstance is the phase of gait where the foot assumes more of a support and overall stability role. The complete sole of the foot is weight-bearing. During this phase the foot is in transferring from becoming a shock absorber to a body propeller. Knee flexion peaks at the same time as the rearfoot is maximally pronated (De Wit et al., 2000; Hardin et al., 2004). The body mass center is moving forward over the fixpoint of the supporting leg on the ground and as it does so the foot commences a change toward propulsion.

Toe-off is the final stage of the stance phase. It begins immediately as the heel lifts off the ground. During this period, the rearfoot begins to supinate, allowing the collapsed foot arch to recoil, releasing the stored energy, and turning the midfoot into a rigid level, on which the calf muscles can act to produce propulsion (Albertus-Kajee et al., 2011). As the bodyweight shifts on the big toe, the ankle starts to plantarflex, the knee and hip joints extend to straighten the whole leg.

Kinematics

For typical rearfoot strike running, contact occurs in the posterior third of the foot. As the foot is loaded, the ankle joint is rapidly plantarflexed and the calcaneus everts simultaneously, which results in pronation of the rearfoot and increased motion in the transverse tarsal joint (Svoboda et al., 2016). Motions of the foot joints during running normally fall within a certain range, though substantial variations among individuals may exist as confounding factors, such as running style, footwear (McPoil, 2000), and surface condition, are involved.

For an average runner running at the speed of 4.4 m/s, the calcaneus is inverted by approximately 6 degrees (4–21 degrees) at touchdown and undergoes quick eversion shortly after that. In the meantime, the rearfoot is increasingly pronated until reaching its peak angular value of 12 degrees (4–25 degrees) at midstance (De Wit et al., 2000; Stacoff et al., 2000). The tibia is also internally rotated (to 5–9 degrees) during this period (Bergstra et al., 2015; Bonacci et al., 2013). This rearfoot pronation is one of the mechanisms that assist in shock absorption during early stance. Runners with pes cavus (high foot arch) generally absorb forces more poorly than do those with pes planus (flat foot), because foot pronation was found to reduce in pes cavus foot (Williams et al., 2001).

After the midstance of running, the rearfoot begins to supinate. The subtalar joint goes through inversion to return to the neutral position or a slightly inverted position at the end of stance. Making the total subtalar excursion of 10–12 degrees when runners wear a typical neutral/cushioning shoe (Dierks et al., 2010). At the beginning of the second half stance, the foot is fully loaded as the center of mass passes the base of the supporting leg along the direction of motion. External rotation of the lower leg causes inversion of the calcaneus through the close kinematic chain (Donatelli, 1996). As the foot rising up onto the metatarsal heads at midstance, the obliquity of the metatarsal break—the axis passing the lateral four metatarsophalangeal (MTP) joints in the transverse plane, also helps to supinate the foot by enhancing external rotation of the tibia (Hsu et al., 2008). The foot arch is stabilized by the plantar connective tissues, tightened by MTP joint dorsiflexion (Bolgla and Malone, 2004), and turns itself into a rigid lever, on which the foot muscles can act and facilitate propelling the body (Vogler and Bojsen-Møller, 2000). The external rotation of the lower leg is initiated by the forward swing of the opposite leg that brings the pelvis forward. Because the femur of the stance leg is fixed to the pelvis by the adductors, the femur rotates externally (Dugan and Bhat, 2005). This rotation is passed through the knee/ankle joint and reaches to the foot.

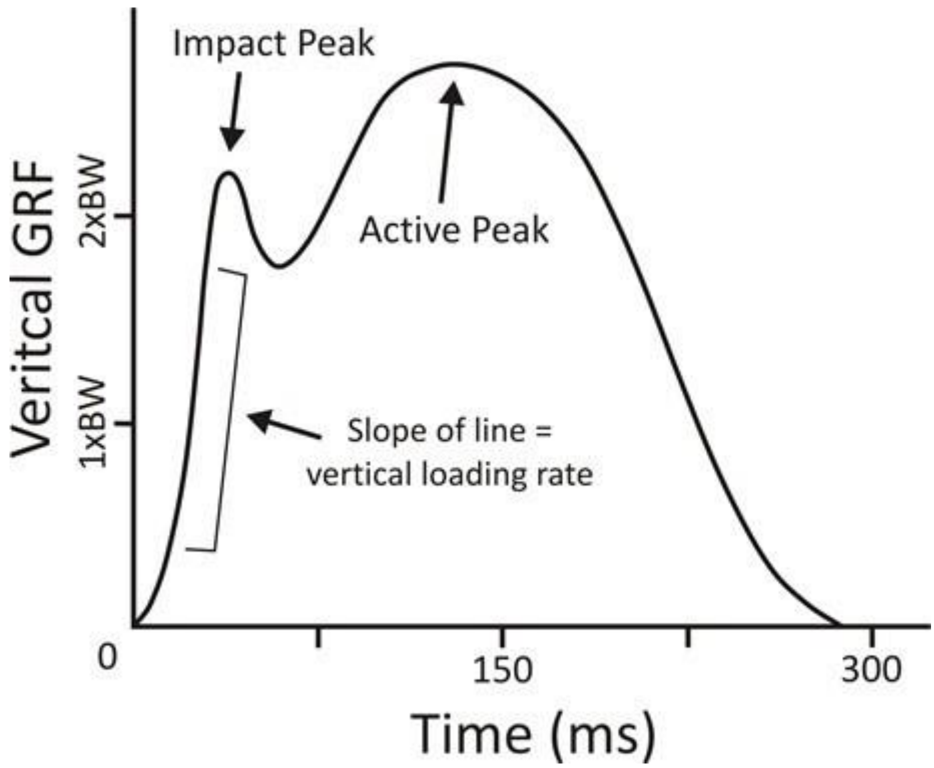


Figure 6. Representative vertical GRF from stance phase of typical rearfoot strike running. (Reprinted from B. Bazuelo-Ruiz et al., 2018, Peer J, 6, e4489, Copyright 2018 by Bazuelo-Ruiz et al).

The "impact peak" (the first and smaller peak of the graph) represents the forces absorbed at initial contact. The "active peak" (the second and larger peak of the graph) represents the forces absorbed at midstance.

Kinetics

During running, vast force is generated by the collision between the foot and the ground. The magnitude of vertical ground reaction force can range from 2.4 to 3.0 times body-weight (BW) depending on a variety of influential factors (Aguinaldo et al., 2002; Clarke et al., 1983; Hamill et al., 1983). Localized loading may be as high as 5–6 times BW at the ankle (Chen et al., 2016; Rooney and Derrick, 2013; Sasimontonkul et al., 2007), 11–15 times at the knee (Edwards et al., 2008; Rooney and Derrick, 2013), and 3–10 times at the Achilles tendon (Almonroeder et al., 2013; Aniţaş and Lucaciu, 2013; Burdett, 1982; Gruber et al., 2011). In rearfoot strike running, a typical plot of the vertical ground reaction force usually shows a spike preceding a single peak (Hamill and Gruber, 2017). The spike indicates an impact peak caused by initial contact/foot strike and its magnitude is usually smaller than the active peak at midstance (Figure 6). The actual peak is considered to relate to/associated with propulsion at the beginning of the late stance (Hunter et al., 2005). With midfoot or forefoot strikers, the impact peak generally flattens out and dissipates (Williams et al., 2000).

Ground reaction force in the shear directions during running is usually less characteristic, most likely because of wide variations in anatomic alignments and foot placements among runners. the anteroposterior ground reaction force peaks at approximately half of the BW during the first part of stance and contributes to a braking force to the movement. A second peak, similar in the magnitude but in the opposite direction, appears after midstance and is associated with forwarding propulsion (Hunter et al., 2005). A small amount of medial shear ground reaction force usually occurs in the second half stance phase, which may also relate to body propulsion due to the external rotation of the stance leg at push-off (Heise and Martin, 2001).

The center of plantar pressure is another variable that reflects kinetics in running gait. A rearfoot striker normally has the center of plantar pressure located on the lateral heel at touchdown. The pressure center then moves anteriorly and medially through the rest of the stance. Map of plantar pressure high-density in the proximal portion of the foot because the forces are concentrated in this region for the first two-thirds of the stance phase (Mann et al., 2016; Tessutti et al., 2010). For the forefoot striker, the central or lateral forefoot region is first loaded at the beginning of the stance. The pressure center then moves posteriorly through loading response and progresses to more anterior and medial portions of the foot during push off (Kernozek et al., 2016).

Muscle activity

Two active groups of the extrinsic foot muscles are important for running—the anterior compartment muscles and the posterior compartment muscles. For rearfoot strike running, the anterior compartment muscles are active during the early stance phase. They contract concentrically to dorsiflex the ankle and secure the ankle stability for shock absorption (Landreneau et al., 2014; Shih et al., 2013; Yong et al., 2014). The mechanism also contributes to tibia deceleration over the fixed foot. For forefoot strikers, a period of inactivity of the anterior compartment muscles at pre-stance is associated with plantar flexion of the foot. The posterior calf muscles become active immediately at foot strike, they undergo an eccentric contraction, which controls forward movement of the tibia and stores elastic energy over the ankle and soft tissues themselves (Almonroeder et al., 2013; Kulmala et al., 2013; Rice and Patel, 2017). In addition to that, forefoot strike running also showed an apparent co-contraction of the anterior and the posterior compartment muscles at which is thought to provide the muscular stability for the plantarflexed ankle (Hashizume and Yanagiya, 2017). For both running styles, the posterior compartment muscles generate a large forward momentum during the late stance phase and contribute the most to push-off.

Intrinsic foot muscles are also widely recruited during running to facilitate the movement.

Unlike the extrinsic foot muscles that actuate joint motions, the intrinsic foot muscles are the major stabilizers of the foot and the foot arch. They also aid in the energy cycle during gait. At the stance phase, most of the intrinsic foot muscles firstly undergo a slow process of active-lengthening to cater to foot arch compression and then shorten rapidly as the arch recoil for propulsion (Kelly et al., 2015). Likewise, the magnitude of muscle lengthening and muscle activation is found to change upon varying running conditions (Masumoto et al., 2017; Nishida et al., 2017; Vernillo et al., 2017; Voloshina and Ferris, 2015).

The foot arches

The foot arch is the bony arch formed by the tarsal and metatarsal bones and strengthened by the ligaments and tendons. The complex structure of the foot arch confers it with two important functions: weight-bearing and propulsion in locomotion. To achieve these functions, the foot arch should possess a high degree of both stability and flexibility. A great part of the foot bones was articulated in an arch array that allows it to support large loading over the top, while the joints, mostly with a small range of motion under the constraints of the soft tissues, are able to deform under loading conditions to fulfill its dynamic functionality.

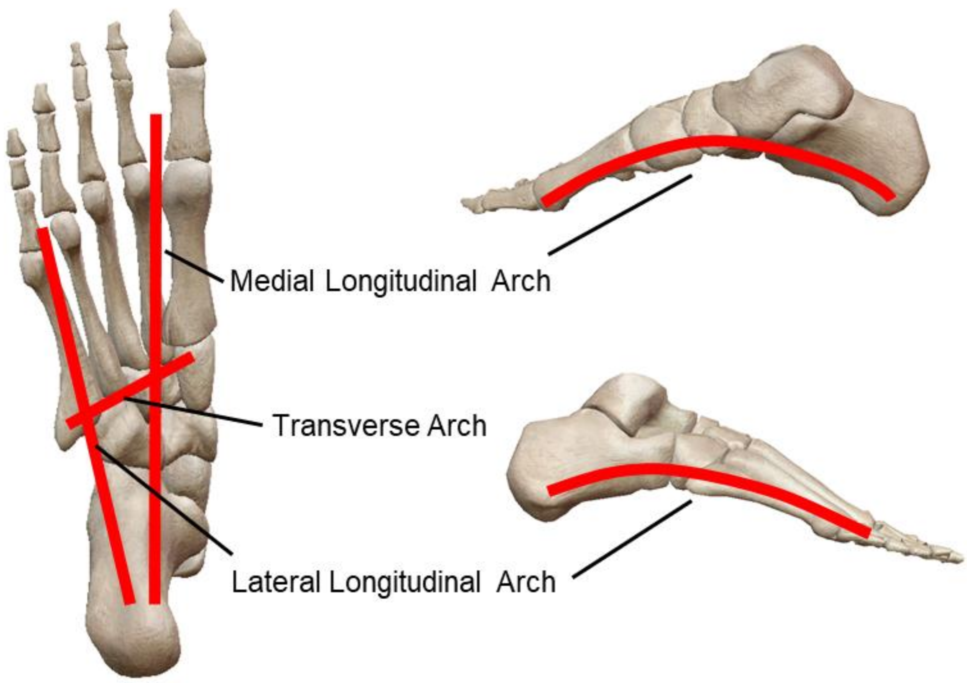


Figure 7. Arches of the foot.

The foot arch has three distinct arches—two longitudinal arches, stretching from the heel to the forefoot and one transverse arch across the midfoot on the coronal plane (Figure 7).

The MLA is the most prominent foot arch that is usually referred to as “the arch”. Its summit is at the superior articular surface of the talus, and its two extremities, on which it rests in standing, are the tuberosity on the plantar surface of the calcaneus posteriorly and the heads of the first, second, and third metatarsal bones anteriorly. The medial arch is formed by the calcaneus, talus, navicular, three cuneiforms, and the medial three metatarsal bones. This arch absorbs most of the impact shock during walking, running, and jumping. Muscles supporting the MLA are the tibialis anterior and posterior, peroneus longus, flexor digitorum longus, flexor hallucis, and the majority of the intrinsic foot muscles. Some plantar ligaments, in particular the long plantar and plantar calcaneonavicular ligaments, together with the plantar fascia, help to consolidate the MLA.

The lateral longitudinal foot arch runs along the lateral edge of the foot and parallel to the MLA. The lateral arch is usually flat on the ground in the static standing position and can be more clearly visualized in the high-arch foot (Chang et al., 2010). It is formed by the calcaneus, cuboid, and the lateral two metatarsal bones. Most of the soft tissues that support the medial arch also play a role in stabilizing the lateral arch.

The transverse arch is formed by the metatarsal bases, the cuboid, and the three cuneiforms on the coronal plane. Except for the peroneus longus and tibialis posterior, ligaments that connect two or more of the above bones also support the transverse arch.

Among the three foot arches, the MLA contributes the most to the foot functionality. The medial arch creates a large space in which most of the plantar connective tissues, e.g. the plantar fascia, pass through. They can serve as springs that are fixed to the two arch piers and counteract collapse of the arch when it is loaded. These soft tissues can also spread ground contact reaction forces over a longer time period and thus reduce wears or damage to the musculoskeletal system. Due to the material elasticity, the medial arch, along with the plantar soft tissues, can store energy to attenuate the impact of landing and return the energy for body propulsion. This function is particularity effective in saving the costs of walking and running.

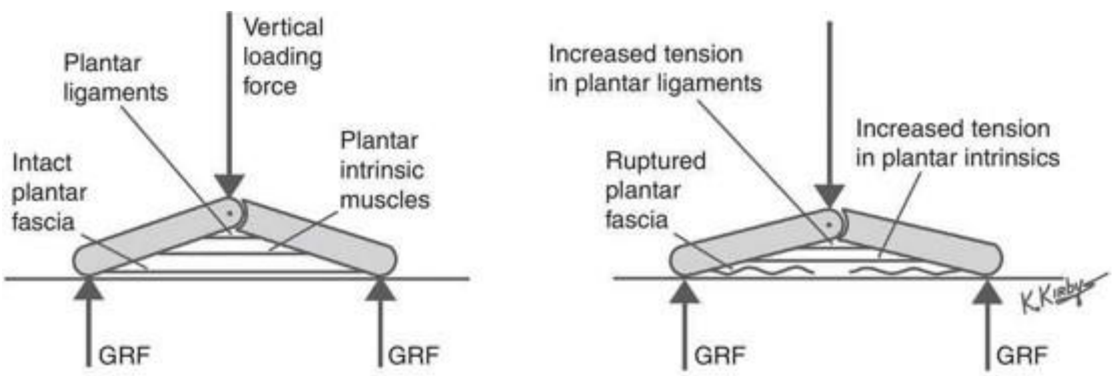


Figure 8. The longitudinal arch load-sharing system. (Reprinted from K. Kirby, 2017, Revista Española de Podología, 28(1), pp. e18–e26, Copyright 2017 by Elsevier Ltd.).

Load-sharing system of the foot arch

Load-sharing is a concept commonly used in the field of electrical mechanics that describes the intricate mechanical balance of forces acting on the components of a system. Examples of load-sharing include the use of multiple supporting cables in suspension bridges, multiple engines in aircraft, and multiple electric generators in power-generation systems. The concept was borrowed from the industry and used for the human foot arch by Kirby in 2002 (Kirby, 2002). In a load-sharing system, the total load applied to the system is distributed among the components, equally or unequally, depending on the integrity of each component (Figure 8). Normally, the amount of the components should be redundant to secure the reliability of the system. In case that if one component fails, the loads acting on the remain will increase to keep the system operational.

The longitudinal foot arch is subject to large external loads during weight-bearing activities. The arch system should be able to resist these loads with degrees of stiffness. In a single running gait cycle, the longitudinal arch was early compressed by the combined actions of multiple forces and undergoes elongation for weight acceptance. When the body center passes the stance foot and prepares to forward. The ground reaction force is gradually reduced and finally removed from the sole of the foot. The longitudinal arch regains its normal height and restores its unload shape. This load-unload cycle could occur thousands of times on a daily basis. In order to fulfill this spring-like function enduringly, the longitudinal arch should have multiple components that allow both passive and active control of its mechanical behavior.

Passive components of the foot arch load-sharing system

Passive components of the foot arch load-sharing system include the bones, joints, plantar fascia, and plantar ligaments. The osseous elements of the longitudinal arch are constructed to resist compression over the arch summit. The rearfoot section, consisting of the talus and calcaneus, and the forefoot section, consisting of the navicular, cuboid, cuneiforms, and metatarsals, are able to dorsiflex/plantarflex relative to each other. Together with the small mobility of the midfoot joints, the longitudinal arch is able to flatten or elevate, on the bases that the soft tissues connecting the bones provide the sufficient strength to stabilize the arch. The plantar fascia and plantar ligaments are the passively stretchable components. Since they are not governed by the central nervous system, they yield tension only when they undergo elongation. The plantar fascia alone constitutes the most superficial layer of the arch load-sharing system. Research showed that plantar fascia could elevate the foot arch when it is tensioned by hallux dorsiflexion The mechanism later on became well known as the "windlass effect" (Hicks, 1954). Tension force on the plantar fascia was estimated to be 0.96 times BW in simulated walking experiments (Erdemir et al., 2004). The plantar ligaments are the deepest layer of the arch load-sharing system. Similar to the plantar fascia, the plantar ligaments exert tension force when the fibers are elongated by dorsiflexion of the forefoot. Research has demonstrated that, in the loaded cadaveric feet, the average strain within the spring and long plantar ligaments increased significantly following the plantar fasciotomy (Crary et al., 2003). The findings indicated that the plantar fascia and plantar ligaments work synergistically to compose the longitudinal arch stiffness.

Active components of the load-sharing system

Active components of the foot arch load-sharing system are constituted of extrinsic and intrinsic foot muscles. The foot muscles are under the control of the central nervous system. They produce tensile force to constrain arch deformation through the active contraction.

The extrinsic foot muscles originate in the lower leg, cross the ankle, and insert on the foot. They are the global movers that generate gross joint motion (McKeon et al., 2015). Four extrinsic muscles are particularly functional in stabilizing the foot arch: the posterior tibialis, flexor digitorum longus, flexor hallucis longus, and peroneus longus. The posterior tibialis and peroneus longus have their insertion to the plantar midfoot areas. During contractile activities, they provide support to the longitudinal arch by exerting a forefoot plantarflexion moment (a torque that resists arch collapse). The flexor digitorum longus and flexor hallucis longus run across nearly the full plantar arch and insert into the bases of the distal phalanges. By active contraction, they produce a proximally directed compression force at the metatarsophalangeal joints and plantarflex the forefoot.

The intrinsic foot muscles originate and insert on the foot. Unlike the extrinsic foot muscles, the intrinsic foot muscles are more of the local stabilizers of the foot and the foot arch. Their functions are well documented in research tracking corresponding bony movements that are triggered by individual muscle contraction (Soysa et al., 2012). Previous studies observing electromyography (EMG) signals of several intrinsic muscles reported an overall enhanced contractile activity in the abductor hallucis, flexor digitorum brevis, and quadratus plantae, in response to increased loading on the longitudinal arch during walking and running (Kelly et al., 2015, 2012). Additionally, simulating these muscles elevated the foot arch (Kelly et al., 2014). Depending on the type and intensity of specific weight-bearing activities, the central nervous system adjusts the magnitude and temporal patterns of the motor activities for some or all foot muscles, either extrinsic or intrinsic, to regulate the arch stiffness (Kirby, 2002).

Synergy of the components

During the first half of the stance phase in running, the supporting lower leg experiences a large deceleration preceding initial contact. The longitudinal arch must allow a certain degree of flexibility to dampen or attenuate the impact shock. For the second half of the stance phase, the deformed foot arch recoils to return the elastic energy for propulsion. During this period, the foot arch should be able to retain its stiffness to form a rigid lever on which the foot muscles can act to yield the driving force. To turn the foot arch into an effective weight-bearing structure, alterations in the longitudinal arch must be regulated from time to time through the synergy of the passive and active components. The myofascial linkage of the Achilles tendon, plantar fascia, and the plantar ligaments confers an auto-stiffening mechanism on the longitudinal arch (Kirby, 2017). As the calf muscles increase their outputs during the progression of the stance phase, the Achilles tendon is tensioned to create a plantarflexion moment on the rearfoot. A plantarflexed rearfoot relative to the forefoot will tighten the plantar fascia and the plantar ligaments until that, they generate sufficient dorsiflexion moments to counterbalance the rearfoot plantarflexion. At this point, the longitudinal arch ceases flattening and becomes stiffened. While this auto-stiffness mechanism is passively activated by arch deformation, the central nervous system can further adjust the arch stiffness through contractile activities of the foot muscles. To accommodate a variety of surface conditions and fulfill the task requirements of different stance phases, the central nervous system actively controls the muscular outputs to create an either more solid or more compliant foot arch.

Overall, the foundation of the load-sharing system is that each of the system components performs similar functions to support the arch. They can act independently while also cooperate to optimize the functionality of the foot arch. If one element fails, the remaining elements can compensate to maintain the arch function. However, the failure of one element also increases the tension loads on others.

2.2. Plantar fascia

2.2.1. Anatomy

The plantar fascia is a connective tissue of unevenly thicken fibrous sheet that starts from the medial tubercle on the inferior calcaneus, fans out distally, and attaches to the plantar facets of the MTP joints (Roxas, 2005). A fully developed plantar fascia can be structurally divided into three components: the medial, central, and lateral bands (Hedrick, 1996). The central band is the main body of the plantar fascia that possesses its major functionality (Figure 9), while the appearance of medial and lateral bands is rather variable among individuals (Dylevský, 1988). Research has reported a wide range of structural variety for the two sideway bands from apparent thickness to complete absence (Wearing et al., 2006b).

The plantar fascia is triangular in shape that constitutes the most superficial layer of the plantar soft tissues. At its many attachments onto the bases of the digits, plantar fascia diverges into many fibro bundles that can cross each other and insert into adjacent soft tissues such as flexor tendon sheath, interosseous fascia, the fascia of the transverse head of the adductor hallucis, and the deep transverse metatarsal ligament (Sarrafian, 1993). From proximal to distal, the plantar fascia is also continuing to intermuscular longitudinal septa through its medial and lateral aspects of the central band. The broad extension of the fascial tissue and its connection to many adjacent anatomic regions increase its structural stability and underlie its role of substantiating the foot-ankle mechanical chain in human locomotion (Wearing et al., 2006b).

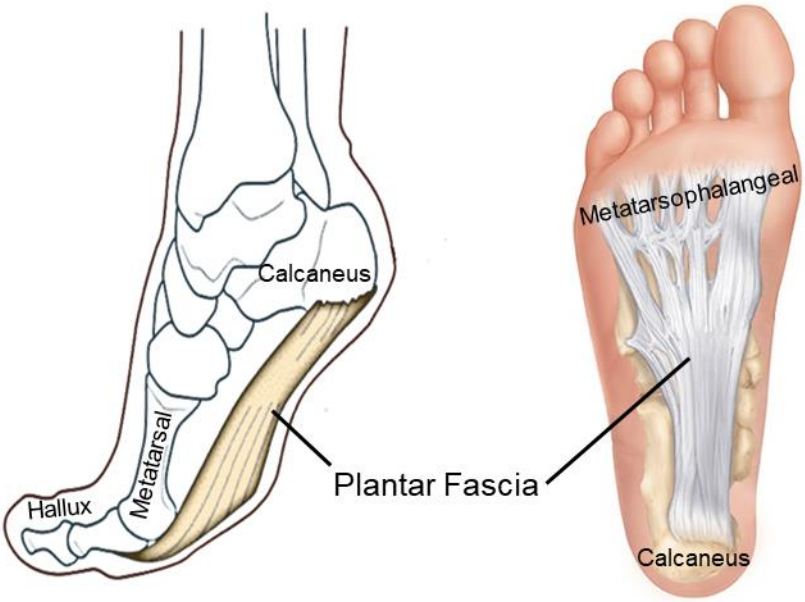


Figure 9. Plantar fascia.

2.2.2. Mechanical property

The plantar fascia is geometrically asymmetric and varies in thickness and width from end to end, which results in an altered mechanical property on different sites. At its origin approximating the calcaneus tubercle, the thickness and width of the central band can range from 3 to 4 mm and 12 to 29 mm respectively (Cardinal et al., 1996; Thordarson et al., 1997; Tsai, 2000). These dimensions generally reduce as the band fans out and diverges distally to the forefoot region. In line with these geometrical characteristics, most of the research interests reside in the proximal insertion of the plantar aponeurosis because plantar fasciitis commonly occurs at this site (McPoil et al., 2008).

Existing studies investigating the property of plantar fascia usually employed a quasi-static approach (Wearing et al., 2006b). Due to the complexity of the microfiber structures, plantar fascia has the time-dependent viscoelasticity that is slightly different from those of tendons and ligaments (Sakalauskaite and Satkunskienė, 2012). Research has reported an average stiffness and ultimate strength of 209 ± 52 N/mm and 1189 ± 244 N for the plantar aponeurosis (Blevins et al., 1994; Kitaoka et al., 1994), which is similar to that of the anterior cruciate ligament but larger than most of the foot intrinsic ligaments (Harner et al., 1995). This structural property of the plantar fascia, though, is subject to many research factors, such as specimen modality, loading rate, and force distribution (Wearing et al., 2006b).

In contrast to structural property, material property of the plantar fascia is independent of tissue shape/geometry. Instead, it is mainly determined by the complex interaction between the water content and the composition of collagen fibers. The elastic modulus of plantar fascia was reported to be 342–822 Mpa. The values fall within the normal range for human connective soft tissues (Boabighi et al., 1993). Likewise, the elastic modulus of the facial tissue is also influenced by loading rate and is suspected to be underestimated under small testing loads within the 'toe' region of the stress-strain curve (Wearing et al., 2006b). Besides, the differences between *in-vitro* and *in-vivo* studies regarding tissue property should also be noted. Factors associated with the physiological conditions of the donor play an important role in modifying the testing results. (Wearing et al., 2006b).

2.2.3. Mechanical function

Static arch stability

The plantar fascia is the major part of the passive components that support the longitudinal arch and constrain the arch deformation during static standing. A number of cadaveric studies invariably shown that the foot arch, with the plantar fascia removed by sectioning, was flattened (Huang et al., 1993; Murphy et al., 1998; Thordarson et al., 1998). In an intact foot, an external force applied to the apex of the arch could lengthen the foot and increase the fascial tension. Research using a finite element method reported a peak tensile force of 285–352 N on the plantar fascia during standing (Wu et al., 2007). The force was reported to account for 15% of the total arch loading (Arangio et al., 1997; Kim and Voloshin, 1995).

Gait support

The plantar fascia, along with other passively stretchable components, confers two major functions to the longitudinal foot arch in gait—energy storage and force transmission (Davis et al., 2017). As the longitudinal arch flattens during early stance of running, the plantar fascia and the plantar ligaments are strained by 6–17% for bodyweight acceptance (Ker et al., 1987; Stearne et al., 2016). A cadaveric study found a peak plantar fascia loading of 538 N in walking (Erdemir et al., 2004). Computational simulations that accounted for foot muscular forces also reported a peak tension of 464 – 922N on the plantar fascia for running gait (Chen et al., 2014; Lin et al., 2014). The strained plantar fascia implies that the deformed foot arch converts, in a part, the impact of ground reaction force into elastic potential energy. The stored elastic energy could be unleashed to propel the body later during the stance when the arch recoils. In transferring from weight acceptance to propulsion, plantar fascia is usually the most tensioned. It locks the small joints of the foot and turns the longitudinal arch into a rigid lever, which the foot muscles can anchor and generate large plantarflexion moment to accelerate the body (Chan and Rudins, 1994). Although the osseous structure forms the basic stability for the foot arch, the plantar fascia, as well as the plantar ligaments, can add on to its rigidity to facilitate the force transmission from active muscle contraction to forward momentum.

The windlass mechanism

The “windlass mechanism” was firstly introduced by Hick in his landmark study (Hicks, 1954). The concept likened the mechanical effects of digits dorsiflexion to the function of a windlass. Stretching from its origin at the calcaneus. the plantar fascia fans out to five branches wrapping the corresponding metatarsal heads and attached to the phalanxes. As the toes dorsiflexed, the plantar fascia rolls around the metatarsophalangeal joint and glides towards to the toe tip. The motion shortens the effective length of the plantar fascia beneath the longitudinal arch and tightens the fascial band (Figure 10). For an unload condition, the mechanism causes midfoot plantarflexion and directly raises the arch. The statement is supported by cadaveric studies, which showed the lapsed windlass mechanism following fasciotomy (Thordarson et al., 1997) and an increased arch height with digit dorsiflexion in the intact feet (Thordarson et al., 1995). In-vivo studies also found similar results. Individuals with healthy foot elevated their foot arch following passive dorsiflexion of the great toe (Kappel-Bargas et al., 1998). For a condition in which the foot arch is loaded, the integrity of the windlass mechanism can be compromised by excessive compressive force. During static standing, the total arch loading is bearable to the plantar soft tissues, whereupon the windlass mechanism is still successful through segmental supination and external rotation of the foot (Wearing et al., 2006b). However, previous studies frequently reported flattening of the longitudinal arch during the push-off phase of walking gait, in which the plantar fascia was thought to reach the point of maximal strain and fail to yield sufficient strength to counteract arch collapse (Hunt et al., 2001; Wearing et al., 2004). Outcomes from computational simulations demonstrated that the fascial loading was influenced by both MPT joint angulation and Achilles tendon tension (Carlson et al., 2000; Cheung et al., 2006). The concurrence of calf muscle contraction and toe dorsiflexion during push-off can amplify loading on the fascial band. Once the plantar fascia cannot withstand this tensile force and undergoes elongation, the windlass mechanism is forced disabled.

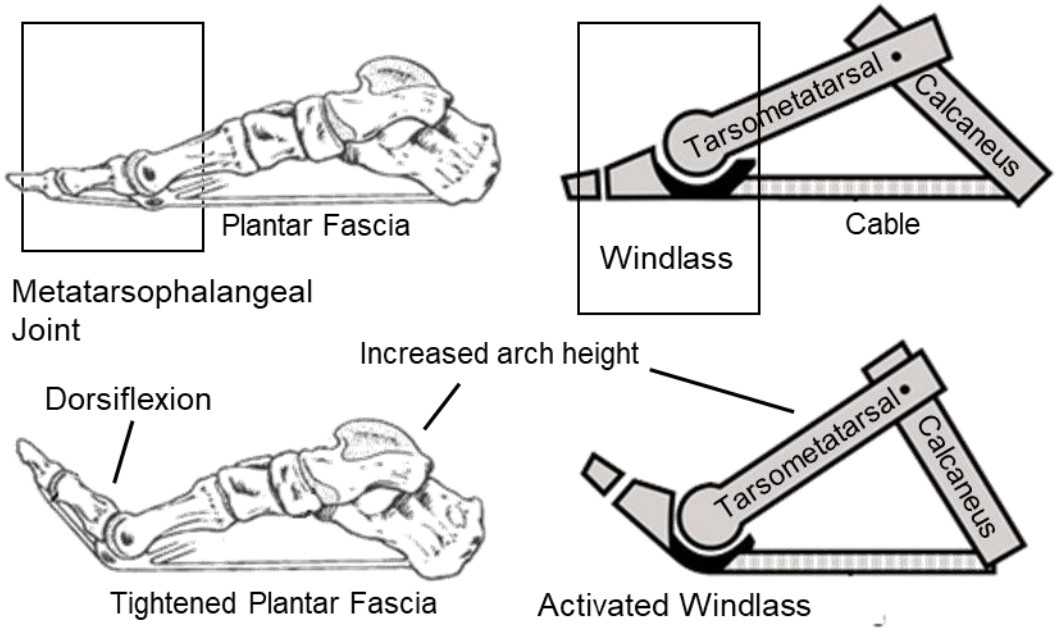


Figure 10. The windlass mechanism of the foot.

2.3. Plantar fasciitis

2.3.1. General information

Plantar fasciitis is one of the most commonly diagnosed foot disorders presenting to foot and ankle specialists. Plantar fasciitis is frequently referred to in the literature by many terms including plantar fasciosis, heel spur syndrome, and runner's heel, which oftentimes confuses the injured people and lends the complexity to its pathology (Roxas, 2005). Though not fully understood, the causes of plantar fasciitis are thought to be largely mechanic-based (Thomas et al., 2010). Repetitive overload on the plantar fascia and its enthesis on the calcaneal tuberosity can induce continuous micro tears and repair procedures, which subsequently contribute to the aggravation of plantar fasciitis in the given of time (Lake, 2000).

The typical symptom of plantar fasciitis is intense, sharp heel pain occurring during the first couple of steps in the morning. The pain is primarily at the origin of the plantar fascia where it is attached to the calcaneus and can radiate distally in more severe cases (Figure 11). Some patients with moderate symptoms may experience this type of morning pain after prolonged walking or standing. Activities such as sprinting and jumping can particularly exacerbate the pain. Footwear with little arch support also contributes to symptom aggravation. In addition to heel pain, patients may report concomitant foot stiffness and localized swelling in the heel, which, in combination, could impair body function and cause a cut-off from routine exercises.

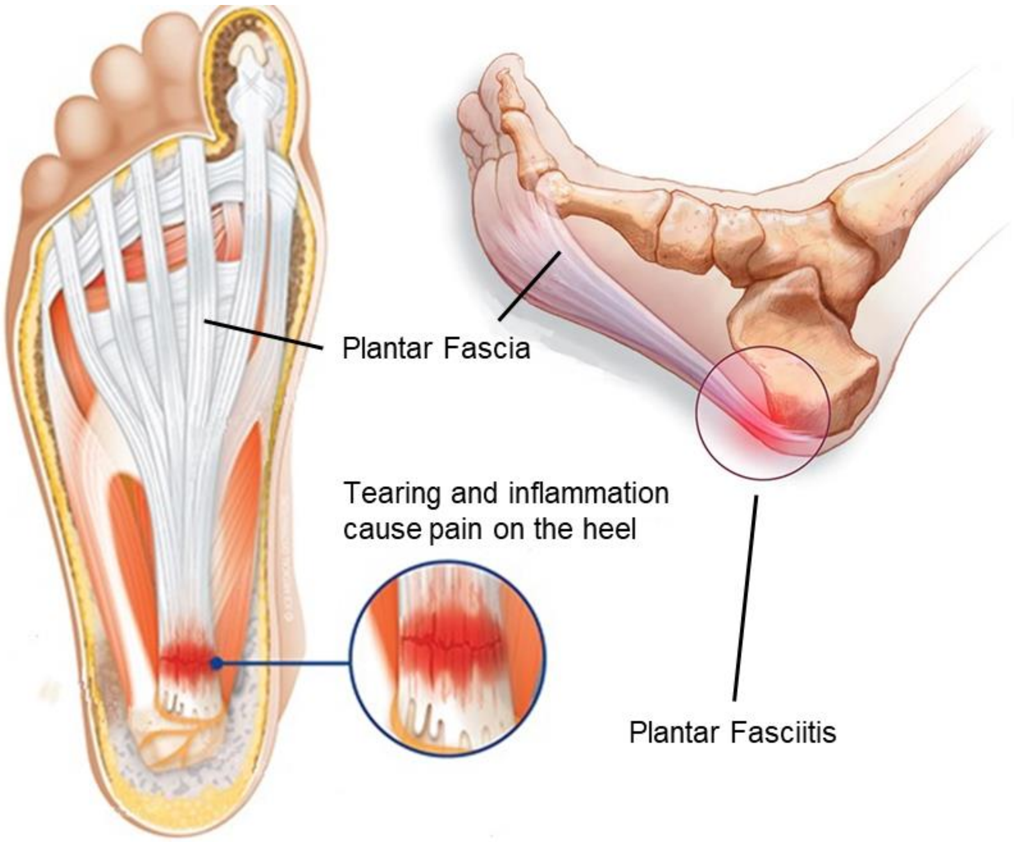


Figure 11. Plantar fasciitis.

Diagnosis of plantar fasciitis should follow the clinical guideline targeting a series of findings from medical history and physical examination (Martin et al., 2014): plantar medial heel pain with initial steps after a period of inactivity, which can be worsened following prolonged weight bearing; heel pain precipitated by a recent increase in weight-bearing activity; pain with palpation of the proximal insertion of the plantar fascia; positive windlass test; negative tarsal tunnel tests; limited active and passive talocrural joint dorsiflexion range of motion; abnormal foot posture index score; high body mass index in nonathletic individuals (Redmond et al., 2006). When the patient's presentation is not typical, or the symptoms are not resolving with interventions, differential diagnosis should also be performed to identify other types of injuries: calcaneal stress fracture, the heel fat pad syndrome, longitudinal arch strain, and the nerve entrapment syndrome (Tahririan et al., 2012).

2.3.2. Epidemiology

On an annual basis, more than two million individuals were diagnosed with plantar fasciitis in the United States, which was about 15% of all adult foot complaints (Roxas, 2005). In addition, about 10% of the population were expected to experience heel pain in their lifetime (Riddle et al., 2003). The percentage could be multiple times more than projection because a large portion of sufferers may not seek medical consultation even when they are symptomatic (Diamond, 2017). Athletes and physically active individuals represented the most prevalent and vulnerable group to plantar fasciitis (Taunton et al., 2002). Overweight and elderly people are also more susceptible to plantar fasciitis (Orchard, 2012; Wearing et al., 2006a), and females exhibited higher morbidity in comparison to males (Scher et al., 2009).

2.3.3. Economic and social impact

For most cases, patients with plantar fasciitis are unsatisfied with the pain and functional limitations and seek treatments. Clinicians also suggested an early medical intervention to prevent possible risk of recurrence and complication (Young et al., 2001). Medical expenses of plantar fasciitis are considerably costly, which is further exaggerated by the growing prevalence. The annual expenditure on various treatments for plantar fasciitis and heel pain syndrome was estimated to be 284 million dollars in the United States in 2010 (Tong and Furia, 2010). The annual cost of medical care per individual could reach 10 thousand dollars, wherein surgical treatment predominates the cost, as compared with medication and physiotherapy (Tong and Furia, 2010). Besides, footwear and orthoses for patients with plantar fasciitis cost over 10 million (Rome et al., 2004). The burden of medical care further upsurges when the patients have a lengthened recovery period and post-fasciitis chronic symptoms, which brings in other health problems associated with long-term immobility caused by foot pain (Beeson, 2014). Plantar fascia used to receive less attention from the medical professional until recently the injury becomes a serious public health challenge to the society, in which a growing number of the population are less physically active (Orchard, 2012). The rehabilitation procedure will sometimes be enduring and frustrating for individuals. The chronic symptom of plantar fasciitis may cut-off regular physical activities for them and add to the vicious circle. Plantar fasciitis is normally activity based and could be recurrent due to its infamous stubborn nature. The amounts of time and money caused by the tedious medical procedure are bothersome for either individuals or society.

2.3.4. Pathomechanics

The pathologies of plantar fasciitis are currently not well understood. In regardless, there is a consensus in the literature that attributes the injury to repeated micro tears of the tissues secondary to mechanical overload and excessive strain on the fascial band (Cutts et al., 2012). The progression of plantar fasciitis is thought to be similar to that of tendinitis. Histological studies have reported multiple degenerative changes occurring to the collagen fibral structure of the plantar fascia and the fibrocartilage on its origin in the injured group (Wearing 2006). Though not necessarily coexisting, tissue degeneration of the plantar fascia is usually accompanied by acute and chronic inflammation, which indicates localized tissue rupture when vascularization cannot provide timely repairmen.

2.3.5. Risk factors

Both intrinsic and extrinsic factors that can overload the plantar fascia are thought to involve in the development of plantar fasciitis. For most of the cases, the onset of plantar fasciitis is likely attributed to a combined effect of multiple factors. With few exceptions, the underlying mechanism of the injury is invariably an increased strain on the plantar fascia caused by either enlarged total force, loading redistribution, or tissue deterioration.

Intrinsic risk factors

Intrinsic risk factors are mostly related to individuals' physical characteristics that predispose them to plantar fasciitis. Based on the nature of causes, intrinsic risk factors of plantar fasciitis can be subdivided into three categories—anatomical, functional, and degenerative risk factors.

Anatomical risk factors include pes planus (flat foot, fallen arches), pes cavus (high arches), leg-length discrepancy, rearfoot overpronation (excessive inward roll of the foot when landing on the ground), femoral/tibial torsion, reduced passive ankle dorsiflexion, and overweight (Dyck and Boyajian-O'Neill, 2004; Irving et al., 2006; Schwartz and Su, 2014). Aberrant foot arch was believed to influence strain distribution on the plantar fascia and cause concentrated strains in certain tendinous regions (Orchard et al., 2004). The foot arch is constituted by hypermobile joints in pes planus and could collapse under body weight, whereby the plantar fascia is overly stretched as it functions to resist arch flattening (Huang et al., 2004). Runners with histories of plantar fasciitis showed significantly lower arch height index (0.315 vs. 0.344, p = 0.01) than the healthy (Pohl et al., 2009). Visa versa, a foot with pes cavus lacks the normal foot flexibility to dissipate the forces of contact. Decreased shock absorption results in increased tension forces being applied to the insertion of the plantar fascia (Bolgla and Malone, 2004). Patients of plantar fasciitis reduced their longitudinal arch index (0.17 vs. 0.22, $p < 0.01$, lower value indicates increased arch height) compared to the healthy control (Ribeiro et al., 2011). The reduction was more exaggerating in acute cases (0.15 vs. 0.17) than chronic cases (Ribeiro et al., 2016). Unequal limb length simply breaks the balance of body-weight-bearing and adds to the loading on the longitudinal foot arch and the plantar connective tissues on the injured side (Mahmood et al., 2010). The leg length discrepancy could be as much as 0.75 cm in patients of plantar fasciitis (Mansur et al., 2019). Rearfoot pronation is a normal segmental movement during walking and running. The motion unlocks the ankle-subtalar complex and aids in absorbing impact from contact. Overpronation, on the other hand, can cause excessive tension on the plantar fascia (Ribeiro et al., 2011). Prichasuk and Subhadrabandhu (Prichasuk and Subhadrabandhu, 1994) calculated the calcaneal pitch to indicate the degree of rearfoot pronation. The injury group had a significantly lower mean calcaneal pitch, which meant higher rearfoot pronation, compared with the control group (15.99 degrees vs. 20.54 degrees, $p < 0.001$). Individuals with limited ankle joint mobility were thought to increase the rearfoot pronation to compensate for the reduced ankle dorsiflexion, which, as above mentioned, could overload the plantar fascia (Riddle et al., 2003). Riddle et al. reported the injury odds of plantar fasciitis were increased by 22.3 times in individuals with the passive ankle dorsiflexion lower than 0 degrees compared to those who had the passive ankle dorsiflexion larger than 10 degrees (Riddle et al., 2003). The same trend was also found in athletes, wherein the injured group produced fewer degrees of passive ankle dorsiflexion (38 degrees vs. 63 degrees, $p < 0.01$) than the healthy (Kibler et al., 1991). A similar mechanism is also applicable to femoral/tibial torsion. Excessive lower limb internal rotation during the stance phase could enforce larger rearfoot pronation due to the tight coupling of consecutive bodies in the kinematic chain (Kirby, 2000). Individuals with a body-mass index of greater than 30 kg/m^2 were almost six times more likely (odds ratio: 5.6 (1.9—16.6), $p < 0.01$) to develop heel pain compared to those with a body-mass index controlled lower than 25 kg/m^2 (Riddle et al., 2003). The averaged body-mass index was 2.20–3.66 higher in patients of plantar fasciitis than the healthy control (Prichasuk and Subhadrabandhu, 1994; Rano et al., 2001).

Functional risk factors include tightness in the calf muscle complex (Taunton et al., 2002) and weakness of the gastrocnemius, soleus, and intrinsic foot muscles (Huffer et al., 2017). A series of physical examinations showed that the calf muscle tightness was 33.88–80.11% higher in patients of plantar fasciitis compared to the control (Bolívar et al., 2013). Results of the examinations were strongly correlated to the risks of plantar fasciitis (Bolívar et al., 2013). Foot muscles with insufficient strength are less competent to share the total work on the foot arch and the excessive forces are thus transmitted to the passive components, such as the plantar fascia (Headlee et al., 2008). Muscle's volume can directly determine its maximal contraction capacity and patients of plantar fasciitis were reported to reduce the average volume of the intrinsic foot muscles by 5.30–15.91 cm^3 (4.68–11.77%) compared to the healthy (Chang et al., 2012; Cheung et al., 2015).

The primary degenerative risk factor of plantar fasciitis is aging (Irving et al., 2006). Aging was frequently associated with deteriorated material properties of the soft tissue and its altered mechanical behavior under a given loading condition (Hsu et al., 2005; Kwan et al., 2010), which could easily contribute to the pathologies of plantar fasciitis. It was found that patients of plantar fascia were significantly older (mean differences: 2.9–9.1 yr) than the uninjured (Rano et al., 2001; Rome et al., 2001).

Extrinsic risk factors

Unlike intrinsic risk factors, extrinsic risk factors of plantar fasciitis are more related to the biomechanical faults in movement strategies that produce excessive loadings on the plantar fascia. Two major causes of plantar fasciitis for individuals with normal somatotype and healthy foot musculoskeletal structure are inappropriate footwear choice and training regimen (Fredericson and Misra, 2007; Wilk et al., 2000). Shoes with defective motion control on the heel (Wilk et al., 2000), insufficient cushioning in the foot/ground interface (Bowser and Davis, 2010; Pohl et al., 2009), and limited supports to the foot arch (Salzler et al., 2012) could cause excessive rearfoot motion, increase impacts to the tissue regions, and produce large structural deformation on the foot, which subsequently irregulate movements of the plantar fascia and exacerbate strains within the fascia band. Though not fully verified, a sudden change in activity level, e.g., frequency, intensity, or duration, is thought to exacerbate the mechanical burden on the plantar foot as well as the injury risk for runners (Bartold, 2004). Training errors were potentially responsible for 17.2% of all causes of plantar fasciitis reported to clinic (Jack E. Taunton et al., 2002). For an instance in running sports, large weekly volume or sudden increases in weekly mileage could contribute to injury incidence. A cohort study found that novice runners who progressed their weekly running distance by more than 30% were likely more vulnerable to overuse injuries than runners who increased their running distance by less than 10% (hazard ratio = 1.59, $p = 0.07$) (Nielsen et al., 2014). Other high-risk behaviors include speedy workouts, hill workouts, plyometrics and etc., which could stretch the plantar foot ligaments before growth of the foot muscle strength and predispose runners to injuries (Young, 2012).

2.3.6. Plantar fasciitis in runners

Though the knee joint is the most complained region of the lower limb in running events, plantar fasciitis is still one of the top five injury types that affect millions recreational runners (Tenforde et al., 2016). In general population, plantar fasciitis is an important public health disorder that causes heel pain in the outpatient setting. Ten percent of people in the United States may experience heel pain over the course of their lives, with 83% of these patients being physically active adults between the ages of 25 and 65 years old (Riddle and Schappert, 2004).

As a matter of fact, runners are easily vulnerable to plantar fasciitis. Epidemiological studies have revealed an averaged 4%–22% injury rate of plantar fasciitis in regular runners. The odds are slightly higher than those of ordinary people (Ballas et al., 1997; Rome et al., 2001).

The repeatability nature of running sports can particularly expose runners to plantar fasciitis. Risk factors of plantar fasciitis increase tensions on the fascial band for a single stance phase and the effects could accumulate over thousands of gait cycles (Chen et al., 2016; Edwards et al., 2009). Constant overload on the plantar fascia inhibits the normal repair process, resulting in collagen degeneration, which causes both structural changes and perifascial edema (Narváez et al., 2000; Wearing et al., 2004).These changes, in turn, lead to a thicker heel pad, which has been shown to be associated with pain in individuals with plantar fasciitis.

2.4. Running

2.4.1. Popularity

Running is one of the most popular, and perhaps the most important activities due to its known positive influence on the runners' physical fitness and its benefits of reducing the incidence of obesity, cardiovascular diseases, and many other chronic health problems (van Gent et al., 2007). Compared to other forms of sports, running is easy to access for a wide age range of the population. According to an annual-renewed survey conducted by Ipsos, the total number of Americans that participate in regular running has raised from 38.7 million to 55.9 million from 2006 to 2017 (Statista Research Department, 2018). Their averaged weekly mileage could be as many as 21 miles. Most of the runners tended to run year-round and planned to participate in more running events the next calendar year (Running USA, 2017).

2.4.2. Running related injury

Running related injuries remain highly prevalent during the last thirty years since the market saw a blooming running industry. The annual injury rate caused by running was estimated to be 30% to 75% (Hryvniak et al., 2014), depending on the survey methods used and the target population. Nevertheless, running injury is one big challenge to the runners and the problem is still largely unsolved, despite the considerable efforts to eliminate it (van der Worp et al., 2015).

2.4.3. Running style

Running style is a summary of the running techniques that characterize how runners coordinate their body segments and muscles from step to step. Running style could vary considerably in persons and is usually the major factor that influences the running biomechanics (Hall et al., 2013; Perl et al., 2012), which subsequently decides how force is transmitted through the body and the extent to which the loading will cause injuries (Dugan and Bhat, 2005). Among the many aspects that constitute a running style, foot strike pattern is the one that attracts much research attention (Hamill and Gruber, 2017). Foot strike pattern defines which part of the foot collides with the ground first during initial contact as well as how the ground reaction force applied to the lower limb in the stance phase (Hamill and Gruber, 2017). Running is usually comprised of thousands of repeated collisions between the foot and the ground. As the body moves forward and downward in every footstep the lower limb undergoes rapid deceleration and generates a large impact shock on the joints (Bosch and Klomp, 2004). Excessive or abnormal loading to the segments and joints are frequently related to trauma (Norton and Eston, 2018). From this standpoint, research has been focusing on how different foot strike patterns and so-induced changes will affect injury prognosis and running performance (Barton et al., 2016). Though exceptions exist to every rule, there should be a general trend regarding the correct running style to follow to avoid foot problems, e.g., plantar fasciitis.

2.5. Foot strike patterns in running

2.5.1. Types of foot strike patterns

Normally, there are three types of foot strike patterns that runners employ. They are generally referred to as rearfoot strike, midfoot strike, and forefoot strike. The definition is based on which portion of the plantar foot hits the ground first during initial contact (Daoud et al., 2012). Rearfoot strike (rearfoot strike) is the most commonly seen foot strike pattern in which the runner contacts the ground with the lateral aspect of the heel. A midfoot strike is the one in which the runner parallels the plantar foot with the ground surface and touchdown on the heel and metatarsal heads almost simultaneously. In forefoot strike, the ball area of the forefoot contacts the ground before the rest of the foot and the heel is mostly lifted off the ground throughout the stance phase (Daoud et al., 2012). Due to the modern footwear technology that provides the runners with abundant cushioning for impact attenuation, most of the recreational runners nowadays are rearfoot strikers (88.9–95.1%) (de Almeida et al., 2015; Larson et al., 2011). Meanwhile, midfoot strikers and forefoot striker only account for 3.4–4.1% and 0.8–1.8% of the overall runner population respectively (de Almeida et al., 2015; Larson et al., 2011). From the biomechanical point of view, midfoot strike is usually considered as the optimal strike form that neutralizes the deficiencies of the other two strike forms (Kumar et al., 2015). Thought the statement has yet to be justified, the differences between rearfoot and forefoot strike patterns seem to be prominent and therefore received more research attention in the field of running biomechanics (Almeida et al., 2015).

2.5.2. Biomechanics of running with rearfoot strike and forefoot strike

Most of the recreational runners nowadays are habitual rearfoot strikers (de Almeida et al., 2015; Larson et al., 2011). For the alleged purpose of better protection from injuries. some runners are willing to transfer to forefoot strike at some point of their careers (Kaplan, 2014). Despite this drive, there are usually two most important suggestions from the coaches on how to modify the foot strike pattern successfully: hit the road with the ball of the forefoot and lift the heel off the ground during stance (Figure 12). In a general condition, forefoot strike is thought to soften the impact shock for the runner due to the increased recruitment of the calf muscles and ankle joint in the overall running dynamics (Fuller et al., 2016). Research has attributed these merits of forefoot strike to its biomechanical differences from the conventional rearfoot strike in many aspects: spatiotemporal variables, kinematics, kinetics, pedobarographic measures, and muscle activities.

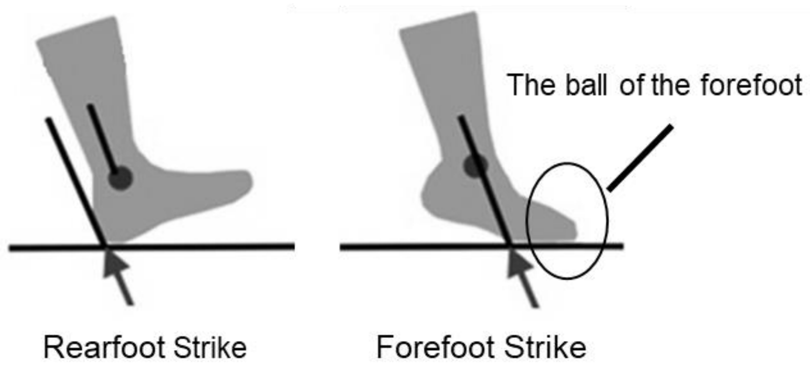


Figure 12. Different foot strike patterns. Rearfoot strikers land on the heel while forefoot strikers land on the ball of the forefoot.

Spatiotemporal variables

Previous studies frequently found a significantly shorter stance phase (mean differences: 3.00–30.00 ms) and stride length (mean differences: 0.01–0.03 m) in forefoot strike than rearfoot strike (Almonroeder et al., 2013; Ardigo' et al., 1995; Bowersock et al., 2017; Goss and Gross, 2013; Knorz et al., 2017; Kulmala et al., 2013; Shih et al., 2013). To keep the same pacing strategy, runners were speculated to increase their cadence to complement the reduced stance time and stride length. However, relevant findings were inconsistent as most of the studies failed to report significant differences in the comparisons of step frequency between the two foot strike patterns, despite that the averaged cadence in forefoot strike was sometimes slightly higher (mean differences: 3–11 times/minute) than that in rearfoot strike (Ardigo' et al., 1995; Goss and Gross, 2013; Kulmala et al., 2013; Shih et al., 2013). The distance between the body center and the stride could be changed in forefoot strike, either increased by 0.15 m (Shih et al., 2013) or decreased by 0.06 m (Kulmala et al., 2013).

Kinematics

The most apparent differences in kinematics between rearfoot strike and forefoot strike resided in the moment of initial contact. By definition, runners need to plantarflex the foot at touchdown when using forefoot strike. The angle between the foot and the ground during initial contact was 8.00–8.92 degrees in forefoot strike and -3.37–-1.6 degrees in rearfoot strike (Altman and Davis, 2012a; Nunns et al., 2013). Several authors noticed that, to allow for a more plantarflexed foot for landing, a greater compliance of the subtalar, ankle, and knee joint was presented in runners using forefoot strike. The subtalar eversion, ankle plantarflexion, and knee flexion at touchdown was increased by 0.10–2.72 degrees, 4.47–28.7 degrees, and 0.16–9.46 degrees respectively in forefoot strike compared to rearfoot strike (Kowalski and Li, 2016; Kulmala et al., 2013; Landreneau et al., 2014; Laughton et al., 2003; Nunns et al., 2013; Rice and Patel, 2017; Shih et al., 2013; Stackhouse et al., 2004; Williams et al., 2012), despite that significances may not always reached in statistics. Hip angle at touchdown varied among the studies as some runners using forefoot strike techniques tempted to increase hip flexion (Kowalski and Li, 2016; Williams et al., 2012) while others did not (Kowalski and Li, 2016; Kulmala et al., 2013). In terms of joint range of motion, ankle excursion (mean differences: 1.54–3.6 degrees) and rearfoot excursion (mean differences: 0.4–5.9 degrees) were commonly increased while knee excursion was reduced (mean differences: 0.03–4.71 degrees) in forefoot strike running (Goss and Gross, 2013; Landreneau et al., 2014; Laughton et al., 2003; Nunns et al., 2013; Pohl and Buckley, 2008; Stackhouse et al., 2004; Williams et al., 2000).

Kinetics

Lower limb loading was reported to redistribute between the knee joint and ankle joint in forefoot strike running, as the ankle joint offloads the knee joint by yielding more mechanical work to absorb the impact of landing (Fuller et al., 2016; Rooney and Derrick, 2013; Stearne et al., 2014). Research demonstrated a reduced ankle stiffness (mean differences: 5.00–7.09 Nm/degree) and an increased knee stiffness (mean differences: 0.7–5 Nm/degree) in runners using forefoot strike (Butler et al., 2003; Hamill et al., 2014). Likewise, ankle joint moment was generally higher (mean differences: 0.05–0.12 Nm/kg) in forefoot strike than rearfoot strike. Oddly, forefoot strike did not alleviate the knee joint loading for the runners as expected. forefoot strike induced a mild increases in the ankle joint contact fore (0.90–1.08 BW), while knee joint contact force was frequently found to be comparable between forefoot strike (6.56–12.46 BW) and rearfoot strike (6.66–13.14 BW) (Bowersock et al., 2017; Boyer and Derrick, 2018; Rooney and Derrick, 2013).

Ground reaction force

The primary kinetic variable that attracts much research interests is the vertical GRF during stance. Forefoot strike is most characterized by an absence or minimization of an impact peak in the vertical GRF compared to rearfoot strike (Giandolini et al., 2014) (Figure 13). Subsequently, the mean and peak loading rate of the vertical GRF during the early stance phase were lower in forefoot strike than rearfoot strike. Compared to rearfoot strike, forefoot strike reduced the mean and peak loading rate of the vertical GRF by 1.7–17.46 BW/s and 11.09–46.6 BW/s respectively (Chen et al., 2016; Goss and Gross, 2013; Kulmala et al., 2013; Laughton et al., 2003; Shih et al., 2013). On the contrary, the magnitude of maximal vertical GRF was usually similar between the two foot strike conditions (mean differences: 0.11–0.32 BW) (Hashizume and Yanagiya, 2017; Knorz et al., 2017; Kowalski and Li, 2016; Kulmala et al., 2013; Williams et al., 2000; Yong et al., 2014).

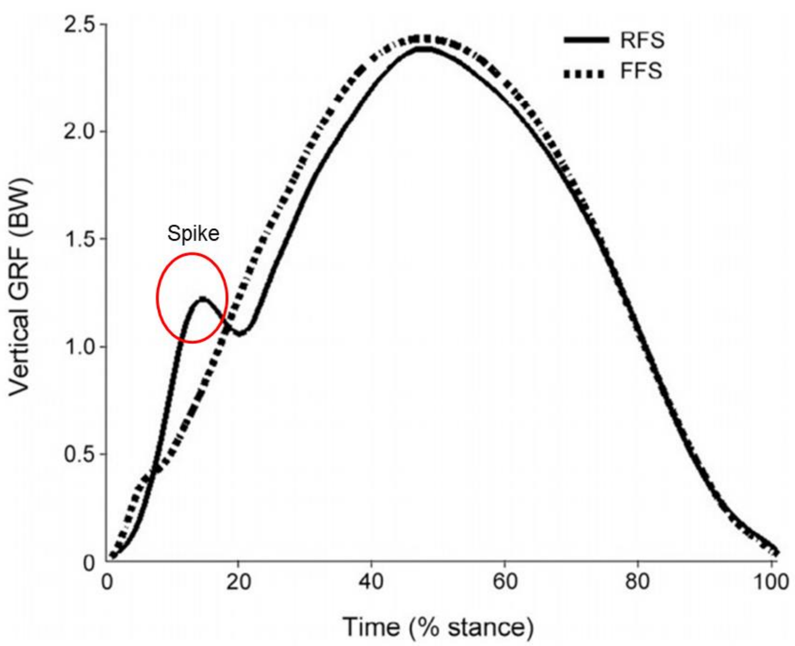


Figure 13. A typical presentation of the vertical GRF produced by a rearfoot striker and a forefoot striker. (Reprinted from I. Davis et al., 2017, Journal of Sport and Health Science, 6(2), pp. 154–161, Copyright 2017 by Elsevier Ltd.).

There is clearly a spike on the curve of rearfoot strike running that indicates an impact peak during the early stance. This spike disappears on the curve of forefoot strike running. RFS: rearfoot strike. FFS: forefoot strike.

Muscle activity

Theoretically, forefoot strike requires more muscular output from the ankle plantarflexors to incline the foot for landing and lift the heel during the stance phase, whereas rearfoot strike may pre-activate the ankle dorsiflexiors in late swim phase (Figure 14). The statement was supported by previous studies (Landreneau et al., 2014; Shih et al., 2013; Yong et al., 2014), in which the authors reported an increased activation of the gastrocnemius (0.44–0.48 RMS) and reduced activation of the tibialis anterior (0.25–0.79 RMS) in forefoot strike running compared to rearfoot strike running. The Achilles tendon tension was also increased significantly (mean differences: 2.48–5.96 BW) in forefoot strike (Almonroeder et al., 2013; Hashizume and Yanagiya, 2017; Kulmala et al., 2013; Rice and Patel, 2017). Due to a higher knee stiffness in forefoot strike, the runners seemed to exert more neuromuscular control and muscle co-contraction over the knee joint. As a result, the changes in the activation of the hamstrings were inconsistent in the literature (Landreneau et al., 2014; Shih et al., 2013; Yong et al., 2014). The activity of the intrinsic foot muscles in different foot strike patterns are barely observed. A landmark study conducted by Kelly and the colleagues used an invasive approach to acquire EMG signals of selected plantar foot muscles during running (Kelly et al., 2018). The results showed that forefoot strike induced a higher activation burst in both the abductor hallucis and flexor digitorum brevis than rearfoot strike, indicating increased recruitment of the intrinsic foot muscles in forefoot strike running. Two longitudinal studies investigated how minimalist shoes, which assumedly promoted forefoot strike running, affected the foot muscle strength for the habitual rearfoot runners (Chen et al., 2016; Miller et al., 2014). After undergoing a training course with the shoes, the experimental group exhibited an overall voluminal growth in the intrinsic foot muscles. The findings unveil the role of foot support that the intrinsic foot muscle playing in converted forefoot strike running.

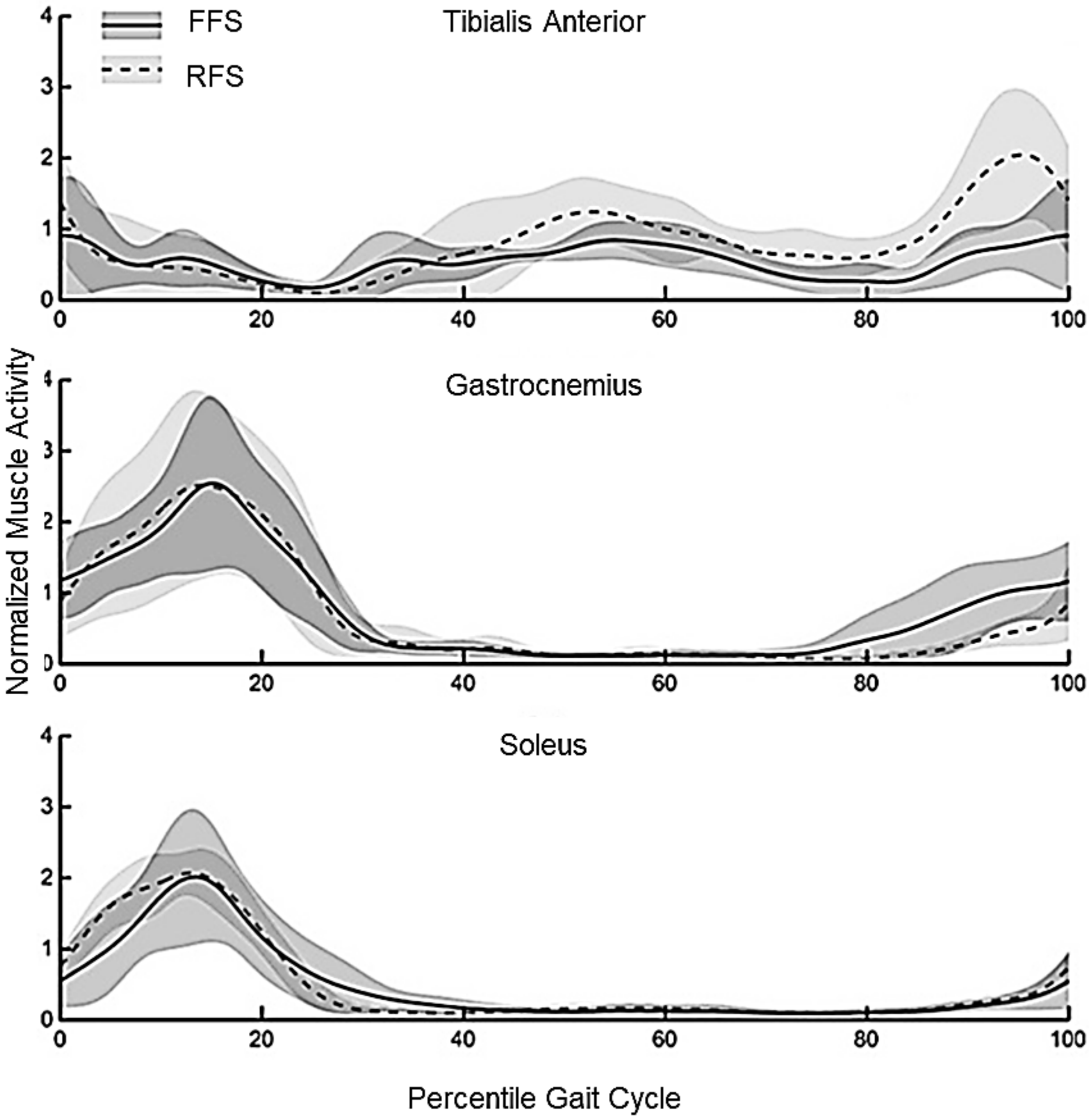


Figure 14. Normalized muscle activation of the tibialis anterior, lateral gastrocnemius, and soleus during running in rearfoot strikers and forefoot strikers. (Reprinted from J. Yong et al., 2014, Journal of Biomechanics, 47(15), pp. 3593–3597, Copyright 2014 by Elsevier Ltd.).

The variables are plotted as a function of the percentile gait cycle. RFS: rearfoot strike. FFS: forefoot strike.

Plantar pressure

For forefoot strike running, the applied point of GRF is shifted anterior to the ankle center and usually located in the forefoot, whereas the heel region is barely in contact with the ground (Kernozek et al., 2016). Pedobarographic studies of forefoot strikers consistently demonstrated that the peak plantar pressure increased in the forefoot region (mean differences: 22.07–126.04 kPa) and reduced in the rearfoot region (mean differences: 147.94–324.88 kPa) (Kernozek et al., 2014, 2016; Nunns et al., 2013). Pressure on the midfoot was usually similar between the two foot strike conditions (Figure 15).

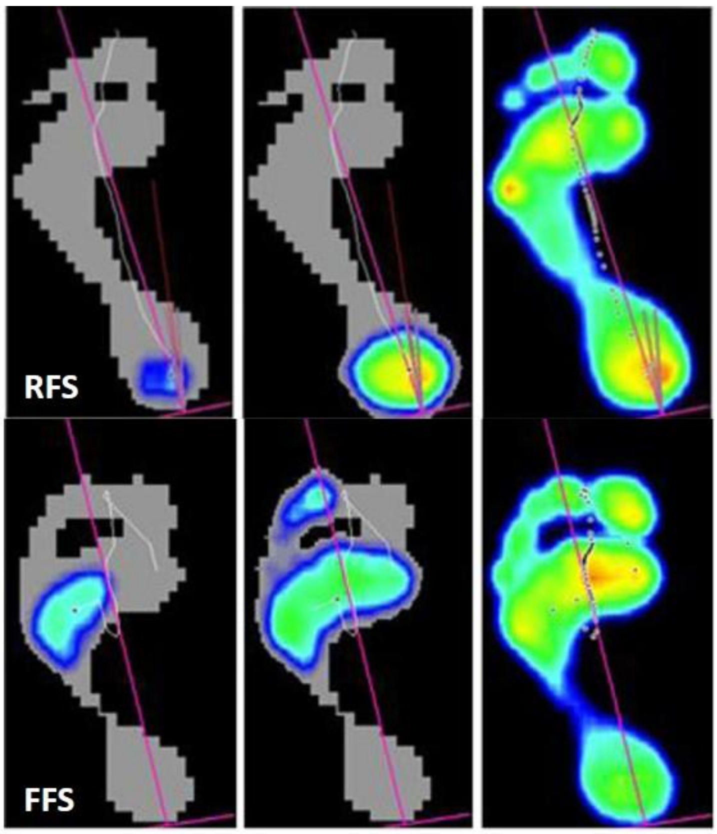


Figure 15. Pedobarographies at early stance of running gait for rearfoot strikers and

forefoot strikers. (Reprinted from M. Nunns et al., 2013, Journal of Biomechanics, 46(15), pp. 2603–2610, Copyright 2013 by Elsevier Ltd.).

For each row, the left-to-right columns show the time-dependent progression of pressure map after initial contact for rearfoot strike running (upper row) and forefoot strike running (lower row). RFS: rearfoot strike. FFS: forefoot strike.

Summary

Due to the large heterogenicity among existing studies, biomechanical differences between rearfoot strike and forefoot strike are not completely understood. Nevertheless, forefoot strikers consistently demonstrated a more plantarflexed ankle and more flexed knee at touchdown compared to rearfoot strikers. Meanwhile, ankle excursion was also increased and knee excursion was reduced in forefoot strike. The more compliant ankle joint in forefoot strike appeared to redistribute the lower limb forces and provide additional cushioning for the foot, causing an overall decreased loading rate of the vertical GRF while also increasing the mechanical burden on the ankle. Because the heel is lifted during stance, a larger and earlier burst of the calf muscle activation and Achilles tension was frequently shown in forefoot strike running, which shifted the center of plantar pressure to the forefoot region and resulted in stress concentrated on the metatarsal bones. These results provide an insight of the biomechanics underlying the different injury patterns between rearfoot strike and forefoot strike, whereby the corresponding treatment strategies can be adjusted accordingly to benefit the runners.

2.5.3. Correlation of foot strike pattern and plantar fasciitis

By far, the correlation of plantar fasciitis and foot strike patterns in running is still tenuous at best. Very few studies addressed and compared the injury rate of plantar fasciitis in a subgroup of runners based on their foot strike patterns. There are authors believing that forefoot strike running can benefit patients of plantar fasciitis because the landing style eliminates direct impacts to the rearfoot and can relieve heel pain for runners (Almeida et al., 2015; Altman and Davis, 2012b; Daoud et al., 2012; Lieberman et al., 2010; Pohl et al., 2008). However, this statement seems to be anecdotal, in a large part, with little to no supportive evidence from controlled studies. Besides, the statement is inconsistent with the findings of biomechanical research, which generated a set of variables that indirectly reflected the tendency in forefoot strike to overload the plantar fascia. A clearer consensus among the scholars is that forefoot strike redistributes the total work done by the knee and ankle joints during running (Fuller et al., 2016). Forefoot strike may reduce mechanical burdens to the upper segments of the lower limb, but its effects on the foot and the plantar soft tissues are still unknown and arguable.

2.6. Taping treatment for plantar fasciitis

2.6.1. Background

Nonoperative treatment is currently the mainstream treatment method for plantar fasciitis (Martin et al., 2014). Nearly 90% of the injured cases underwent conservative treatments and had the heel pain resolved over the course of 12–18 months (Brotzman and Manske, 2011). Based on the underlying mechanism of functionality, conservative treatments for plantar fasciitis can include: activity modification, stretching, night splints, shock wave therapy, localized steroid injection, orthotics, taping, etc. Among these treatment options, taping usually serves as the first-line intervention of the hierarchical treatment system for plantar fasciitis (Orchard, 2012). On some occasions, taping is applied as the palliative treatment in conjunction with other methods for short-term heel pain management (Radford et al., 2006a).

Compared to other treatment options, taping has many advantages. For a starter, taping is a mechanic-based approach that targets on reducing loading on the plantar fascia. Tapes attached to the foot were opined to correct faulty biomechanics, regulate segmental movements, and thus take strains off the plantar fascia (Podolsky and Kalichman, 2015). Second, taping is easy to access from a clinical point of view. It is a convenient, budget-friendly treatment option that is widely used by the patients. Finally, individuals who are physically active, e.g. regular runners, are more compliant with the taping treatment because they can maintain their training routine while wearing the tapes. Some authors claimed that the tapes could give in-situ and step-to-step protections to the plantar foot (Franettovich et al., 2008; Smith et al., 2004), showing the good balance of convenience and effectiveness in the treatment.

2.6.2. Taping modalities

Up to date, a variety of taping modalities have been introduced in the literature. Some of them were constructed based on the same principal but were later on modified by the operators to fit different purposes (Podolsky and Kalichman, 2015). To a certain extent, terminology for taping modality is also equivocal. A same taping modality could be named differently across studies, which makes it difficult to summarize the profile of the taping method for the best interests of both clinicians and patients. Depending on their effects on foot biomechanics, existing taping modalities can be divided into two major categories: the Low-Dye taping and Fascia taping.

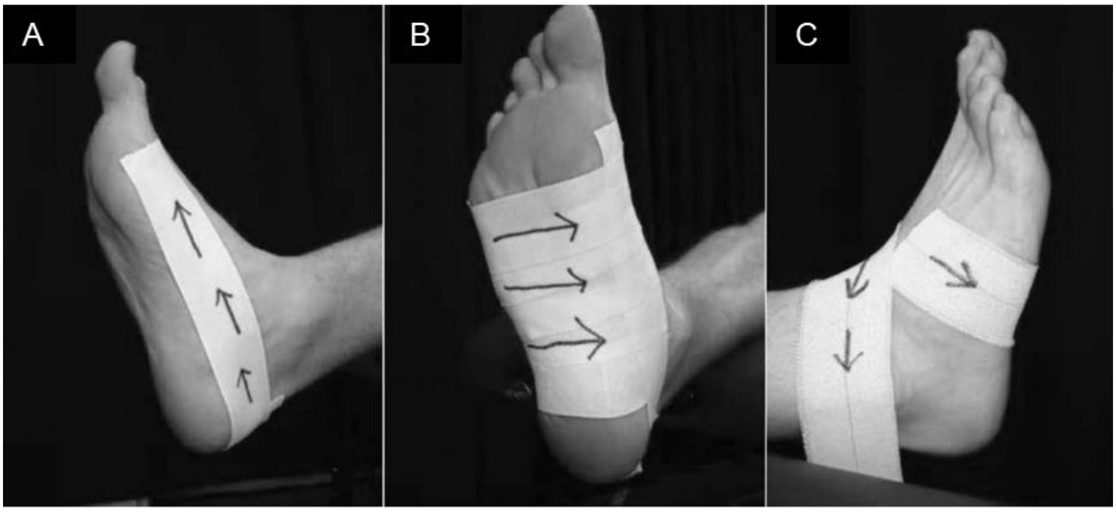


Figure 16. The Low-Dye taping technique. (Reprinted from T. Newell et al., 2015, Journal of Athletic Training, 50(8), pp. 825–832, Copyright 2015 by National Athletic Trainers' Association, Inc.).

Arrows point the directions of stretch. A: an anchor strip wraps around the periphery of the foot; B (bottom view) and C (side view): plantar strips travel under the foot, pull the foot arch inward from lateral to medial, and attach to anterior ankle.

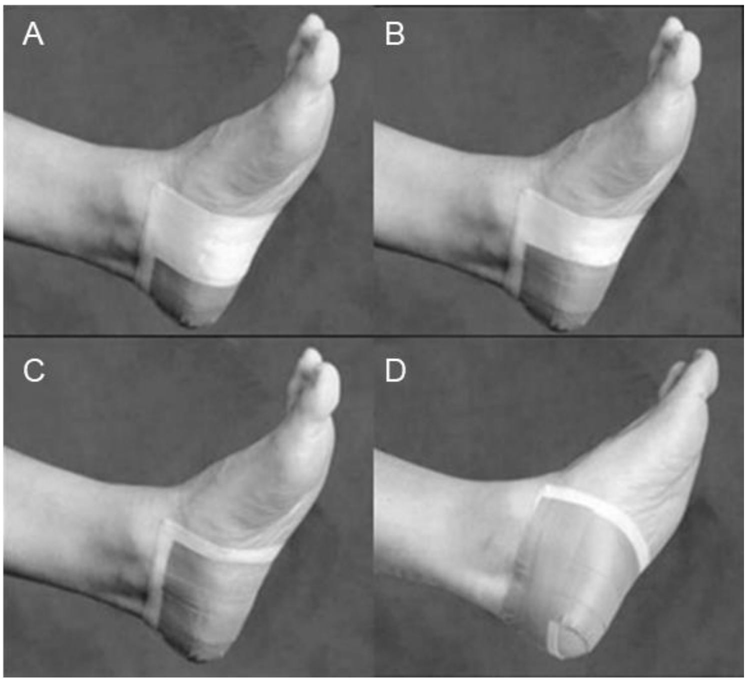


Figure 17. The Calcaneus taping technique. (Reprinted from M. Hyland et al., 2006, Journal of Orthopaedic & Sports Physical Therapy, 36(6), pp. 364–371, Copyright 2015 by Journal of Orthopaedic & Sports Physical Therapy).

A: the first strip starts from the lateral malleolus, travels under the foot, pulls the calcaneus medially, and attaches to just below the medial malleolus; B and C: strip 2 and 3 followed the same pattern with overlap of approximately one third of the tape width, moving in the distal direction; D: strip 4 starts distal to the lateral malleolus, wraps around the back of the heel, and anchors distal to the medial malleolus.

The Low-Dye taping, which is also referred to as the "anti-pronatory" taping, was firstly created by Dr. Ralph Dye (Berkson, 1977). This technique is used predominantly for injuries or pain associated with excessive rearfoot pronation during gait (Figure 16). If properly applied. Low-Dye taping is anticipated to reduce subtalar joint motion and, through the kinematic coupling of the foot segments (Donatelli, 1996), constraint foot arch deformation, and support the plantar fascia (Keenan and Tanner, 2001). Plantar fasciitis that is aggravated by or secondary to rearfoot overpronation should be responsive to this taping technique. Other taping methods (though sometimes named differently) such as "calcaneus" taping (Hyland et al., 2006) (Figure 17), augmented Low-Dye taping (Van Lunen et al., 2011; Vicenzino et al., 2007), and High-Dye taping (Keenan and Tanner, 2001), also fall in this category. Augmented Low-Dye taping and High-Dye taping were extensions to the standard Low-Dye taping by increasing strap coverage over the ankle joint or the midfoot. They add extra tape straps, normally in the forms of U-shape sling or revered six, that wind around one-third up the length of the leg to further control movements of the rearfoot (Vicenzino et al., 2006). Calcaneus taping can be considered as a simplified version of the standard Low-Dye taping method. It wraps the heel with only four pieces of short tapes and avoids involving the medial foot arch. The technique focuses on neutralizing the position of the calcaneus rather than supporting the foot arch (Hyland et al., 2006).

Figure 18. The plantar fascia taping technique.

Arrows point the directions of stretch

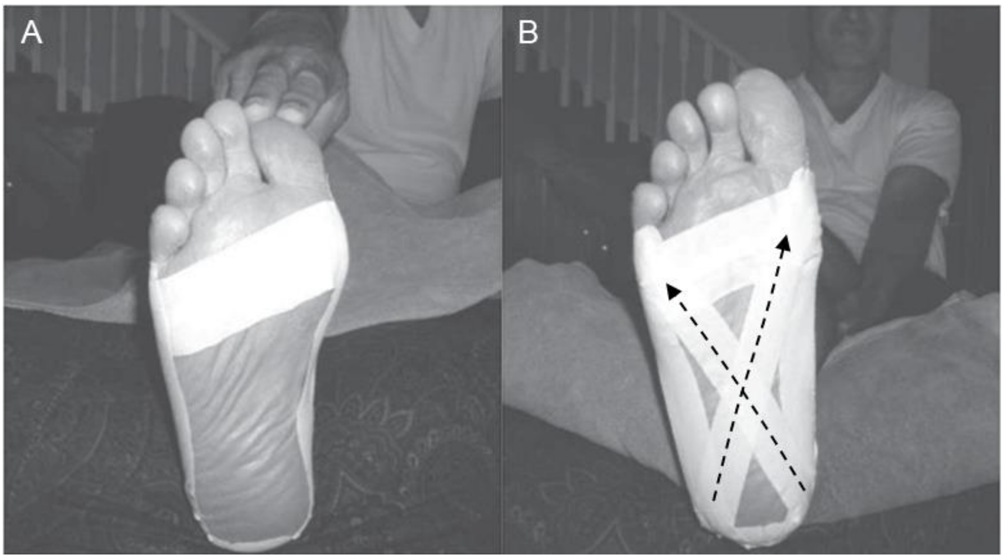


Figure 19. The Windlass taping technique. (Reprinted from B. Jamali et al., 2004, Journal of Sport Rehabilitation, 13(3), pp. 228–343, Copyright 2004 by Human Kinetics Publishers, Inc.).

Arrows point the directions of stretch. A: anchor strips are applied around the periphery of the foot and across the metatarsal heads beneath the foot; B: the windlass strip starts from first metatarsal head, goes obliquely to the rearfoot, wraps around heel, and attaches to the fifth metatarsal head.

The Fascia taping usually features tapes attached directly to the plantar foot surface along its longitudinal axis (Vishal et al., 2010). It is designed to enhance the arch raising effects of the foot "windlass" mechanism (Hicks, 1954) by strengthening the connection between the forefoot and heel (Figure 18). When installed properly, Fascia taping can stabilize the plantar foot ligaments and limit their movements, whereby the fascial tissue is protected from abnormal or excessive stretching. There are many types of taping modalities that can be classified as Fascia taping—"windlass" taping (Jamali et al., 2004) (Figure 19), plantar fascia taping (Tsai et al., 2010), and double x (Ator et al., 1991). Generally, "windlass" taping and double x taping refer to the same layout of tape straps beneath the foot—straps start from the first metatarsal head, go obliquely across the midfoot, wrap around the heel back, and finish at the fifth metatarsal head (Jamali et al., 2004). For other Fascia taping methods, the tape straps are usually placed under the foot in parallel to either the foot longitudinal axis or the direction of the fascial bands (Tsai et al., 2010).

2.6.3. Effects of taping on foot biomechanics

Kinematics and kinetics

Many studies have been conducted to assess the effects of taping treatments on foot kinematics and kinetics. The most research interests lie in the outcomes of rearfoot eversion, navicular height, and plantar pressure during gait. In spite of its designated purpose of controlling rearfoot motion, the standard Low-Dye taping appeared to have limited effects of reducing rearfoot eversion for the wearers. Most of the existing studies using Low-Dye taping failed to reported significant differences in maximal rearfoot eversion between untapped and tapped conditions (Harradine et al., 2001; Keenan and Tanner, 2001; Moss et al., 1993). Speaking of foot arch deformation, both Low-Dye taping (Del Rossi et al., 2004; Jamali et al., 2004) and Fascia taping (Ator et al., 1991; Holmes et al., 2002) could elevate the foot arch during static standing except for that, this efficacy diminished quickly after brief exercises. In most cases, taping studies that measured plantar pressure used the method of Low-Dye taping or augmented Low-Dye taping (Radford et al., 2006a). Some of them mixed Low-Dye taping with Fascia taping (Kim et al., 2011; Lange et al., 2004). Except for one study recruiting patients of plantar fasciitis (Van Lunen et al., 2011), others included individuals with either excessive navicular drop or rearfoot overpronation. Peak pressure in selected plantar regions during walking (sometimes during jogging) was the most frequently reported outcome variable. In spite of the discrepancies in research setups, taping was commonly showed to increase peak plantar pressure in the lateral midfoot (Kim et al., 2011; Newell, 2012; Nolan and Kennedy, 2009; O'Sullivan et al., 2008; Russo and Chipchase, 2001; Vicenzino et al., 2007) and reduce peak pressure in the medial midfoot as well as the whole rearfoot (Lange et al., 2004; O'Sullivan et al., 2008; Pinto Guedes Rogerio et al., 2016). Taping also caused a lower peak pressure in the lateral forefoot (Newell, 2012; Pinto Guedes Rogerio et al., 2016), though this conclusion is less evidential. Two studies reported plantar contact areas pre- and post-taping and found reduced contacts in the rearfoot (Vicenzino et al., 2006) and midfoot (Boergers, 2000) during exercises after the foot was taped.

Muscle activity

A few taping studies performed EMG measurements to the lower leg muscles. Boergers et al. reported that the tibialis anterior was less activated in running under the Low-Dye taping condition (Boergers, 2000). Franettovich and his colleagues found (Franettovich et al., 2012) that flat footers exhibited reduced activation of the tibialis posterior and the peroneus longus in walking after treated by augmented Low-Dye taping. They used the same taping method for patients with leg pain and observed a similar trend of changes in muscle activities (Franettovich et al., 2010). Despite the limited amount of evidence, the usage of tapes appeared to reduce the involvement of foot muscles for locomotion by assisting in organizing joint (tibialis anterior for ankle dorsiflexion at early stance) (Yong et al., 2014) and segmental movements (tibialis posterior and peroneus longus for foot arch support) (Kokubo et al., 2012). Lower muscle activities could be beneficial for preventing injuries associated with overuse and material fatigue (Nur et al., 2015).

Improvement in symptom and function

The current evaluation of taping treatment effects is mainly based on subjective feedback. In most cases, patients rated the level of pain relief and function recovery pre- and post-taping. Despite the variations in research setups, patients of plantar fasciitis receiving taping treatments frequently reported alleviated pain symptoms (Abd El Salam and Abd Elhafz, 2011; Aishwarya and Sai, 2016; Ha et al., 2012; Hyland et al., 2006; Jamali et al., 2004; Landorf et al., 2005; Lynch et al., 1998; Radford et al., 2006b; Tsai et al., 2010; Van Lunen et al., 2011; Vishal et al., 2010). In contrast, the effects of taping on foot functions were less conclusive. Though the patients' functional score was usually improved after taping treatments (Abd El Salam and Abd Elhafz, 2011; Tsai et al., 2010; Vishal et al., 2010), the increments were not always significant in the statistical terms (Hyland et al., 2006). Some studies compared taping to other treatment methods. The results showed that taping might be better than anti-inflammation medicine and physiotherapy (Hyland et al., 2006; Lynch et al., 1998), but worse than arch supports (Abd El Salam and Abd Elhafz, 2011) for pain relief. In the meantime, two studies found that Low-Dye taping was more effective than Fascia taping for short-term pain management (Aishwarya and Sai, 2016; Vishal et al., 2010).

Summary

Existing studies that investigated the effects of taping treatments appeared to center on the proximal-to-middle portion of the foot, as the most frequently reported outcomes were changes in subtalar eversion, navicular height, plantar pressure, EMG signals, and heel pain. All these variables are directly or indirectly related to the loading status of the plantar fascia. The increased navicular height and decreased plantar pressure in the medial midfoot can imply a reduction in rearfoot pronation (Radford et al., 2006b), the latter is allegedly able to loosen the fascial band (Kocaman et al., 2017). EMG measurements showed that the activities of the tibialis anterior and tibialis posterior were changed by taping. These two muscles are known to restrict foot arch flattening as well as shelter the plantar soft tissues from excessive loading (Sarrafian, 1993). Heel pain reflects, in part, the pathological progression of plantar fasciitis, which can be aggravated by foot arch collapse and rearfoot overpronation (Donatelli, 1996). Overall, these findings all point to a positive effect of taping treatments on offloading the plantar fascia, in spite of that direct evidence is currently scarce.

2.7. Research gap and formulation of research questions

It is well-documented that running kinematics affects force transmission in the lower limb and thus influences the incidence of plantar fasciitis (Daoud et al., 2012). After transferring from rearfoot strike to forefoot strike, runners are expected to increasingly engage the plantar foot muscles and ligaments to cope with landing on the forefoot (Kelly et al., 2018, 2014). Due to these alterations, habitual rearfoot strikers were normally suggested to enhance foot strength before changing their landing modes (Daoud et al., 2012; Kelly et al., 2018; Miles et al., 2013).

However, many runners are not aware of these premises for forefoot strike running and expose themselves to an abrupt change of running style, which could possibly overstrain the plantar fascia and predisposes them to a high injury risk (Tan et al., 2008). Clinical studies have raised the concerns of altered material property in the plantar fascia and the resulting plantar fasciitis involved in foot strike pattern modification (Daoud et al., 2012; Lieberman et al., 2010), despite that supportive evidence is largely absent. Very few studies so far have compared the differences between rearfoot strikers and forefoot strikers with focuses on the plantar fascia.

The speculated injury risks of plantar fasciitis involved in forefoot strike running are based on the changed loading environment in which the foot is placed. In forefoot strike running, the vertical GRF at touchdown is shifted anteriorly to the ankle center and applied on the ball area of the foot (Kernozek et al., 2014, 2016). A larger Achilles tension is generated to counteract the ankle dorsiflexion moment caused by GRF and lift the heel off the ground (Landreneau et al., 2014; Yong et al., 2014). Together with the downward ankle joint force, a three-point-bending force is formed upon the longitudinal foot arch (Lieberman, 2012; Perl et al., 2012) and thus increases the tension on many arch load-sharing components, e.g., the plantar fascia (Kirby, 2017). This statement is by far only confined to a theoretical level and yet to be justified. The difficulties of real-time measuring of loading on the soft tissues hinder research and evidence-mining in this field. Fortunately, the well-developed computational technology has enabled dynamic simulations of human movements using a comprehensively constructed foot model, whereby stress/strain on the plantar fascia during running can be quantified and evaluated.

Though many studies have been conducted to examine the treatment effects of taping on plantar fasciitis, the reported outcomes are still conflicting (Radford et al., 2006a; Salvioli et al., 2017; van de Water and Speksnijder, 2010). One explanation for this inconsistency is a lack of unified taping standard. On a clinic base, selection of taping methods to treat plantar fasciitis is largely anecdotal (Franettovich et al., 2008). Variances in taping modalities could affect the generalization of the results and render the integrated outcomes inconclusive.

Besides, direct comparisons are barely performed among taping methods. They may change the foot biomechanics differently while few details are known by the researchers and users. Current measures of the taping treatment effects are mainly patient-rated. Causes of symptom improvement can be multi-factorial and it does not necessarily reflect the changes of loading on the plantar fascia, which is the key element influencing the risks of plantar fasciitis. More quantitative evidence is needed to determine the extent to which different taping treatments can offload the plantar fascia.

CHAPTER 3 OVERVIEW

3.1. Overview on the sub-studies

In the first article, the mechanical property of the plantar fascia, namely the plantar fascia elasticity, was evaluated and compared between rearfoot strikers and forefoot strikers (Figure 20). The underlying rationale is that the elasticity of the soft tissues could be affected under repeated loading upon an accumulation of gait cycles in running. Information of the fascial elasticity could imply the loading status of the plantar fascia as well as the possibly altered injury risks of plantar fasciitis.

The second study aimed to provide quantitative evidence to support the presumed outcomes of changed plantar fascia elasticity in forefoot strikers: loading on the plantar fascia was different between rearfoot strike running and forefoot strike running, more specifically forefoot strike running could produce a higher tensile force on the fascial band (Figure 20). Mechanical overload is one of the major causes of tissue tears and the subsequent degenerative pathologies. Changes in fascial loading are strongly related to the incidence of injury.

The third study investigated how different taping methods could reduce loading on the plantar fascia during running (Figure 20). Taping serves as one of the first-line treatment options for plantar fasciitis and it is particularly suitable for runners because it can provide on-site protection to the plantar fascia on a step-to-step basis. Enthusiastic runners can maintain their training routine while receiving taping treatments. The study sought to support the clinical application of taping for plantar fasciitis through a biomechanical and computational approach.

3.3. Method overview

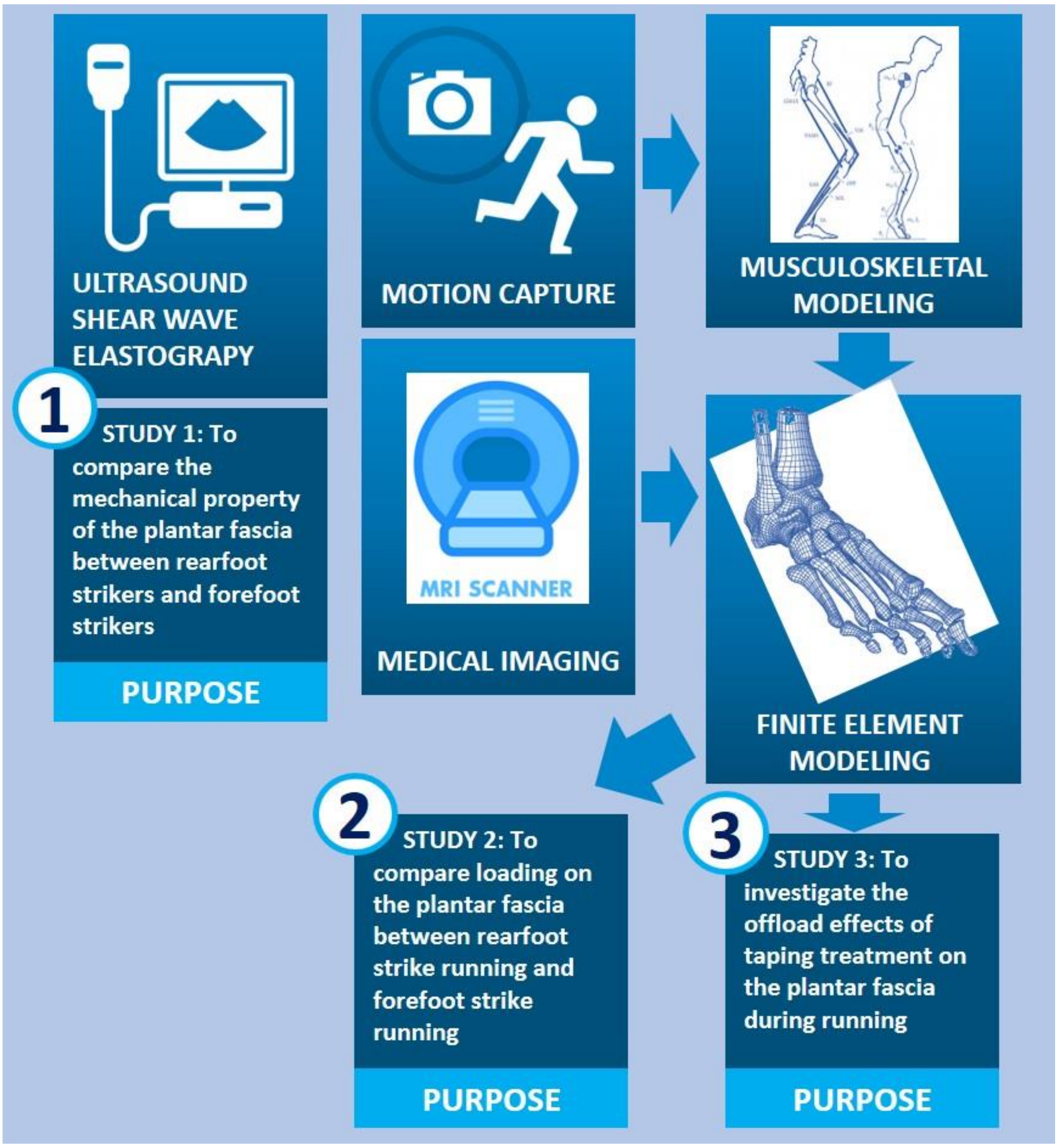


Figure 20. Overview of the methodological workflow.

The ultrasound elastography identifies changes in mechanical property of the plantar fascia for rearfoot strikers and forefoot strikers. This information can reflect the loading status of *the plantar fascia and is associated with injury risks. Computational simulations of running movements were conducted to verify the speculation that, the plantar fascia is loaded differently between rearfoot strike running and forefoot strike running. A finite element analysis using the foot model was also performed to ascertain the effects of taping treatments on offloading the plantar fascia, whereby to support the usage of taping as the front-line therapeutic option for plantar fasciitis.*

3.3.1. Ultrasound shear wave elastography

Background

It used to be difficult to perform noninvasive measurements to the mechanical property of soft tissues until the most recently, the established ultrasound shear wave elastography has enabled an in-vivo and quantitative assessment of the tissue elasticity. Currently, many elastographic techniques are available for clinical application and diagnosis purpose. All of these methods, in regardless of the varied modes, share a same physical principle that underlies the assessment: the imaging modality creates a distortion in the tissue, observes and processes the tissue response to infer the mechanical properties of the tissue (Sigrist et al., 2017). More specifically in the shear wave elastography, a high-intensity pulse (acoustic radiation force) is generated by the ultrasound beam and transmitted deeply into the tissue. The disturbance created by this pulse travels sideways as a shear wave, which extends laterally from the insonated structure. The shear waves can be tracked with low-intensity pulses to determine its velocity and this velocity is proportionally related to the tissue's shear modulus, upon the assumptions that, the measurement is conducted within a uniform and purely elastic system (Sarvazyan et al., 1998):

$$\mu = \rho c_s^2 \quad (1)$$

where μ is the shear modulus, c_s is the shear wave velocity, and ρ is the tissue density that normally ranges from 916 to 1060 kg/m^3 (Nowicki and Dobruch-Sobczak, 2016). Usually, the Young's modulus of a tissue is much higher than its shear modulus. Their relationship can be expressed as following:

$$E = 3\mu = 3\rho c_s^2 \quad (2)$$

where E is the Young's modulus, Young's modulus was reported to be about 10 kPa for parenchymal tissue, about 20 kPa for muscles, and 50 kPa for connective tissues (Nowicki and Dobruch-Sobczak, 2016). In either way, shear wave velocity is assumed linearly correlated to the soft tissue's elasticity and can be used as an indicator reflecting changes in its mechanical property.

Method

Up to date, many studies have applied shear wave elastography to assess human tendons and ligaments. The majority of these studies are experimental studies with the most research interests focusing on the Achilles tendon (Arda et al., 2011; Aubry et al., 2015; Chen et al., 2013; DeWall et al., 2014; Helfenstein-Didier et al., 2016), patella tendon (Hsiao et al., 2015; Taş et al., 2017; Zhang et al., 2014), and joint collateral ligaments (Wu et al., 2016). A few of them investigated the plantar fascia (Shiotani et al., 2019; Zhang et al., 2014). Approximately half of the studies reported the absolute values of shear wave velocity while the rest reported the converted Young's modulus as the primary outcome variable. The Aixplorer ultrasound scanner (Supersonic Imagine, Aix-en-Provence, France) was the dominant instrument for shear wave elastographic measurements. Most of the studies utilized a 4–15 MHz transducer operating in a “musculoskeletal” probe mode for acquiring images.

In this study, a similar research setup, including the instrument model and scanning parameters, was adopted to assess the elasticity of the plantar fascia for recreational runners (Chen et al., 2019a). More details can be found in the method section of study one of this thesis. The scanning parameters were configured with the intention of achieving the best imaging resolution while not compromising the penetration depth of the input acoustic wave (Hill et al., 2002). In addition to equipment tuning, an established algorithm was also employed for post-scan image processing (Lee et al., 2015; Leonardis et al., 2017) for the purpose of increasing the overall measurement accuracy. The algorithm analyzes the quality map (provided by the instrument manufacturer) of each image, screened out poor-quality pixels, and calculated the values of shear wave velocity from the remainders. Shear wave elastography has shown moderate feasibility and validity in measuring many musculoskeletal structures, which paves the way for its application in the plantar fascia.

3.3.2. Motion capture (for tracking running movements)

The incidence and prognosis of foot injury are influenced by changes in the joint and segmental movements occurring during locomotion. In the field of sports biomechanics, motion capture technique is frequently applied to track and record human motions, whereby the acquired data can be processed not only to compute the basic kinematics, but also to provide additional information, e.g., internal loading in the musculoskeletal system, upon data mining.

In this study, a recruited runner was asked to perform over-ground running under several experimental conditions, during which his gait patterns were recorded by a camera-based motion capture system incorporated with force platforms (Figure 21). A set of retroreflective markers were attached to selected anatomic landmarks to track segmental movements.

Locations for marker installment were configured as following to accommodate the OpenSim environment: acromioclavicular joints, posterior/anterior iliac spines, greater trochanters, lateral/medial femoral epicondyles, lateral/medial malleoli, calcaneal tuberosity, the base/head of the first and fifth metatarsals, and the distal phalanx of the hallux. The force platforms measured ground reaction force for steps of interest. Running trials were repeated until target numbers of data set were obtained. Data of kinematics and kinetics can be used to set-up and govern computational simulations of running with either musculoskeletal models or finite element models.

Figure 21. The Vicon system for motion capture.

The system is constituted of many optical-based cameras that can locate frame-to-frame positions of the retroreflective markers. By placing a preset number of markers to selected body landmarks, the movements of the participant are visually trackable within the capture volume. The system can be incorporated with force platforms, by which the GRF of each step in gait can be recorded synchronously with data of marker trajectories. Operations of the system and data processing can be implemented in associated software—the Nexus.

3.3.3. Computational simulations (for predicting loading on the plantar fascia)

In computational simulations of sports, there are two well-developed but separate modeling domains: musculoskeletal modeling (multibody dynamics) for body movements and finite element modeling for tissue deformations. The solutions to many clinical problems, e.g., the pathomechanics of foot injuries, should mostly emerge from both domains. Finite element modeling is particularly adaptive to studies of plantar fasciitis due to the mechanical nature of the injury. Loading on the plantar fascia can be assessed through deformations of the model parts and correlate to injury risks (Campbell and Glüer, 2017). However, predictions of tissue stress/strain during running is subject to setups of boundary conditions, such as the whole-body anatomy, mass distribution, and gait patterns. These factors are not typically represented in finite element models. Musculoskeletal models, on the other hand, can account for these factors in a completely rigid bony construct and calculate time-dependent changes in muscular outputs and segment kinematics by means of inverse dynamics, mathematical optimization, and forward dynamics (Moissenet et al., 2017). As a matter of fact, exploration of foot problems should be better accomplished by integrating movement simulations in both domains.

Musculoskeletal model

Background

A musculoskeletal model usually consists of many rigid bodies connected by mechanical joints with pre-defined degrees of freedom. Muscle units are included in the model to actuate joint rotation and generate requisite momentum for body movements. Outputs from musculoskeletal modeling are informative of running dynamics and can facilitate setting up the finite element analyses, by which loading on the soft tissues can be assessed. Currently, there are two dominating simulation tools in the field of musculoskeletal modeling: AnyBody and OpenSim (Trinler et al., 2019). Except for the variations in platform environments, the two simulation tools are greatly similar in the workflow to calculate desired variables and both of them can produce established results (Trinler et al., 2019).

Method

Musculoskeletal modeling in this study was conducted on the OpenSim platform. To fit the specific research purpose, generic models that incorporated different musculoskeletal structures and muscle functions were obtained from the OpenSim model repository and used for running simulations (Arnold et al., 2010; Delp et al., 2007; Rajagopal et al., 2016). These models are constructed with osseous and muscular details on the lower limb for studies focusing on the knee, ankle, and foot biomechanics.

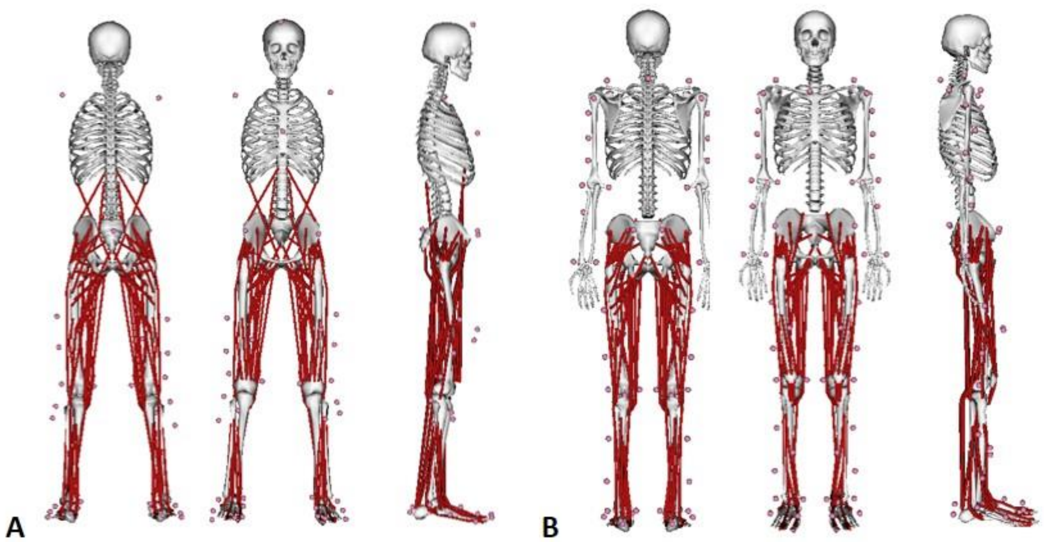


Figure 22. The generic musculoskeletal models used in the study. A: the OpenSim 2392

model (Delp et al., 2007). B: the OpenSim full-body model (Rajagopal et al., 2016).

Predefined virtual markers are created and affiliated to corresponding body segments. Positions of the virtual markers can be adjusted to match up to their counterparts in reality at the beginning of the simulation. Markers on designated body landmarks are used to scale the generic model.

Experimental data collected from motion capture, including marker trajectories and ground reaction force, were imported to the OpenSim platform to drive the movement simulations (Figure 22). The generic model was firstly scaled to the subject' anthropometry using data of a static trial. Scaling ensures that the model's mass distribution and segmental dimensions in the virtual environment matchup those in reality. Inverse kinematics were then solved to compute generalized coordinate values, with which the scaled model best tracked the maker trajectories and reproduced the running gait patterns (Figure 23). Muscle excitations and muscle forces were calculated through an iterative forward dynamics approach (Thelen and Anderson, 2006). By combining proportional-derivative control and static optimization (Thelen and Anderson, 2006), the Computed Muscle Control modulus computed a set of muscular actuator controls that drive the model to track the desired kinematics in the presence of applied external forces (GRF in this case). Finally, the Analyze tool was launched to generate more variables concerning the model states, such as segment orientation and position.

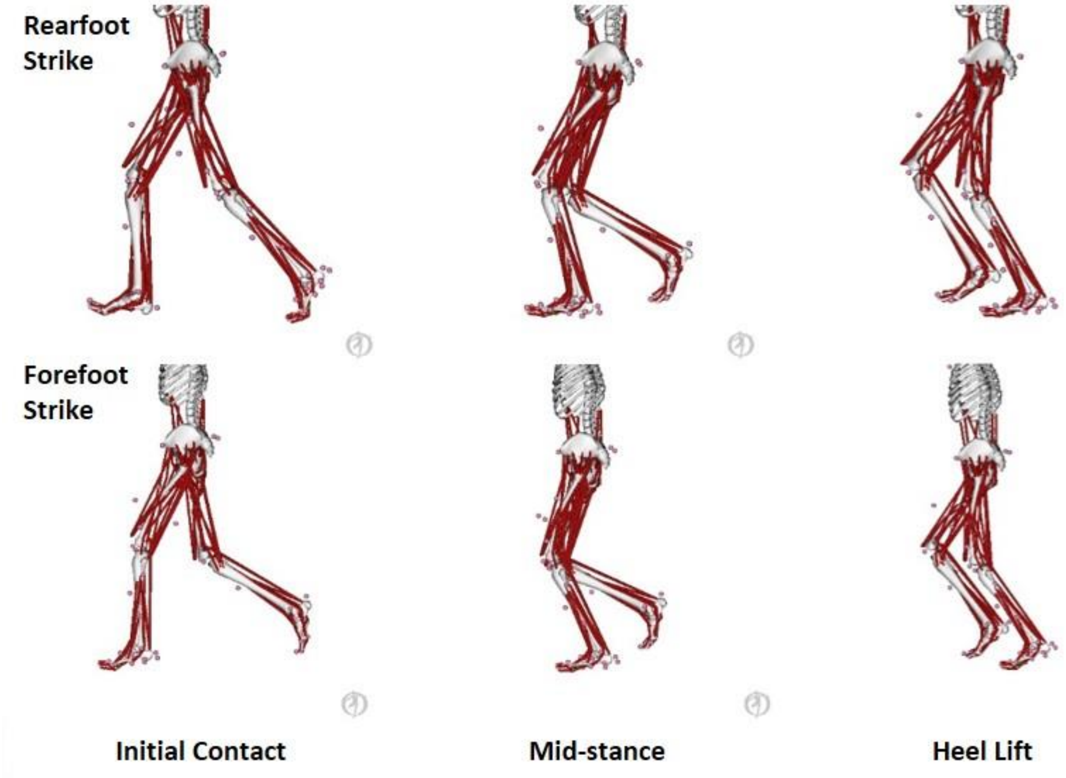


Figure 23. Simulations of running with rearfoot strike and forefoot strike using the scaled musculoskeletal model.

Finite element model

Background

In this study, the finite element model of the foot also includes many independent solid parts that represent the bones. The layout and relative displacements of these bony parts are regulated by either connectors or structures that mimic the functions of the soft tissues, such as the foot muscles and ligaments. One major difference that distinguishes finite element models from musculoskeletal models is that, finite element models are deformable, which can be accessed for information of internal stress/strain within the materials. Another major difference is that, finite element models are usually established for selected body segments rather than the whole body to save the procedure from time-consuming and high computational costs. As a matter of fact, finite element analyses of running movements require inputs of localized loading and positioning from musculoskeletal modeling to initiate and govern the simulations.

Geometry reconstruction

In the field of finite element analyses of the human foot and footwear biomechanics, two main methodological approaches were found for the geometric design: The use of realistic representations of foot geometry or the use of idealized geometry. The former is more complex as it aimed to produce a model specific to the subject and it is adopted by most of the existing finite element studies (Behforootan et al., 2017). For this approach, geometries of the lower limb are directly defined by segmentation and reconstructions from the medical imaging. Early practice of finite element methods was based on X-ray images, while at present most of the research built the foot model upon magnetic resonance images (MRI) or computed tomography (Carey et al., 2014). As the primary goals of this study were to observe the mechanical behavior of the plantar fascia, a finite element foot model was established from the MRIs of a runner who voluntarily participated in the experiment. The regions of interests, including the bones and the bulk soft tissue, were outlined and segmented from a series of cross-sectional images. Layers of the segmented images were superimposed to constitute the three-dimensional geometries. The three-dimensional geometries were smoothed and quality-edited to reduce the geometrical inaccuracy caused by image burr. All processed parts were assembled according to their anatomic positions showed in the MRIs (Figure 24).

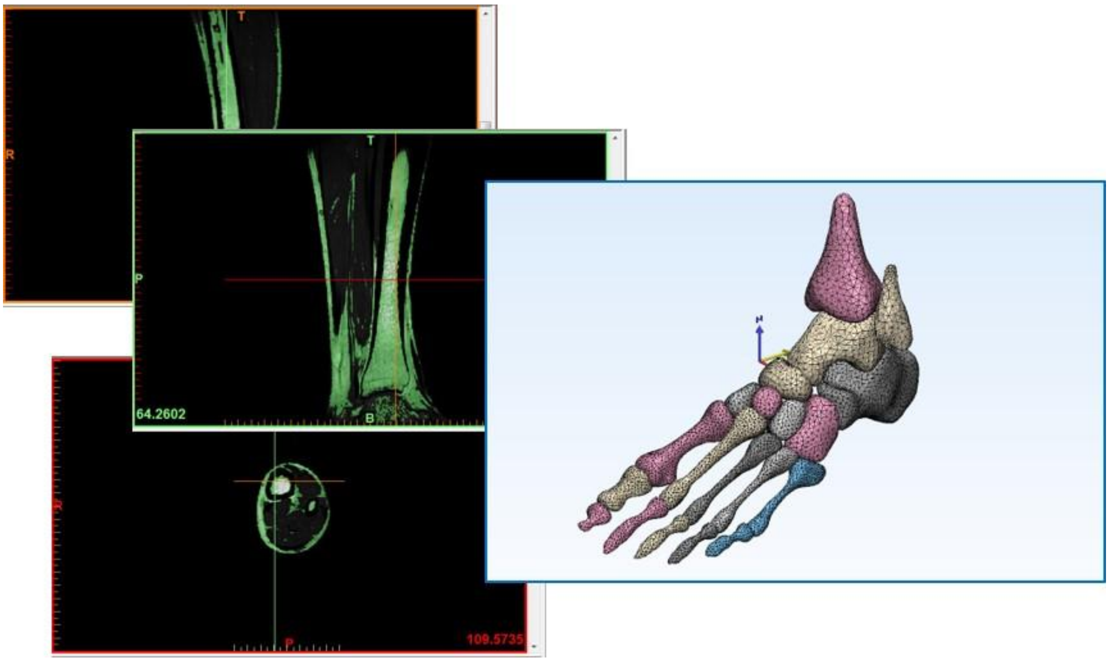


Figure 24. Image segmentation and reconstruction.

From left to right, the bony regions of the cross-sectional MRIs are firstly identified and cropped to create a set of new images. The new images are then overlaid to form the three-dimensional geometry of the target skeletal part.

The bulk soft tissue was modeled as a cluster of SPH particles encapsulated in a shell unit using a cohesive property. The shell unit consisted of interior and exterior surfaces representing the periosteum layer (tied to the bony structures) and skin layer (in contact with the ground) respectively (Figure 25). Ligaments, tendons, and foot muscles were modeled as the wire-based units or connectors that attached to selected bony landmarks based on descriptions of human anatomy (Gray, 2011). Depending on the purposes of the independent studies, the plantar fascia was modeled as either a slip-ring connector or a three-dimensional solid part in the foot model. All parts of the foot model were assumed to be linearly elastic and isotropic except for that, the plantar fascia was assigned with a hyperelastic material property. Detailed information of material property and element type for each model components is listed in tables in the independent studies.

In the study, SPH method was applied to model the bulk soft tissue, instead of the finite element solid mesh, because SPH particles possessed a superior ability to solve large deformation occurring in a short time span (Yang et al., 2016; Zhang and Batra, 2009). Regular deformable elements (either tetrahedral or hexahedral mesh) could be reluctant to simulating speedy deformation due to the fixed connections between nodes and the weakness in handling mesh distortion (Ma et al., 2009). Conversely, SPH was more competent to simulate material crush in a high-impact phenomenon with the discrete meshless points. A test simulation was conducted to illustrate the differences between finite element meshing and SPH method (Figure 26). A cube, meshed by either hexahedral solid element or SPH particles, was assigned with the material property of human foot soft tissue and compressed on the top at a rate of 25000 N/s (approximately equaled to the loading rate of vertical GRF in rearfoot strike running at early stance) (Table 4). The results showed that the finite-element-meshed body could not withstand excessive deform speed and reported simulation failure shortly after the simulation began. On the contrary, deformation was congruently progressed in the SPH model.

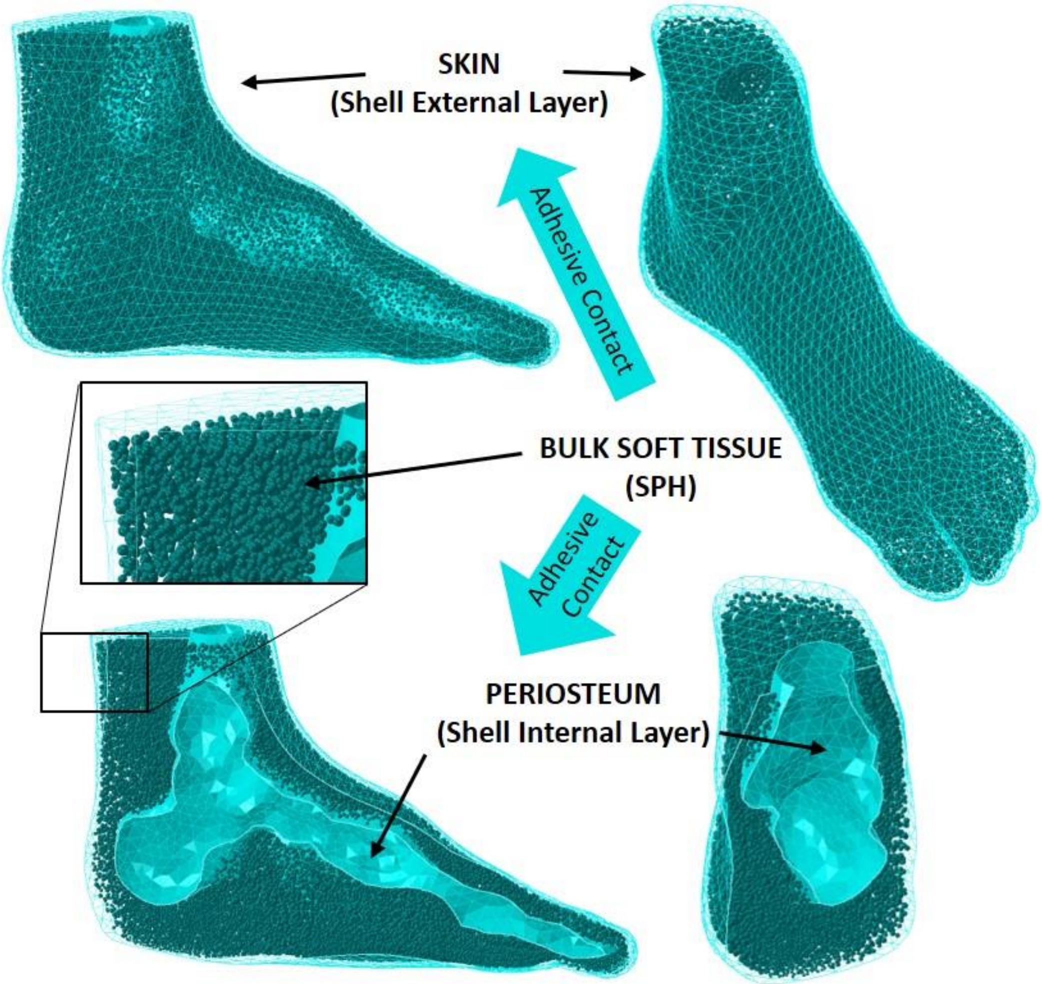


Figure 25. Modeling of the bulk soft tissue.

The SPH particles modeling the bulk soft tissues are encapsulated by a shell unit, of which the internal layer represents the periosteum (tied to the corresponding bony surface) and the external layer represents the skin (in contact with the ground).

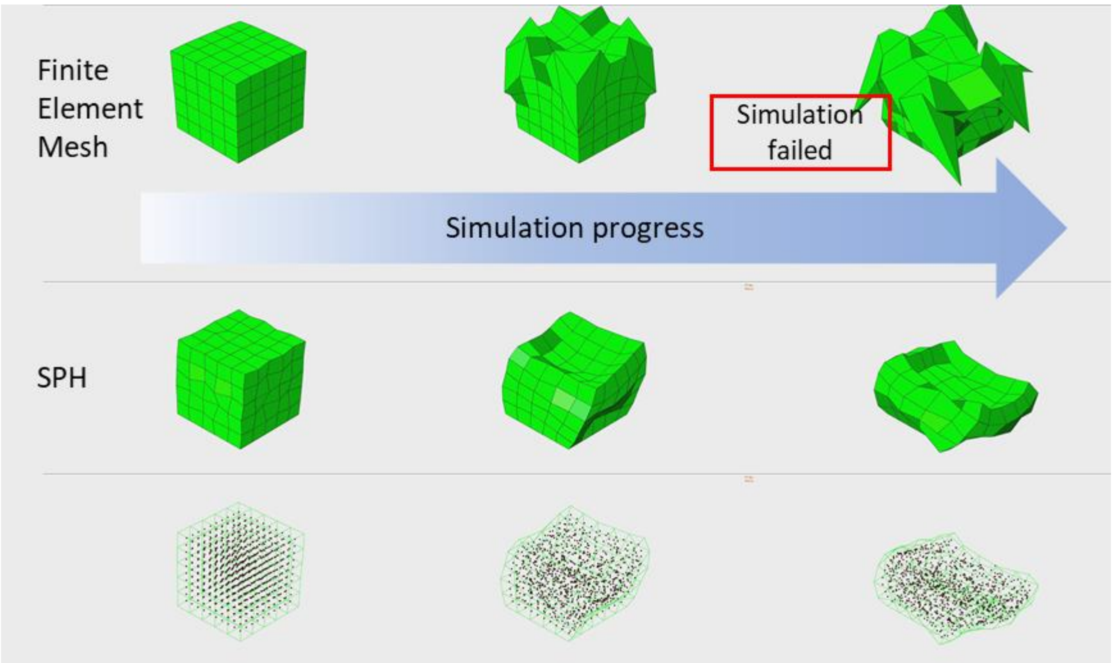


Figure 26. Modeling cube using either finite element mesh or SPH particles and the simulation results.

Interaction

The interaction property of different model parts, particularly the bony segments, can represent the level of complexity in the simulation. In this study, a general "hard" contact property was assigned to the model. Relative bone-to-bone displacements were frictionless to mimic the function of the cartilages. Given that loading on the plantar fascia is barely influenced by interphalangeal movements, the distal and proximal interphalangeal joints were fused in the lateral four toes. The contact between the foot and the ground plate was assigned with a friction coefficient of 0.6 (Zhang and Mak, 1999).

Boundary/loading conditions

The finite element foot model was assigned with the boundary and loading conditions generated by musculoskeletal modeling of the same participant (Figure 27). Concentric connector force was applied to the slip-ring connectors to simulate extrinsic foot muscle force. Three-dimensional ankle joint reaction force was loaded on the ankle surface of the talus. All force data were input in a tabulated time-series matrix to drive the finite element analysis. The whole foot model was initially placed at the position and orientation corresponding to the instant before foot strike. A pre-defined velocity was also conferred to the whole model. Gravity was enabled throughout the simulation steps using the force/mass ratio of 9.8.

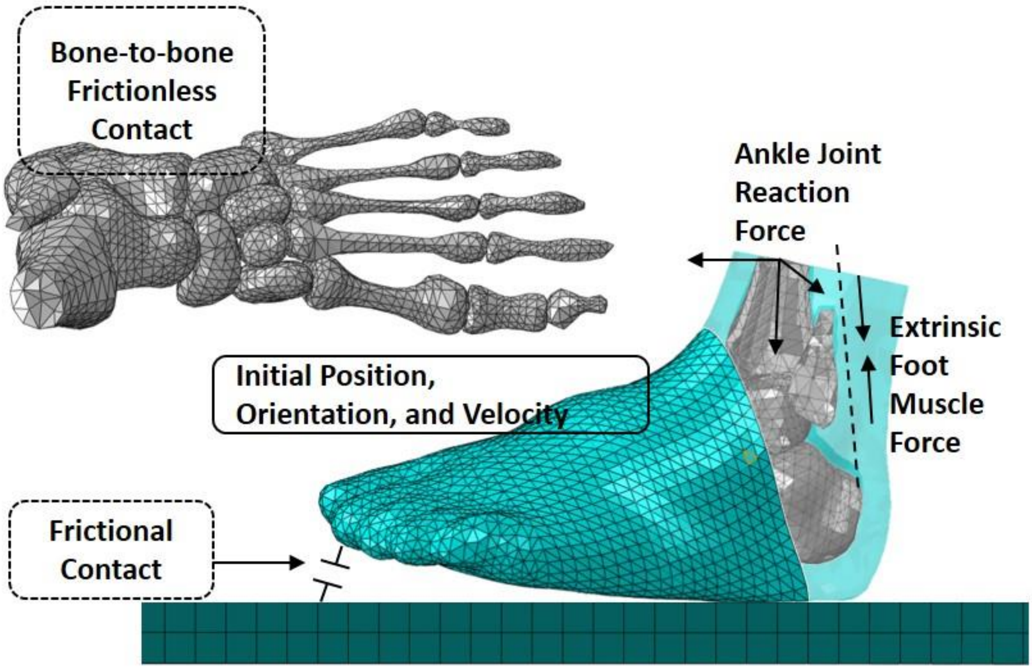


Figure 27. Boundary/loading conditions for the running simulations.

Contact properties are circled by the dashed box. Initial kinematic states are circled by the solid box. Arrows denote applied forces as well as the force directions.

Meshing

The accuracy of simulations is reportedly sensitive to the meshing density of the finite element model (Henninger et al., 2010). In general, meshing should be refined to maximally represent the underlying morphologies. However, a large mesh number also increases the modeling complexity as well as the costs of computational power, which consequently lengthens the time lag between the onset of the simulation and getting the results (Behforootan et al., 2017). Therefore, a mesh convergence analysis is usually performed to determine the optimal mesh size that satisfies the accuracy requirement while saves the model from computational expensiveness.

In this study, the averaged mesh size for the finite element foot model was determined through mesh convergence tests. Briefly, the foot model was repeatedly loaded under the same set of boundary/loading conditions while the mesh size was refined at 10% interval at each time (Figure 28). The primary outcome variable, e.g. strains on the plantar fascia, was compared between consecutive mesh refinements. The optimal mesh size was found when the change in outcome solution was less than 5% (Henninger et al., 2010).

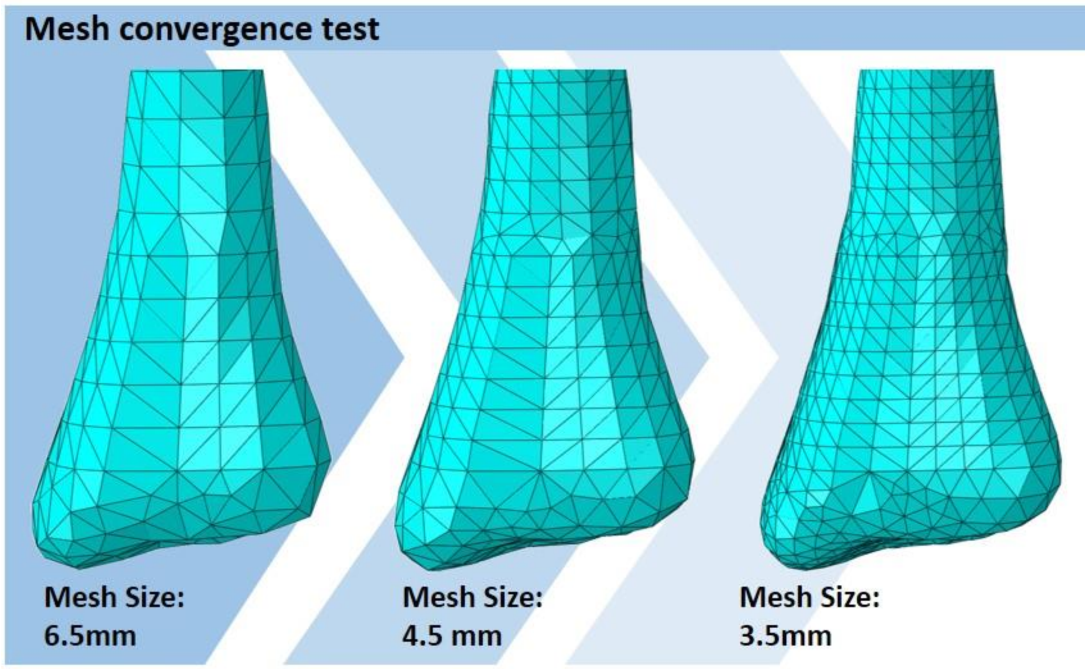


Figure 28. Mesh size refinement for the finite element models. From left to right, the general mesh size of the distal tibia is reduced from 6.5 mm to 3.5 mm.

Validation

Validation is a process in which calculations of a finite element model is compared to experimental data to determine the model's predictive capability (Anderson et al., 2007).

In study two, the established model was validated by reproducing the loading condition on a cadaveric foot from Sharkey's study (Sharkey et al., 1995). Briefly, the foot model was gradually heel-lifted by Achilles/flexor digitorum forces against the ground until the GRF reached 750 N. Meanwhile, either the peroneus brevis/longus or tibialis posterior was loaded by 200 N, depending on the condition simulated. The model-predicted and experiment-measured strains on the second metatarsal head were compared.

In study three, the foot model was upgraded by incorporating a three-dimensional solid unit that represented the plantar fascia. The model was validated again using the loading conditions of another cadaveric study (Clark et al., 2009). More specifically, the foot model was fixed at proximal tibia and constantly loaded by 98.1 N on the plantar forefoot, which imitated a plantar fascia stretching exercise. Strains on selected regions of the fascia band approximal to its origin were reported. Likewise, the predicted and measured strain values were compared.

For both validation procedures, the agreement between our simulations and the cadaveric studies was tested by linear regression. The regression parameters (slope, intercept, R2, and RMSE%) were calculated to demonstrate the accuracy level. The percentile root mean square error (RMSE%) was expressed as percentages of the peak of the absolute values of strains measured under each loading condition (Gray et al., 2008). Statistics were performed in SPSS (Version 19.0, IBM, Armonk, USA) at a significance level of 0.05.

Figure 29A (condition of peroneus brevis/longus activated) and Figure 29B (condition of tibialis posterior activated) shows the regression analysis between the predicted and measured second metatarsal strain under the heel-lift condition (Sharkey et al., 1995). The Pearson correlation coefficients were between 0.925 and 0.996 (p < 0.001, slope: 0.644–0.913, intercept: 0.069–1.602), indicating a good agreement. Figure 29C shows the validation results of the predicted and measured fascial strains under the loading condition of plantar fascia stretching exercise (Clark et al., 2009). The Pearson correlation coefficient was 0.954 (p < 0.001) with a slope of 1.050 and an intercept of 0.174.

To the best knowledge, there is not a statistical standard that quantifies the accuracy level for a finite element model at the time of this research. Instead, a previous study (Gray et al., 2008) suggested to demonstrate the model's accuracy by comparing the linear regression parameters to those of previous studies focusing on the similar modeling areas, e.g. finite element model (Table 1). The results showed that all parameters (slope, R2 of regression line and RMSE%) of this study were within the range of those published, indicating a satisfied level of model accuracy and a congruent agreement in the outcomes between simulations and experiments in the study (Gray et al., 2008). The slight differences in the absolute values of RMSE% among research could be attributed to the variances in model geometries, boundary/loading conditions, and modeling assumptions, which was acceptable considering the complexity of current foot model in the validation procedure.

In the study, the foot model was not validated by comparing measurable variables, e.g. plantar pressure, between simulations and experiments conducted on the recruited participant. The reason was two-fold. For one, to record the real-time plantar pressure during running, an insole system should usually be installed in conjunction with the usage of proper footwear, which could not be included in the modeling procedure in the study. Shoes were frequently reported to affect running biomechanics (Morio et al., 2009; Murley et al., 2009) and thus introduce a confounding factor to the results. Because the condition of footwear could not be replicated in the simulation, it was infeasible to validate the foot model using plantar pressure. Alternatively, the insole system can be adhered to the plantar foot by means of tapes or glue. However, this approach could change the attributes of the foot/insole interface, e.g. generating negative pressure values, and produce low-fidelity measurement data for reliable model validation. Another reason for not using plantar pressure for model validation was that, plantar pressure is a state of external loading on the foot surface and it does not necessarily reflect the profile of internal loading on the musculoskeletal structures (Williams, 2017). Its efficacy in validating a foot model specific for plantar fascia was thus questionable.

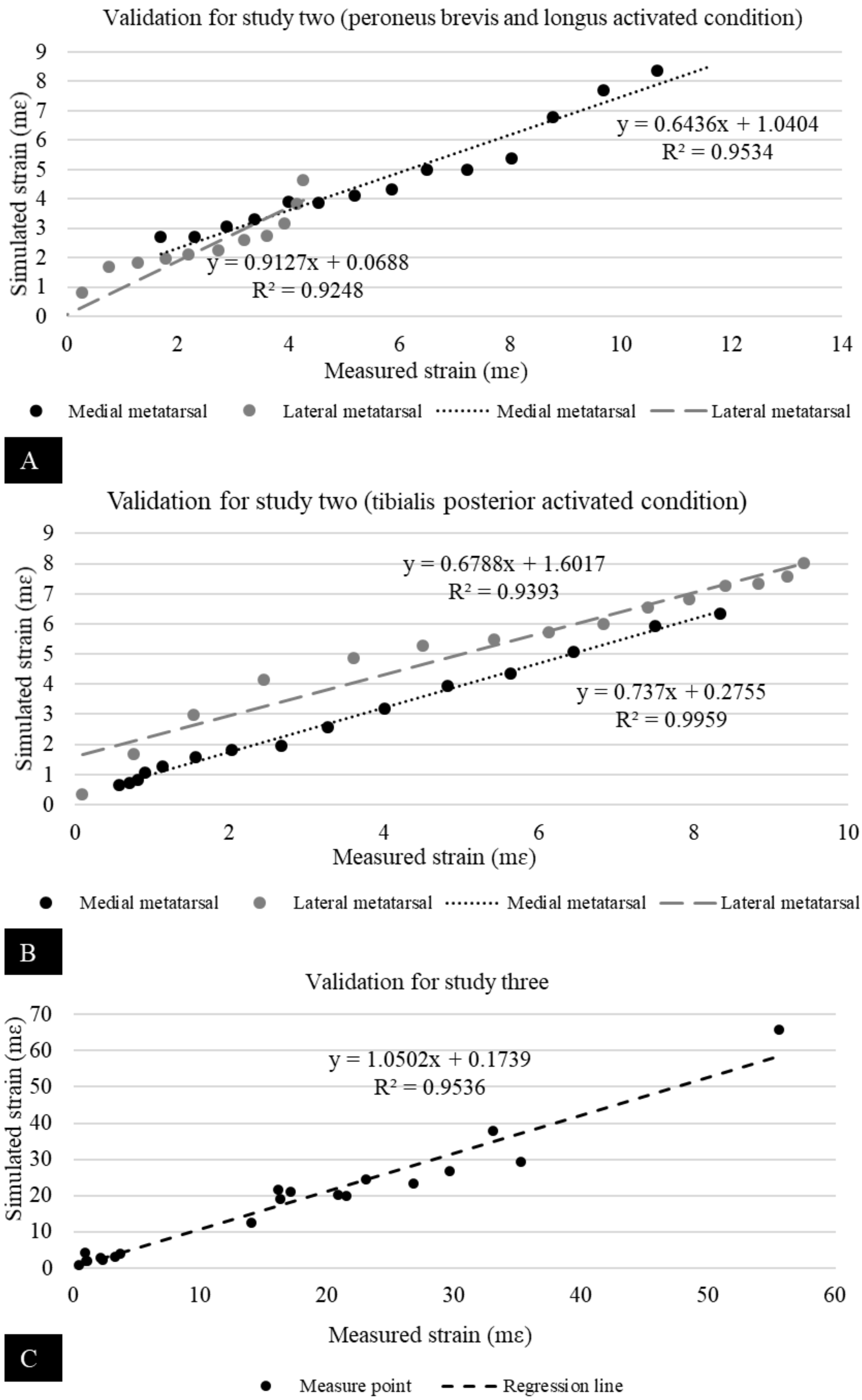


Figure 29. Validation of the finite element foot model by comparing the predicted strain

with existing literature.

(A-B) Linear regression of the second metatarsal bone strain between finite element prediction and existing cadaveric study (Sharkey et al., 1995) under a heel-lift condition with (A) peroneus brevis/longus activated or (B) tibialis posterior activated. Each dot denotes the measured and simulated strain value of one measurement point as depicted in Sharkey's study. Parameters of the four trendlines (p < 0.001, R2: 0.925–0.996, slope: 0.644–0.913, intercept: 0.069–1.602) indicate a good agreement between the experimental outcomes and model estimation; (C) linear regression of fascial strain between finite element prediction and existing cadaveric study (Clark et al., 2009). Each dot denotes the simulated and measured strain value of selected fascial regions as depicted in Clark's study. Parameters of the trendline (p < 0.001, R2: 0.954, slope: 1.050, intercept: 0.174) indicated a good agreement between the model estimates and experimental outcomes.

Table 1. Comparisons of the regression parameters of this study with those of other publications on finite element models as an indication of model accuracy level.

	***p*-value**	**Regression line slop**	**R square**	**RMSE%**
This study (metatarsal strain)	< 0.001	0.644–0.913	0.925–0.996	10.08–15.80%
This study (fascial strain)	< 0.001	1.050	0.954	5.55%
Kiapour et al. (ACL strain)	< 0.0005	-	0.792–0.865	9.31–13.04%

	***p*-value**	**Regression line slop**	**R square**	**RMSE%**
Kiapour et al. (MCL strain)	< 0.0005	-	0.941–0.960	4.50–6.31%
Gray et al. (tibia strain)	-	1.018–1.246	0.961–0.994	6.0–9.0%
Taddei et al. (femur strain)	-	1.010	0.910	8.6%

ACL: anterior cruciate ligament; MCL: medial collateral ligament.

CHAPTER 4 PLANTAR FASCIA SHEAR WAVE ELASTICITY IN RUNNERS USING REARFOOT STRIKE AND FOREFOOT STRIKE

4.1. Summary of the study

Most of the recreational runners nowadays are rearfoot strikers because modern footwear with cushioning in the midsole allows them to land comfortably on the heel. Most recently, there is a growing trend among the runners to advocate a “natural” running form, which is partially equivalent to running in barefoot or minimalist shoes that provide limited supports to the foot. Research has observed that, to accommodate the biomechanics of natural running, runners usually switch from rearfoot strike to forefoot strike. Forefoot strike is found to better attenuate impacts of footsteps by increasing the mechanical workload on the ankle and plantar foot. Scholars thus speculate a potentially higher risk of foot problems, e.g., plantar fasciitis, involved in forefoot strike. Up to date, this concern remains largely theoretical upon a scarce of supportive evidence from the clinic.

Previous studies have identified a reduced elasticity in tissues traumatized by repeated tensile force. Given this relationship between tissue loading and material property, this study sought to use ultrasound elastography to assess and compare the shear wave elasticity of the plantar fascia between rearfoot strikers and forefoot strikers. A total of 35 recreational runners (21 rearfoot strikers and 14 forefoot strikers), who were free of lower limb injuries, diseases, and foot deformities prior to the experiments, were recruited from a local running club. Their foot arch types and foot strike patterns were confirmed through the measured plantar pressure during treadmill running trials. The B-Mode ultrasound images and shear wave elastographic images of the plantar fascia were collected from each runner. Two independent investigators reviewed the images and examined the plantar fascia qualitatively and quantitatively.

The results of intraclass correlation coefficients (ICC) demonstrated an overall good agreement between the investigators in the image review outcomes (ICC:0.96–0.98, κ: 0.89). There were no significant differences in the fascial thickness ($p = 0.490$) and hypoechogenicity on the gray-scale images ($p = 0.542$) between the two groups. Shear wave elastography showed that forefoot strikers exhibited reduced plantar fascia elasticity compared to rearfoot strikers ($p = 0.012$, Cohen's d = 0.91).

A plantar fascia with reduced elasticity is less resistant to strain under loading. Tissue overstrain is frequently related to the incidence of plantar fasciitis, which could be a warning signal for the forefoot strikers in this study. While further investigations are needed to ascertain the underlying pathomechanics, e.g., increased loading on the fascial band, runners using forefoot strike were encouraged to enhance their foot strength for better protection of the plantar fascia.

4.2. Introduction

Transitioning to a forefoot strike pattern has received the most attention among the runners as it has the potential of reducing impact peak of the vertical GRF and eliminating other potential contributors to running-related injuries (Crowell and Davis, 2011; Shih et al., 2013; Williams et al., 2012). However, adoption of a forefoot strike pattern may have negative consequences for runners in terms of changing the loading status of the plantar fascia (Lieberman et al., 2010). Forefoot strike was speculated to induce a higher tensile force on the plantar fascia than rearfoot strike (Chen et al., 2019b). Repeated overload has been commonly considered as the primary cause of microtears in fascia and a contributor to plantar fasciitis (Wearing et al., 2006).

Ultrasound imaging is an effective tool to examine the plantar fascia and has been widely used to assist diagnosis of plantar fasciitis. Histologically, plantar fasciitis is a composite result of fiber microtears, collagen degeneration, chronic inflammation, and calcification caused by repetitive overstrain (Wearing et al., 2006), which underlies the typical appearances of plantar fasciitis on the conventional B-mode ultrasound images: thickened plantar fascia and a diffuse hypoechoic area within the fascia band (McMillan et al., 2009). Hypoechoic change in the plantar fascia usually presents as a loss of the normal fibrillar pattern on the gray-scale images (Kim et al., 2016). However, research showed that not all patients with plantar fasciitis exhibited these changes in tissue morphology and hypoechogenicity (Kapoor et al., 2010; Wu et al., 2012). Recently, ultrasound shear wave elastography has emerged as a novel imaging technique that can detect early-stage plantar fasciitis through the assessment of tissue elasticity (Sabir et al., 2005). Shear wave elastography can autogenerate and track the transient shear waves propagating in the tissues, whereby the shear wave velocity (SWV) is proportional to the tissue elasticity (Nowicki and Dobruch-Sobczak, 2016). Reduced elasticity was frequently related to tissue rupture caused by repeated loading (Eby et al., 2013; Zhang and Fu, 2013) and subsequent plantar fasciitis (Wu et al., 2011). The use of elastography may help identify signs of negative adaptation to forefoot striking via a loss of elasticity in the fascial band. Together with the already available fascial thickness and hypoechogenicity measurements, this information could be valuable for the runners to better understand their foot health.

Therefore, the purpose of this study was to utilize the ultrasound shear wave elastography to document plantar fascia elasticity in runners using forefoot strike patterns. Since the majority of recreational runners are rearfoot strikers (de Almeida et al., 2015; Larson et al., 2011), we used this group as controls. We hypothesized that forefoot strikers would exhibit increased fascial thickness, increased hypoechogenicity, and reduced plantar fascia elasticity compared to rearfoot strikers. The results of this study may provide greater insights of whether there were altered material properties of the plantar fascia involved in forefoot strike running and help to guide training programming, given that tissue strength can influence its injury risk.

4.3. Methods

4.3.1. Participants

A pilot study with two-independent-group design (2-tailed test) was conducted to estimate the proper sample size. Statistical power was set at the level of $\alpha = 0.05$, $\beta = 0.2$, and an assumed effect size of 0.54. Data of the first 10 eligible participants (6 rearfoot strikers and 4 forefoot strikers) recruited in the study were used for the sample size calculation. The ratio of rearfoot-to-forefoot-striker was therefore 3:2 based on their availability and kept consistent throughout the research. The primary outcome variable was mean SWV of the plantar fascia (5.11 ± 0.30 m/s for rearfoot strikers and 4.63 ± 0.32 m/s for forefoot strikers). The results showed that, at the preset statistic level, the minimum participant number for rearfoot striker group and forefoot striker group was 9 and 6 respectively.

Finally, a total of 35 recreational runners (21 rearfoot strikers and 14 forefoot strikers) were included in the actual experiment. The runners' foot strike patterns were later confirmed by undergoing an instrumental treadmill session. The experimental protocol was approved by the University Institutional Review Board (IRB NO.: HUM00149062). Participants were recruited from the local running community via recruitment flyers and word of mouth. Inclusion criteria were: (i) aged between 18 and 35 yr.; (ii) currently had a weekly mileage of at least 15 km; (iii) had a running experience of at least 2 years prior to the experiment; (iv) originally rearfoot striker at the beginning of her/his running career; (v) had no ongoing symptoms or injuries of the lower limbs at the time of entry, and (vi) had no musculoskeletal diseases, such as rheumatoid disorders. Runners were excluded if they: (i) had an abnormal foot arch (i.e., pes planus or pes cavus), which was later confirmed by measures of foot arch index in the study (Cavanagh and Rodgers, 1987); (ii) modified foot strike pattern within the 6 months prior to the experiments; (iii) currently used orthotics, prosthetic devices, or footwear with motion control function; (iv) received lower limb surgeries in the past 6 months prior to the experiments; or (v) habitually ran barefoot. Runners with a foot arch index ≥ 0.26 (pes planus) or ≤ 0.21 (pes cavus) were excluded from the study (Wong et al., 2012). All participants were fully informed of the research procedures and provided informed consent prior to participation. Each participant also completed a questionnaire that collected information regarding basic anthropometry and running regime, such as running experiences, usual pacing, days of running per week, weekly mileage, and etc.

4.3.2. Experimental Procedure

The study consisted of two sessions with a 5-minute rest interval: a treadmill session and a shear wave elastography session. The purpose of the treadmill session was to confirm foot arch type and foot strike pattern for the participants. To minimize the influences of physical activities on imaging quality (Skou et al., 2012), we performed the shear wave elastographic measurements to the participants prior to the treadmill session.

Shear wave elastography session

Shear wave elastography was performed using the Aixplorer ultrasonic scanner (Supersonic Imagine, Aix-en-Provence, France; software version 5). The participants laid prone with the knee fully extended, foot suspended at the bedside, and the ankle slightly plantarflexed in a resting position. The position for examination was kept consistent across all participants in the study. Gentle compression was applied to the heel with a linear array transducer (7–14 MHz). In the longitudinal view, the color-coded image of the shear wave (a rectangle 2 cm × 4 cm) was superimposed over the B-mode ultrasound image obtained simultaneously (Wu et al., 2015). The following parameters were set and kept consistent throughout the measurements: “Muscle” probe mode, “Penetration,” “High Definition,” “Contrast” = 65%, and “Gain” = 100%. The region of interest was initially positioned at the anterior edge of the inferior calcaneal border (Lee et al., 2014) and slightly moved towards the toe direction if shear wave signals were difficult to acquire. Ten images were collected for each participant. All ultrasonic measurements were performed by a same investigator. The primary variable was the mean SWV of the plantar fascia, which represented the degree of tissue stiffness. The fascial thickness and hypoechogenicity on the gray-scale images were also examined as secondary outcomes.

The ultrasound images were analyzed qualitatively and quantitatively by two investigators, who were blinded to the grouping results. Plantar fascia thickness was measured as the vertical distance between the anterior edge of the inferior calcaneal border and the inferior border of the plantar fascia (Ríos-Díaz et al., 2015) (Figure 30A). The hypoechogenicity of the plantar fascia was assessed based on a grading scheme (Archambault et al., 1998): grade I: normal appearance (parallel margins, homogeneous echogenicity); grade II: enlarged structure (bowed margins, homogeneous echogenicity); and grade III: hypo-echoic area with or without enlargement.

Shear wave elastography images were analyzed using an established algorithm (Lee et al., 2015; Leonardis et al., 2017). The algorithm provides quality control to ensure the elastography measurement only includes pixels with sufficient quality in the final calculation of SWV. A customized Matlab (R2014a, MathWorks, Natick, MA) code extracting both SWV and quality map from the elastography images. The quality map, which was provided by Supersonic Imaging, reflects the manufacturer’s calculation regarding the cross-correlation of shear waves propagating in the tissue. In the map, each pixel of the image is assigned with a number that denotes the accuracy of the SWV measures. The number ranges from 0 to 1, with 1 representing the best quality and 0 representing the worse. To start with the algorithm, we manually cropped the region of interest corresponding to the plantar fascia band from the elastography image (Figure 30B). The region of interest was approximately 2 cm in length along the fascial band (kept consistent for all participants), and its size could vary slightly to accommodate the individual variations in fascial thickness. Our algorithm computed the mean SWV of pixels with a quality number > 0.7 from the cropped image (Leonardis et al., 2017). On average, the percentage of qualified pixels within the cropped images was 45.8% for rearfoot strikers and 45.7% for forefoot strikers. We found no evidence of data saturation in any of the images. SWV was averaged across 10 images for each participant.

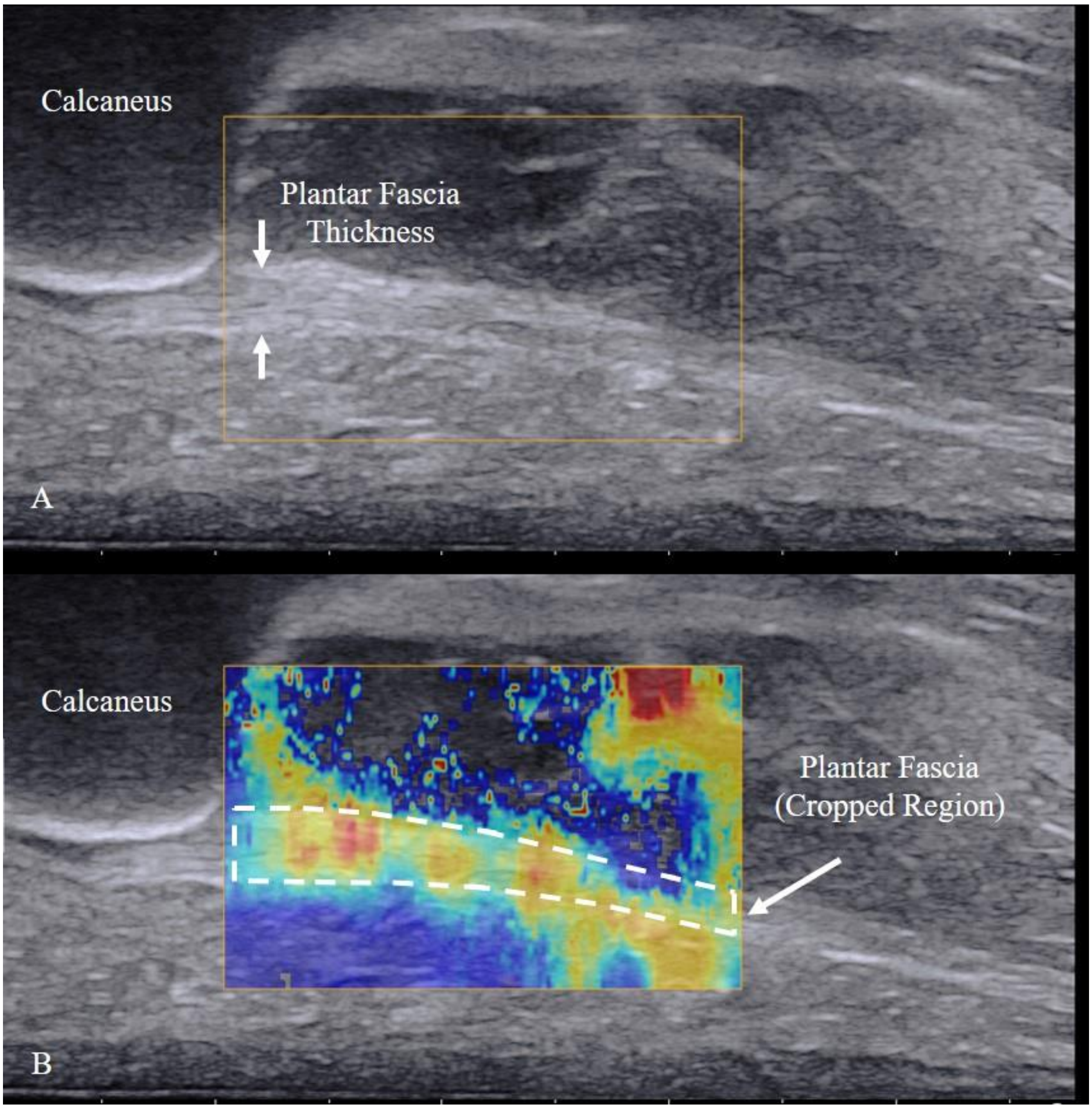


Figure 30. The ultrasound images of one representative participant. A: the longitudinal B-mode image, B: the color-coded shear wave elastographic image superimposed on the B-

mode image.

Treadmill session

Participants stood barefoot on a pressure-sensing treadmill (h/p cosmos Quasar Medical treadmill, h/p cosmos, Nussdorf-Traunstein, Germany) and performed a 6-second single leg stance on right and then left leg. After the standing trial, participants performed a 1-min walking trial at a set speed (4 km/h). Following the walking trial, participants ran on the treadmill for 3 min using a self-selected speed in their own running shoes. No instruction on specific running form (e.g., specific foot strike pattern) was given to the participants.

Foot arch type was quantified using the foot arch index (Cavanagh and Rodgers, 1987). Briefly, the measured foot pressure map was equally divided into three sections along its longitudinal axis (i.e., line intersecting the heel center and second toe), which represented the heel, midfoot, and forefoot regions. Foot arch index was calculated as the percentage of the midfoot area to the total footprint area. A mean foot arch index was averaged through the 6-s standing trial. Runners with an arch index of 0.21–0.26 were considered having the normal foot arch type (Wong et al., 2012).

Foot strike patterns were quantified by the strike index as modified by Graf et al. (Graf et al., 2013). Strike index was defined as the percentile location of the plantar pressure center relative to the full footprint length (Figure 31). The toe region of the footprint was excluded as this modification has been shown to increase the validity of identifying a forefoot strike pattern (Graf et al., 2013). A strike index of 0–33.3% was classified as a rearfoot strike pattern while 66.6–100% was classified as forefoot strike pattern (Lieberman et al., 2010). Because forefoot strikers might produce incomplete footprints during running, the actual footprint of each participant was contoured from the walking trial based on the fact that all of them used a heel-to-toe walking gait, regardless of foot strike patterns used during running. The actual full footprints were then input and re-located in the running trial by aligning to the front edge of the foot pressure profile along the belt, assuming that, for either rearfoot strike or forefoot strike, the foot's tip always touched the ground at push-off and did not drift during the stance phase (Santuz et al., 2016). For each participant, strike index was averaged across the first, middle, and last 10 strides of the running trials.

For the treadmill session, plantar pressure was recorded using the proprietary software (myoRESEARCH MR3.12, Noraxon, Scottsdale, USA) at a sampling rate of 120 Hz. Raw data were exported and processed by a custom-code (Matlab R2014a, MathWorks, Natick, MA).

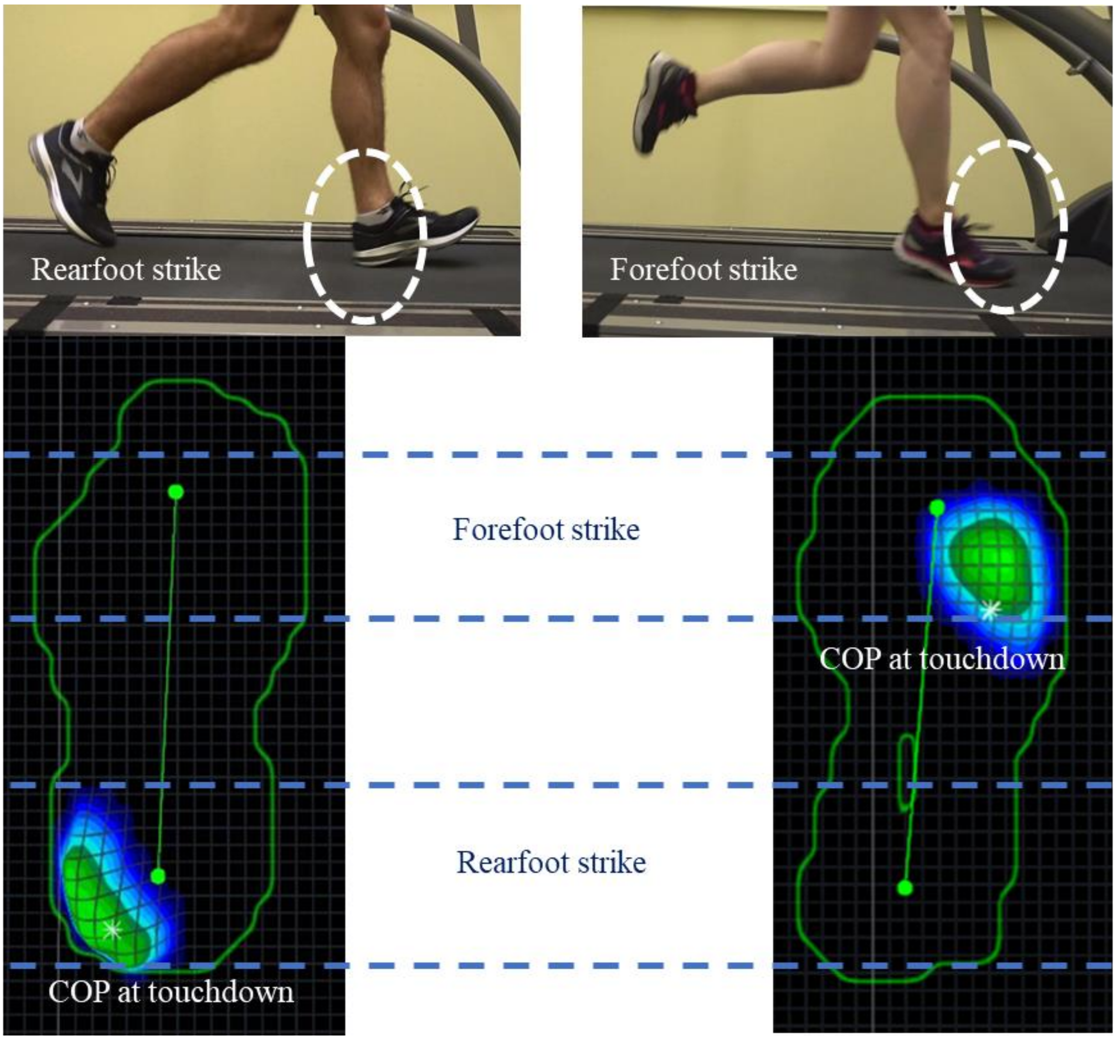


Figure 31. Classification of the runner's foot strike pattern using the data of plantar pressure collected from the treadmill running trials.

The plantar pressure center at touchdown fell within the proximal one-third of the footprint in rearfoot strike and the distal one-third of the footprint in forefoot strike. COP: center of pressure.

4.3.3. Statistics

Statistics was performed using SPSS (V16.0, SPSS, Inc., Chicago, Illinois, USA) with the global significance level set at 0.05. Data were assessed for distribution normality (i.e., Kolmogorov-Smirnov test) and homogeneity of variance (i.e., Levene's test). Numerical outcome variables were compared between the rearfoot strike and forefoot strike groups using the independent Student's t-test or the Mann-Whitney U test—if the assumptions of normal distribution or homogeneity of variance were violated. An effect size (Cohen's *d* for independent Student's t-test and *r* for Mann-Whitney U test) was demonstrated for each comparison pair (Cohen, 1988). A chi-square (X^2) test was used to compare the grade distribution of plantar fascia hypoechogenicity. The effect size (Cramer's V) was also reported (Cohen, 1988).

Intra-subject and inter-subject variabilities of plantar fascia SWV were expressed as coefficient of variation percentage (CV%) (standard deviation/mean × 100). The results were classified as very good, if CV ≤ 10%; good, if 10% < CV ≤ 25%; moderate, if 25% < CV ≤ 35%; and poor, if CV > 35% (Alajbeg et al., 2017).

ICC and Kappa value (κ) were calculated to determine the inter-rater agreement in the outcomes of plantar fascia thickness/SWE and hypoechogenicity grading, respectively. Both

ICC and κ values were reported along with the associated 95% confidence interval. The level of agreement was categorized as: poor, 0 to 0.20; fair, 0.21 to 0.40; moderate, 0.41 to 0.60; substantial, 0.61 to 0.80; and almost perfect, 0.81 to 1.00 (Shrout and Fleiss, 1979).

Table 2. Baseline characteristics and running regimen of the participants (mean ± standard deviation).

	Rearfoot Strikers	**Forefoot Strikers**	***p*-value**	**Effect size**
Age (years) [a]	25.14 ± 4.64	26.85 ± 4.50	0.287	0.36
Height (m) [b]	1.71 ± 0.08	1.74 ± 0.13	0.377	0.14
Body Mass (kg) [b]	65.65 ± 9.70	65.57 ± 9.53	0.853	0.03
Body Mass Index (%) [a]	22.32 ± 2.31	21.42 ± 1.29	0.196	0.44
Shoes' size (US) [b]	9.17 ± 1.19	9.61 ± 1.80	0.583	0.08
Running experience (years) [b]	11.00 ± 5.72	10.39 ± 6.86	0.727	0.05
Weekly running days [a]	4.43 ± 1.43	4.21 ± 1.22	0.648	0.15
Single-run mileage (km) [b]	9.83 ± 3.61	9.89 ± 3.52	0.987	0.01

	Rearfoot Strikers	Forefoot Strikers	*p*-value	Effect size
Weekly mileage (km) [b]	44.58 ± 24.53	43.95 ± 25.73	0.934	0.01
Usual running speed (m/s) [a]	11.45 ± 1.54	11.87 ± 1.25	0.411	0.28
Foot arch index [a]	0.23 ± 0.01	0.23 ± 0.01	0.713	0.13
Strike index (%) [b]	26.38 ± 7.45	86.79 ± 8.57	**< 0.001**	7.12

a: tested by the Student's t-test, effect size: Cohen's *d*; b: tested by the Mann-Whitney U test, effect size: *r*.

4.4. Results

The baseline characteristics including general anthropometries and running regimen were similar between the forefoot strikers and rearfoot strikers (Table 2). Analysis of the B-mode images showed that the majority (65.7%) of the participants had Grade I plantar fascia, while only 3 cases of Grade III plantar fascia presented (Table 3). There were no significant differences in grading ($p = 0.542$, Cramer's V = 0.19) and plantar fascia thickness ($p = 0.490$, $r = 0.10$) between the two groups. Forefoot strikers exhibited significantly lower plantar fascia SWV compared to the rearfoot strikers ($p = 0.012$, Cohen's $d = 0.91$).

A low intra-subject variability was observed in the measurements of plantar fascia SWV, with CV for all recruited participants ranging from 2.2% to 13.5%. Tests of inter-subject variability also reported congruent results for the rearfoot strike group (CV: 9.0%) and forefoot strike group (CV: 10.8%). There were overall good inter-rater agreements in the results of plantar fascia thickness (ICC: 0.96, 95% CI: 0.91, 0.98), plantar fascia SWV (ICC: 0.98, 95% CI: 0.95, 0.99), and hypoechogenicity grading (κ: 0.89, 95% CI: 0.74, 1.04).

Table 3. Outcomes of the ultrasound measurements (mean ± standard deviation).

		Rearfoot Strikers	**Forefoot Strikers**	***p*-value**	**Effect size**
Hypoechogenicity [c]	Grade I	15	8	0.542	0.19
	Grade II	5	4		
	Grade III	1	2		
Plantar fascia thickness (10^{-3} m) [b]		3.08 ± 0.35	3.41 ± 0.89	0.490	0.10
Plantar fascia SWV* (m/s) [a]		6.67 ± 0.48	6.20 ± 0.56	**0.012**	0.91

*a: tested by the Student's t-test, effect size: Cohen's *d*; b: tested by the Mann-Whitney U test, effect size: *r*; c: tested by the chi-square test, effect size: Cramer's V.

4.5. Discussion

This study performed elastographic measurements of the plantar fascia to compare the results between runners using rearfoot strike and forefoot strike patterns. In partial support of our hypothesis, forefoot strikers produced significantly lower plantar fascia elasticity than rearfoot strikers. However, the fascial thickness and hypoechogenicity grade were similar between the two conditions.

Our results echoed previous findings that the outcomes of conventional ultrasound imaging and elastographic measurement to the same plantar fascia can vary due to their different sensitivity in identifying minor intrafascial changes (Kapoor et al., 2010). Generally, a plantar fascia more than 4 mm in the thickness is considered to relate to plantar fasciitis (Chen et al., 2013). Previous studies reported a fascia thickness of 2.6–3.2 mm for the healthy controls and 3.9–5.0 mm for patients with plantar fasciitis (Abdel-Wahab et al., 2008; Chen et al., 2013; Ríos-Díaz et al., 2015; Wu et al., 2015). Our measurements showed that most runners, regardless of foot strike pattern, had a fascia thickness that fell within the normal range. In addition, most runners had the grade I plantar fascia in hypoechogenicity. While three cases of grade III were identified, these participants reported no symptoms during participation and could continue training without complaints. The contrast in outcomes between conventional ultrasound imaging and elastography supported a statement that, shear wave elastography had the potential to identify the "pre-clinical" cases—individuals who are exhibiting biological tissue changes but have yet to report positive symptoms (De Zordo et al., 2009), though further study is required to determine the cut-off value for differentiating injured persons from the healthy.

Previously, our image processing algorithm demonstrated a good inter-rater reliability for muscle tissue evaluation (Leonardis et al., 2017). Similar strong inter-rater reliability was found with the plantar fascia. Previous studies using strain elastography to examine the plantar fascia commonly adopted a qualitative method to analyze the images (Lee et al., 2014; Ríos-Díaz et al., 2015; Wu et al., 2015) (i.e., comparing different colors' distribution frequency in the color histogram). Those methods could not reflect the fascia elasticity quantitatively or facilitate between-study comparisons. Our measurements were quantitative, and the results supported the statement that the plantar fascia possessed an elasticity similar to human tendon and ligament (Ryu and Jeong, 2017; Wearing et al., 2006b). Our measured SWE values (6.20–6.67 m/s) were slightly higher than those of Zhang's study (approximately 5.52 m/s) (Zhang et al., 2014) but lower than Shiotani's results (8.00–9.51 m/s) (Shiotani et al., 2019). The differences may be attributed to the variances in experimental setups. Zhang et al. (Zhang et al., 2014) did not control the imaging quality in their elastographic measurements. Including pixels with poor SWV measures would underestimate tissue elasticity due to the substantial noise in the echo (Barr and Zhang, 2015). Shiotani and coworkers (Shiotani et al., 2019) only recruited physical-inactive persons, and they likely acquired the tissue elasticity from a more tightened plantar fascia since the ankle was fixed at the neutral position for examination.

The present study showed that the plantar fascia SWV was 6.46% lower in forefoot strikers compared to rearfoot strikers. Though that degree of reduction may not be clinically significant because the runners had no symptoms, it suggested that forefoot strikers had a less elastic plantar fascia than rearfoot strikers. Many mechanical factors in running can cause elasticity alterations of the plantar fascia. Most frequently they are associated with sports-related overuse (Abdel-Wahab et al., 2008). Since the baseline characteristics and running regimens were similar between the two groups, the reduced fascial elasticity in forefoot strikers may have resulted from their different running biomechanics. As reported, the most influential changes in forefoot strike may be the increased cadence (loading frequency) (Baggaley et al., 2017; Goss and Gross, 2013; Kulmala et al., 2013) and larger plantar fascia loading (Chen et al., 2019b; McDonald et al., 2016). In addition, a great number of forefoot strikers could not recall the precise date on which they transitioned from rearfoot strike to forefoot strike. As a result, their running experiences with forefoot strike were likely less than the total running years as they reported, which may underestimate the effects of forefoot strike on the fascial elasticity and the subsequent differences in comparison to rearfoot strikers. All these factors, in combination, could play a role in the outcomes of the elastographic measurements. Due to the retrospective nature of the current study, it is not possible to conclude a causal link between forefoot strike and reduced plantar fascia elasticity, and the underlying mechanism for the elasticity difference. For more definitive insights, a prospective, controlled study must be conducted in those who plan to modify their foot strike patterns (Hamill and Gruber, 2017).

The present study may provide valuable information for the runner community. A plantar fascia with reduced elasticity is less resistant to strain and may render the runners to injuries caused by tissue overstretch (e.g., plantar fasciitis). It is currently not possible to determine the extent to which the loss of fascial elasticity in forefoot strikers would affect their running biomechanics or their foot health, unless a longitudinal follow-up study is conducted. Nonetheless, results from the current study suggest that forefoot strikers may focus on strengthening their foot for better protection of the plantar fascia, which has also been advocated by others (Chen et al., 2016; Lynn et al., 2012). Research has found that forefoot strike running could induce a higher loading on the foot arch and increasingly tensioned the plantar connective tissues (Kelly et al., 2018; McDonald et al., 2016). As a part of the loading-sharing system of the foot arch (Kirby, 2017), intrinsic foot muscles can shield the total arch loading for the plantar fascia during running. Habitual rearfoot strikers who tended to adopt forefoot strike are suggested to train on their foot muscle strength (Huffer et al., 2017), to take the strain off the plantar fascia. This may also be advisable for runners already using forefoot strike but showing signs of less elastic plantar fascia as presented in the current study.

This study has several limitations. First, the percentage of qualified pixels within the cropped image was relatively low. The plantar fascia is a deep structure overlaid by other soft tissues, and its visualization is usually not ideal on elastography (Ríos-Díaz et al., 2015). This is a problem commonly faced by research applying ultrasound-based modality. Second, since we did not measure foot muscle strength for the runners, the actual causes of the reduced elasticity in forefoot strikers remain unclear. Finally, the relation between SWV and tendinous elasticity (e.g., plantar fascia) is not fully understood. The plantar fascia is unidirectionally fibrous, and its varied thickness on the transverse plane could be a confounding factor influencing shear wave propagation and the value of SWE read by elastography (Brum et al., 2014; Helfenstein-Didier et al., 2016). Further investigation including measurement on the fascial morphology and mechanical property in different directions would be necessary for a better understanding of the nature of the plantar fascia.

CHAPTER 5 FOOT ARCH DEFORMATION AND PLANTAR FASCIA LOADING DURING RUNNING WITH REARFOOT STRIKE AND FOREFOOT STRIKE: A DYNAMIC FINITE ELEMENT ANALYSIS

5.1. Summary of the study

Recently forefoot strike has gained popularity among recreational runners because it is allegedly a softer landing style than the conventional rearfoot strike. Some runners are recommended to adopt forefoot strike as a mean of saving the lower limb joints from high impact shocks during running. In the meantime, concerns are also raised regarding the downsides of forefoot strike. Running with forefoot strike normally demands more work done by the calf muscles and ankle joint, which is speculated to increase mechanical burdens on the foot arch and the plantar fascia. However, the statement is marginally/barely supported by available evidence, nor is it verified by research.

Given the above situation, this computational study examined and compared foot arch deformation and plantar fascia loading in running with different foot strike techniques. A three-dimensional finite element foot model was reconstructed from the MRIs of a healthy runner. The model contained twenty bones, bulk soft tissue, ligaments, tendons, and the plantar fascia. The general mesh size of the model was determined through a mesh convergence test. Model validation was performed by replicating the loading conditions of two cadaveric studies using the foot model and comparing the results between simulation and experiment. The recruited runner performed several running trials using either rearfoot strike or forefoot strike. The kinematic and kinetic data collected via motion capture were converted to drive a subject-scaled musculoskeletal model, whereby variables of segmental kinematics, foot muscle force, and ankle joint reaction force were generated. These variables were then imported as the boundary/loading conditions to set-up the finite element analysis of the established foot model. Running with either of the two foot strike techniques was simulated using a dynamic solver.

The mesh convergence test showed that an average mesh size of 3.5 mm caused a < 5% changes in the outcome solution. Results of model validation indicated an overall good agreement between the model predictions and experimental measurements. Simulations of running demonstrated the capacity of the established foot model to reproduce running dynamics for both foot strike conditions. Compared to rearfoot strike, forefoot strike increased the foot arch height drop (2.06–9.12% higher), plantar connective tissues stress (18.28–200.11% higher), and plantar fascia tensile force (18.71–109.10% higher).

This study provided quantitative results to support the concerns over forefoot strike, regarding its negative influences on the biomechanics of the plantar fascia. Although it is currently difficult to determine the extent to which the increased foot arch deformation and plantar fascia loading in forefoot strike will jeopardize the foot health, forefoot strike appeared to pose more injury threats of plantar fasciitis to the runners. Cautions should be taken to the prescription of foot strike pattern modification for habitual rearfoot strikers. For runners willing to adopt forefoot strike, they were suggested to pay attention to training foot strength for better injury prevention.

5.2. Introduction

Running kinematics affected the mode of impact attenuation and force transmission associated with the risks of injury (Daoud et al., 2012). To better attenuate impact force, an increasing number of runners adopted a softer landing skill by transferring from rearfoot strike to forefoot strike (Altman and Davis, 2012a). Forefoot strike could change the lower limb loading form during stance, which tended to increase the mechanical burdens on the foot arch and the subsequent plantar connective tissues, e.g. the plantar fascia (Lieberman, 2012). Clinical studies have speculated the risks of plantar fasciitis involved in foot strike pattern modification (Daoud et al., 2012; Lieberman et al., 2010), in spite of that existing evidence is largely anecdotal.

Information of loading on the plantar connective tissues during running is essential to understand the relationship between foot strike patterns and plantar fasciitis. While in-vivo measurement of the plantar fascia force is usually difficult, computational modeling techniques can cater to the methodological needs. However, outcomes of existing simulation studies were inconclusive possibly due to some limitations inherent in the modeling details. Musculoskeletal model was robust to estimate the muscle forces that replicated human motion, but the assumption of using rigid bodies would hinder the constitution of soft tissues, such as the muscular and ligamental stabilizers of the foot arch (Bruening et al., 2012). Finite element model could predict segmental displacement and material deformation. Nevertheless, some existing studies implemented implicit solvers that ignored the large inertial moment of the bony structures generated in running (Behforootan et al., 2017). The load and compression on the soft tissue were heavily dependent on the momentum developed in transient and would be better predicted by an explicit solver (Chen and Lee, 2015). To increase the simulation accuracy, we presented a dynamic finite element analysis with running gait characteristics. The innovation of this study also laid in using a relative accuracy dynamic model in addressing the biomechanics of foot strike patterns.

Therefore, the aims of this study were to 1) establish a finite element foot model for dynamic simulation of running; 2) simulate rearfoot strike and forefoot strike running; 3) compare the foot arch deformation and loading on plantar connective tissues between rearfoot strike and forefoot strike. We hypothesized that forefoot strike would produce a higher foot arch drop, larger plantar ligament stress and plantar fascia tensile force than rearfoot strike.

5.3. Methods

5.3.1. General information

A healthy male aged 29 yr., 170 cm tall, and weighed 65 kg, was recruited for the study. He reported no musculoskeletal disorders, lower limb injuries, or orthopedic surgery history prior to the experiment. The participant was a habitual rearfoot striker with twelve years of running experience. Before the commencement of the study, the participant was fully informed of the research procedure and signed the consent form. The study was approved by the institution authority (IRB NO. HSEARS20170626003).

5.3.2. Equipment

A motion capture system with eight optical-based cameras (Vicon, Oxford Metrics Ltd., Oxford, United Kingdom) and four force platforms (OR6, AMTI, Watertown, USA) were used to measure kinematics and GRF. The data were sampled at 250 Hz and 1000 Hz respectively. The maker set for motion capture was configured to compile with the OpenSim full-body model (Rajagopal et al., 2016). Briefly, markers were affixed to the acromioclavicular joints, lateral/medial humeral epicondyles, radius/ulna styloid processes, posterior/anterior iliac spines, greater trochanters, lateral/medial femoral epicondyles, lateral/medial malleoli, calcaneal tuberosity and the base/head of the first and fifth metatarsals.

5.3.3. Experimental procedure

The participant was asked to run barefoot at 10 km/h (Cheung and Rainbow, 2014) with rearfoot strike and forefoot strike. He was given ample time to warm-up and allowed to rest for ten minutes between the conditions. For each foot strike condition, the participant ran through the motion capture volume in which his running speed was monitored by pairs of photoelectric cells placed 2.6 m apart along the runway (Hamill et al., 2014). During forefoot strike, the participant was instructed to land with the plantar ball area and lift the heel slightly above the ground during stance. The foot strike pattern was visually observed on-site and further confirmed by the foot strike index (Chen et al., 2016). For each condition, the kinematic and kinetic data of one representative trial was selected and processed for the input of subsequent computational simulation.

5.3.4. Musculoskeletal model

The musculoskeletal model (OpenSim, version 3.3, National Center for Simulation in Rehabilitation Research, Stanford, USA) was driven by the kinematic and kinetic data of the selected running trials. The generic model (Rajagopal et al., 2016), featuring 22 rigid-body segments, 37 degrees of freedom, and 80 musculotendonous units, was firstly scaled to accommodate the participant's anthropometry. Inverse kinematics was then solved and the dynamic inconsistency was reduced by fine adjustments to the model mass properties (Arnold et al., 2010). Muscle forces were estimated by the Computed Muscle Control module (Thelen, 2003). Joint reaction force and segmental kinematics were generated by the Analyse toolkit and output as the boundary/loading conditions for the finite element simulation.

5.3.5. Finite element model

Geometry acquisition and reconstruction

A series of MRIs were obtained from the participant's left leg which was fixed at the neutral position using a customized ankle-foot-orthosis (Wong et al., 2014). The 3.0T MRI scanner (GoldSeal Certified Signa HDxt, General Electric Company, Boston, USA) was configurated at T1 sequence, 1-mm slice interval, and a resolution of 0.625 mm pixel size. The images were segmented and processed by Mimics and 3-matics (version 19, Materialise, Leuven, Belgium). Totally twenty bones (including the distal portion of tibia and fibula) and the bulk soft tissue were reconstructed to three-dimensional solids. The bulk soft tissue was modelled as a cluster of SPH particles encapsulated in a shell unit that represented the periosteum (internal layer) and the skin (external layer). The second to fifth interphalangeal joints were fused for simplification. The intrinsic foot muscles and ligaments were modeled as truss units. A slip-ring type of connector was used to embody the function of the extrinsic foot muscles and plantar fascia. The attachments and pathways of the foot muscles, ligaments, and plantar fascia were informed by the human anatomy atlas (Gray, 2011). Details of material assignment and meshing for all model components are listed in Table 4. Overall, the bony parts, ligaments, and muscles were assumed linearly elastic except for that, a hyperelastic material property was conferred to the skin.

To determine an optimal mesh size for the whole foot model, a mesh convergence test was performed by loading the model in an experimental condition that mimicked the setups of a cadaveric study (Sharkey et al., 1995). The simulation was repeated with the general mesh size refined by 10% (from 5.9 mm to 3.2 mm) at each time. As revealed in the results, the optimal mesh size was 3.5 mm because it produced < 5% changes in the outcome solution (Henninger et al., 2010).

Table 4. Element type and material property assigned for the foot model components.

	Element	**Material property**	**Density**	**Poisson's ratio**	**Mesh count**	**Reference**
Skin	Linear triangular shell (S3R)	Hyper-elastic (first-order Ogden model, μ = 0.122 MPa, α = 18) Thickness: 2.0 mm	950 kg/m^3	N/A	4807	Pailler-Mattei et al., 2008
Bulk soft tissue	SPH particle (PC3D)	Linearly elastic (Young's modulus: 0.83 MPa for the plantar heel, 0.70 MPa for the plantar forefoot/toe, 0.67 MPa for the plantar midfoot, and 0.20 MPa for the rest)	950 kg/m^3	0.4	45008	Cheung and Zhang, 2005; Ledoux and Blevins, 2007

	Element	Material property	Density	Poisson's ratio	Mesh count	Reference
Periosteum	Linear triangular shell (S3R)	Linearly elastic (Young's modulus: 0.9 MPa) Thickness: 1.5 mm	1000 kg/m^3	0.4	11576	Uchiyama et al., 1998
Bone	Linear tetrahedral solid (C3D4)	Linearly elastic (Young's modulus: 17000 MPa)	1990 kg/m^3	0.3	18965	Bayraktar et al., 2004
Extrinsic foot muscles	Slip ring connector	Linearly elastic (stiffness: 157.4 N/mm)	1000 kg/m^3	N/A	N/A	Cook and McDonagh, 1996
Intrinsic foot muscles	Two-node truss (T3D2)	Linearly elastic (Young's modulus: 264.8 MPa) Cross-section area: 10 mm^2	1000 kg/m^3	0.4	24	Wong et al., 2016

	Element	Material property	Density	Poisson's ratio	Mesh count	Reference
Rearfoot ligaments	Two-node truss (T3D2)	Linearly elastic (Young's modulus: 100-320 MPa) Cross-section area: 7.1-256 mm^2	1000 kg/m^3	0.4	20	Davis et al., 1996; Kura et al., 2001; Milz et al., 1998; Siegler et al., 1988
Other ligaments	Two-node truss (T3D2)	Linearly elastic (Young's modulus: 264.8 MPa) Cross-section area: 10 mm^2	1000 kg/m^3	0.4	67	Wong et al., 2016
Ground plate	Linear tetrahedral solid (C3D4)	Linearly elastic (Young's modulus: 17000 MPa)	1000 kg/m^3	0.3	12800	N/A

	Element	**Material property**	**Density**	**Poisson's ratio**	**Mesh count**	**Reference**
Plantar fascia	Slip ring connector	linearly elastic (stiffness: 182.4-232.5 N/mm)	1000 kg/m^3	N/A	N/A	Kitaoka et al., 1994

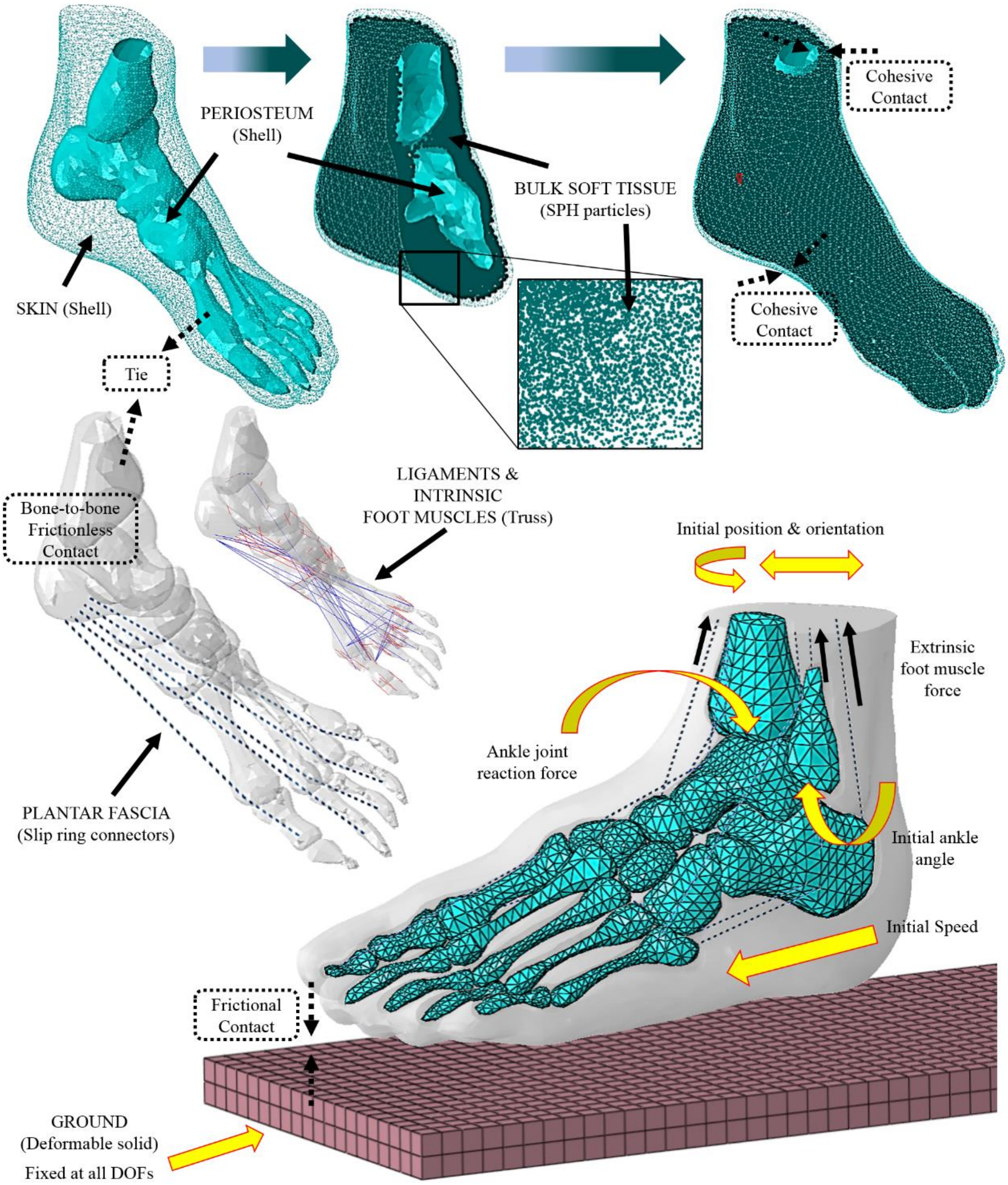


Figure 32. Overview of finite element foot model setup and boundary conditions.

The bulk soft tissue was modeled as SPH particles and encapsulated in a shell unit with an interior periosteum layer and exterior skin layer. The internal layer of the shell was tied to the skeletal structures. The plantar foot was connected by ligaments (truss unit), intrinsic foot muscles (truss unit), and the plantar fascia (slip ring connector). The ground plate was fully fixed and the foot model was placed at an initial position/orientation with an initial ankle joint *angulation. The three-dimensional ankle joint reaction force, extrinsic foot muscle force, and initial transitional velocity were applied to the model to drive the simulation.*

Boundary and loading conditions

The foot model was firstly positioned and orientated with a pre-set ankle joint angulation (Table 5), while the ground plate was fully fixed to sustain the impact of the footstep. The ankle angle (the angle of the tibial axial line and the foot longitudinal axis on the sagittal plane) was adjusted to the targeted value in a short analytical step at the beginning of the simulation. The global coordinate system was configurated based on the OpenSim definition (Delp et al., 2007) to ensure the reference frame was consistent. The initial striking velocity that corresponded to the instant before initial contact was assigned to the foot model. Extrinsic foot muscle force and three-dimensional ankle joint reaction force were applied on the slip-ring connectors and the tibiotalar articular surface of the talus respectively (Figure 32). All force data were input in a tabulated time-series matrix sourced from the musculoskeletal model. Gravity was enabled using a force-to-mass ratio of 9.8.

Due to the small geometry of the cartilages, they were substituted by a frictionless contact property assigned to the bone-to-bone interaction (Athanasiou et al., 1998). The coefficient of friction between the skin and the ground plate was 0.6 (Zhang and Mak, 1999). The surface of the periosteum was tied to the skeletal structure of the foot. The bulk soft tissue was attached to the shell unit using the default cohesive contact property (Figure 32).

Simulation solver and data output

The simulation was conducted in Abaqus (version 6.14, Dassault Systèmes, Waltham, USA) using the dynamic explicit solver. The foot-ground angle, ankle and first MTP joint angle, foot arch deformation, maximum principal stress on the major plantar connective tissues, and the plantar fascia tensile force were reported. The foot-ground angle was defined as the angle between the foot longitudinal axis and the ground surface (positive angle when the foot was upward-tilted). Foot arch deformation was represented by the arch height index (AHI) and medial longitudinal arch (MLA) angle. AHI was the ratio of medial foot arch height at 50% foot length to the truncated foot length (Miller et al., 2014). MLA angle was defined as the angle between the vectors pointing from the navicular tubercle to the first metatarsal head and posterior calcaneus (Prachgosin et al., 2015). Plantar fascia tensile force was the connector force acting on the five fascia bands.

5.4. Results

5.4.1. Spatiotemporal parameters

The running speed was controlled within 5% variance of the target value (Table 5). Foot strike index of each condition fell in the corresponding rearfoot (< 33.3%) and forefoot (> 66.6%) categories. The participant decreased 11.1% of stance phase duration and increased 4.76% cadence in the transition from rearfoot strike to forefoot strike. The peak vertical GRF was comparable between conditions, while forefoot strike reduced the averaged loading rate of vertical GRF by 45.46%.

5.4.2. Boundary conditions from musculoskeletal model

The two conditions showed apparent differences in muscle force during early- and mid-stance (Figure 33). Ankle plantarflexors were more activated (57.90–100.49% higher) in forefoot strike, while some ankle dorsiflexors outputs were reduced (29.10–51.73% lower). Peak ankle joint reaction force in forefoot strike was 39.68% (3.19 BW) larger in the anteroposterior direction and 22.98% (2.61 BW) larger in the axial direction (Figure 34). Forefoot strike also produced higher initial foot strike velocity, tibia internally rotation, tibia supination, and ankle plantarflexion during initial contact (Table 5).

Table 5. General gait parameters and initial segmental kinematics.

	Rearfoot strike	**Forefoot strike**
Measured running speed (m/s)	10.08	10.16
Duration of stance phase (s)	0.27	0.24
Cadence (steps/minute)	168	176
Foot strike index (%)	21.30	69.11
Peak vertical ground reaction force (BW)	2.30	2.44
Averaged loading rate (BW)	52.99	28.90

		Rearfoot strike	Forefoot strike
Initial foot velocity	Anteroposterior (m/s)	2.47 (forward)	2.53 (forward)
	Superoinferior (m/s)	0.54 (downward)	0.70 (downward)
	Mediolateral (m/s)	0.04 (inward)	0.06 (inward)
Initial tibial orientation	Sagittal (degrees)	8.13 (supination)	4.62 (supination)
	Frontal (degrees)	5.01 (inversion)	6.53 (inversion)
	Transversal (degrees)	4.21(external)	-0.05 (external)
Initial ankle plantarflexion (degrees)		-0.67	18.34

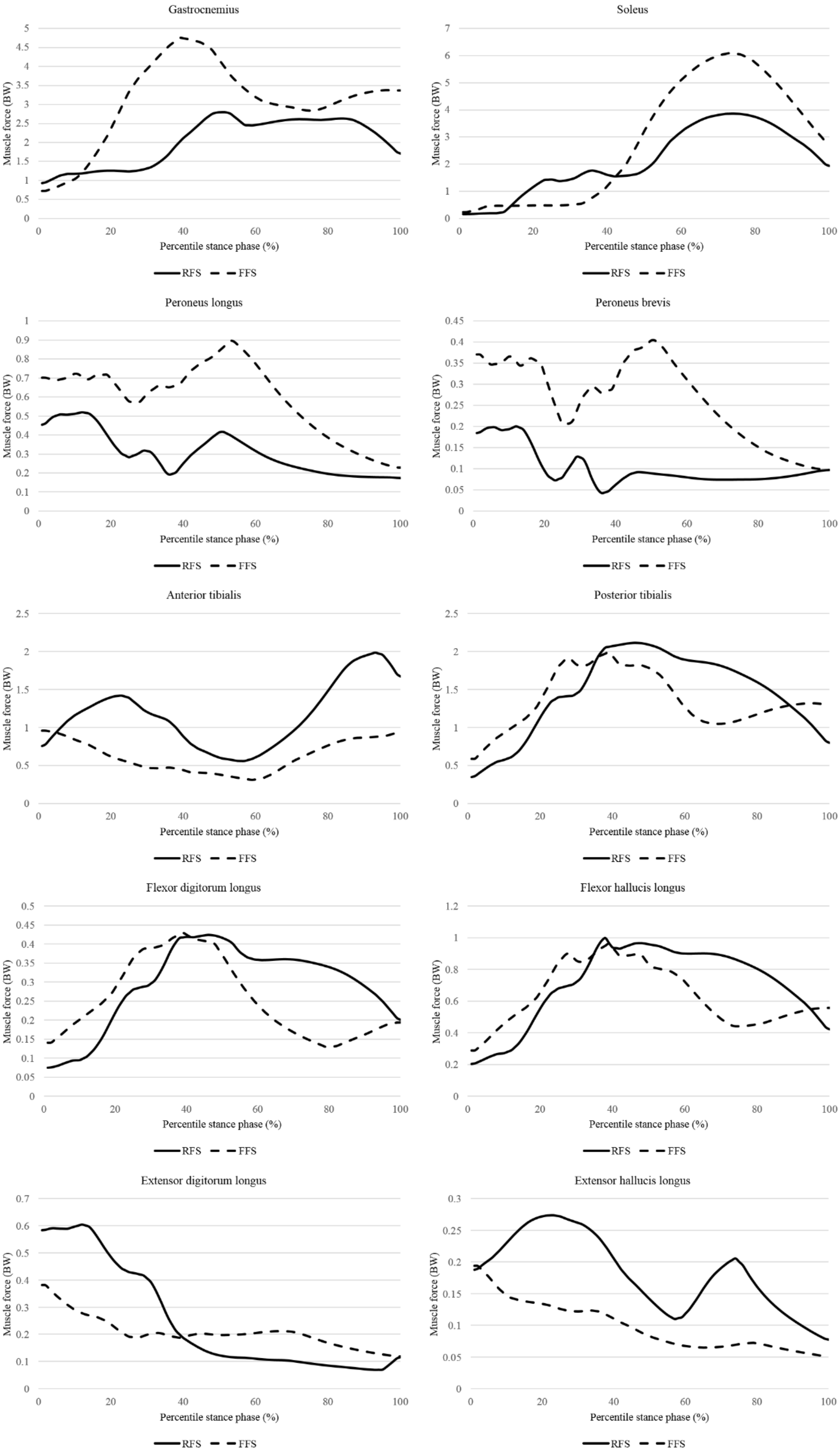

Gastrocnemius
Soleus
Peroneus longus
Peroneus brevis
Anterior tibialis
Posterior tibialis
Flexor digitorum longus
Flexor hallucis longus
Extensor digitorum longus
Extensor hallucis longus
Muscle force (BW)
Percentile stance phase (%)
RFS
FFS

Figure 33. Extrinsic foot muscle force estimated by the musculoskeletal model.

The force expressed in BW and scaled to percentile stance phase for both groups. RFS: rearfoot strike, FFS: forefoot strike.

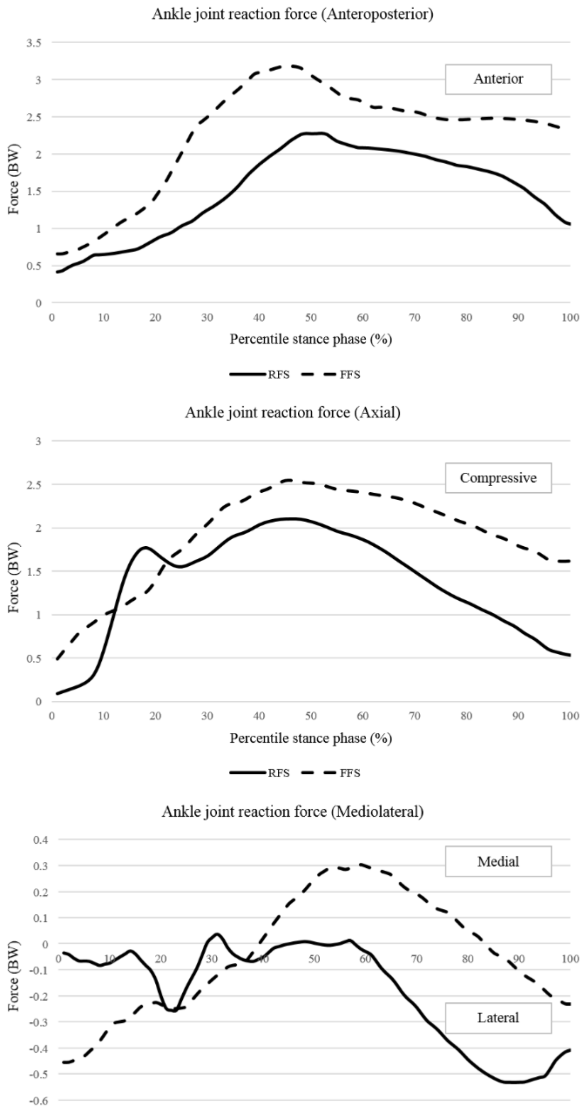


Figure 34. Ankle joint reaction force estimated by the musculoskeletal model.

The force is normalized to the bodyweight and scaled to percentile stance phase for both groups. Positive values mean the force applied in the anterior, inferior, and medial directions *with respects to the tibiotalar articular surface of the talus. RFS: rearfoot strike, FFS: forefoot strike.*

5.4.3. Segmental and joint kinematics from finite element simulation

Figure 35 demonstrated the simulation outcome of running dynamics in our study. As shown in Figure 36(A-C), The most prominent kinematic differences occurred at initial contact. In forefoot strike, the participant reduced the foot-ground angle (down-tilt the foot) by 22.61 degrees and increased ankle plantarflexion by 19.01 degrees. Meanwhile, the peak first MTP joint dorsiflexion in forefoot strike was 15.44 degrees higher compared to rearfoot strike.

Figure 36(E) and Figure 36(D) show that forefoot strike generated a notable drop in foot arch height shortly after the initial contact. The minimal AHI was 9.12% lower in forefoot strike (0.24) compared to rearfoot strike (0.27). Correspondingly, MLA angle increased rapidly in forefoot strike during the first half of the stance phase and reached its peak of 137.66 degrees at mid-stance, which was 2.06% higher than that of rearfoot strike (134.73 degrees).

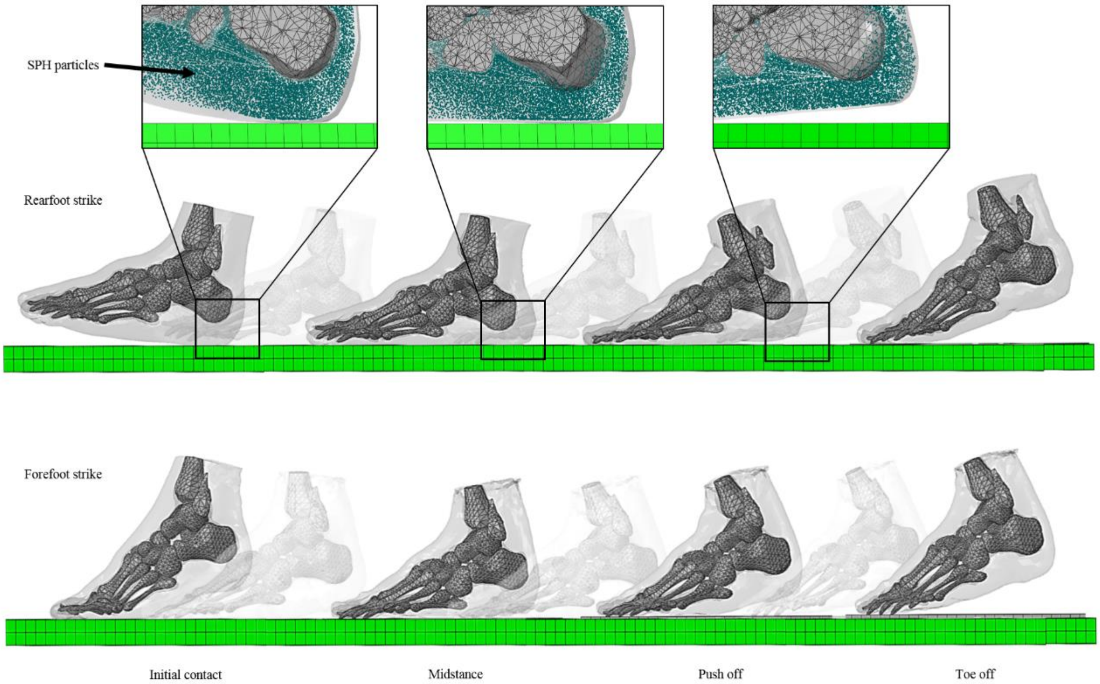


Figure 35. Dynamic simulation of rearfoot strike and forefoot strike running.

The movement of the finite element foot model was continuous from the initial contact to toe-off phase. The SPH particles representing the soft tissue were compressed and attached to the shell unit (the periosteum and skin) during impact.

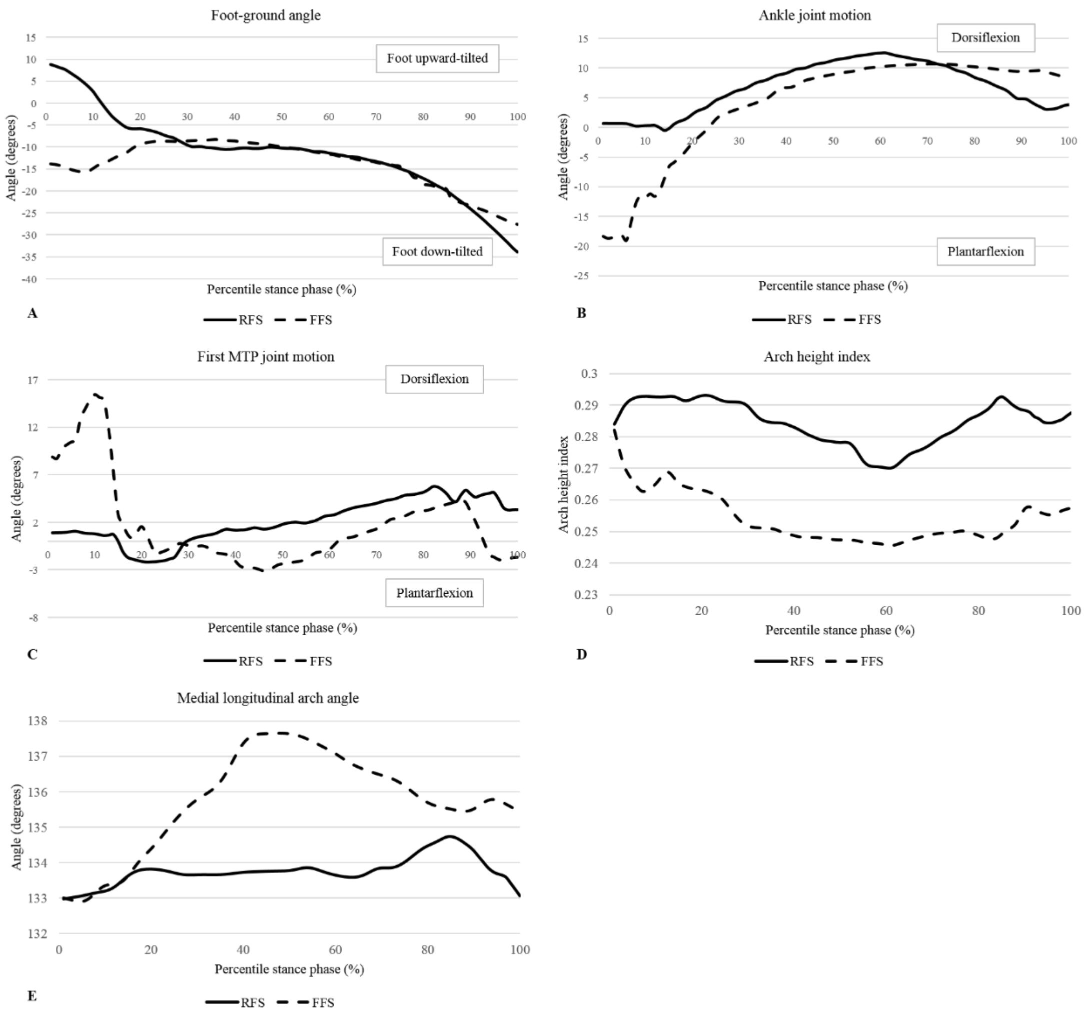


Figure 36. Segmental kinematics predicted by the finite element analysis: (A) foot-ground angle; (B) Ankle joint motion; (C) First MTP joint motion; (D) Arch height index; (E) MLA angle.

All variables are scaled to percentile stance phase for both groups. Foot-ground angle is the angle between the foot longitudinal axis and the ground surface. The ankle and MTP joints are dorsiflexed/plantarflexed on the sagittal plane. Positive values mean that the foot is upward-tilted with respects to the ground and the ankle/MTP joint is dorsiflexed. RFS: rearfoot strike, FFS: forefoot strike.

5.4.4. Plantar connective tissue loading from finite element simulation

All examined connective tissues were apparently more loaded in forefoot strike (Figure 37), except for the flexor hallucis brevis. The peak principal maximal stress was about one-fifth to two-fold higher in forefoot strike. The plantar ligaments shared a greater proportion of the increased foot arch loading in forefoot strike. Stress increments in plantar ligaments (0.93–9.67 MPa) were higher than that of the intrinsic foot muscles (0.35–0.37 MPa). Tensile force of the plantar fascia was generally higher in forefoot strike for the first to fourth bands (Figure 38), in which the maximal force could be 18.71–109.10% higher.

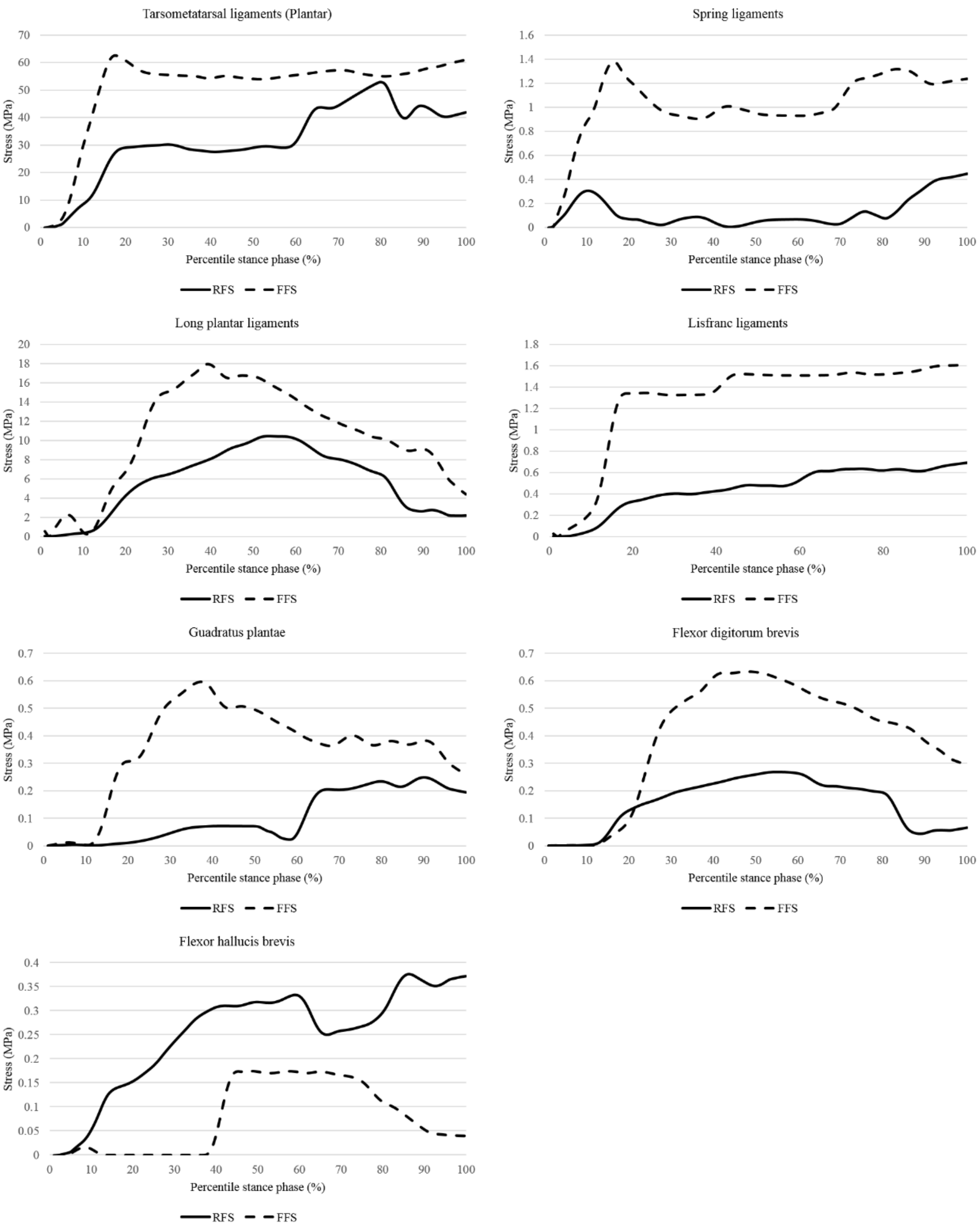


Figure 37. Principal maximal stress on the major plantar foot ligaments and intrinsic foot muscles.

All variables are scaled to percentile stance phase for both groups. RFS: rearfoot strike, FFS: forefoot strike.

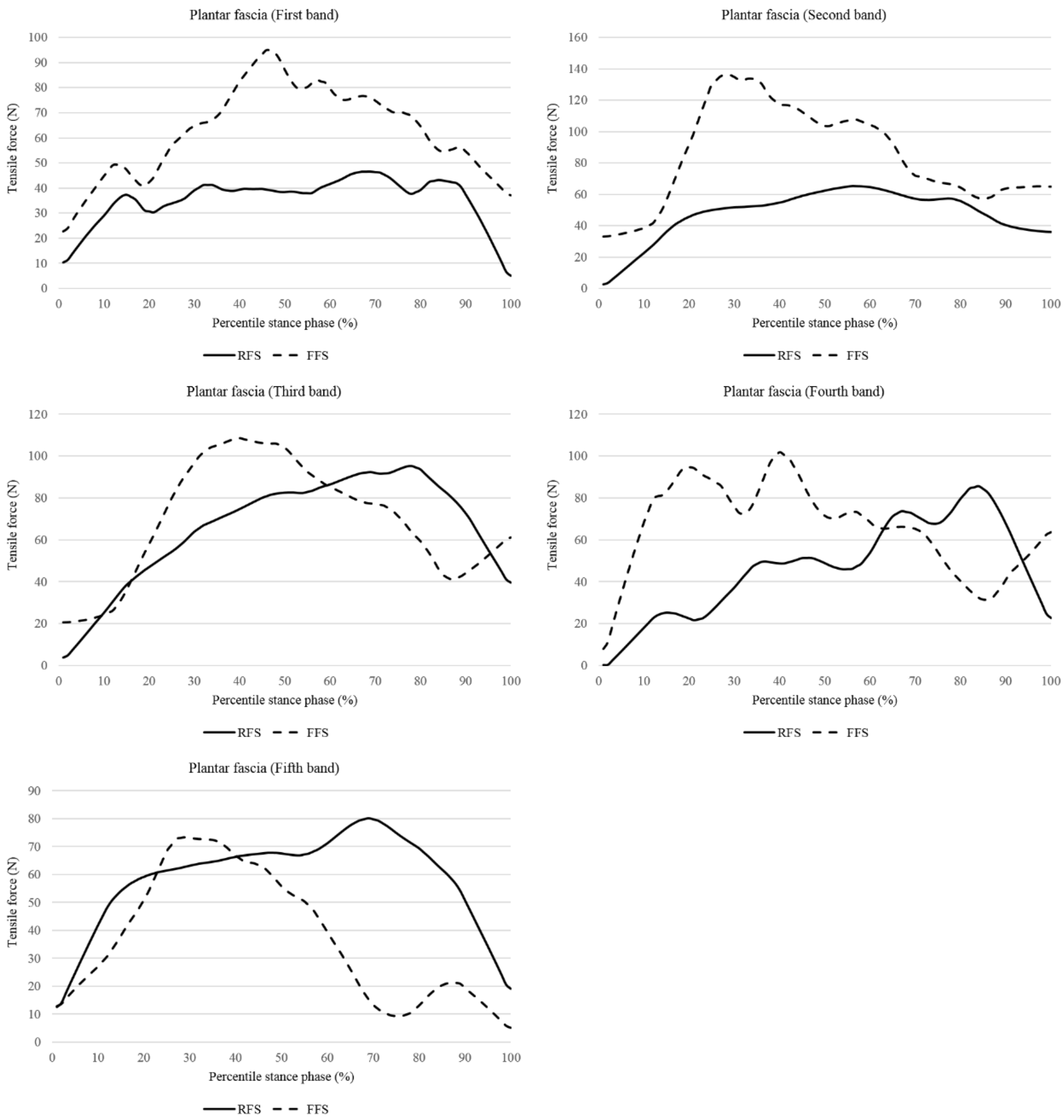


Figure 38. Tensile force on the five plantar fascia bands.

All variables are scaled to percentile stance phase for both groups. RFS: rearfoot strike, FFS: forefoot strike.

5.5. Discussion

The main purpose of this study was to explore the biomechanical behavior of the foot arch and plantar connective tissues in running with different foot strike techniques. The significance of this study resided in its potentials to reveal the risks of plantar fasciitis inherent in foot strike pattern modification. Our findings supported the hypothesis that forefoot strike increased the foot arch deformation, plantar connective tissue stress, and tensile force on the plantar fascia.

We adopted an SPH method in modeling the bulk soft tissue. SPH presented several advantages because it avoided extreme mesh distortions that were frequently encountered in highly impact problems (Johnson et al., 1996). The SPH implemented a cohesive contact property with the boundary finite element mesh and could accommodate high deformation and propagation speed. The interval distance between the SPH particles was 1 mm to 1.5 mm in our study, which was believed to be sufficiently fine (Jankowiak and Łodygowski, 2013). Previous studies reported that SPH provided accurate results in simulating impact (Kulper et al., 2018) and was now applied to running dynamics.

Our results were similar to that of previous literature. Existing studies reported that the alterations of foot and joint angle between rearfoot strike and forefoot strike mostly occurred at initial contact. Depending on research setups, the foot-ground angle at landing could range from 7.60 degrees to 14.85 degrees for rearfoot strike and from -12.46 degrees to -3.37 degrees for forefoot strike (Nunns et al., 2013; Shih et al., 2013; Williams et al., 2000, 2012). Ankle angle at landing ranged from 10.08 degrees to 24.80 degrees for rearfoot strike and from -12.46 degrees to 2.3 degrees for forefoot strike (Kulmala et al., 2013; Shih et al., 2013; Williams et al., 2012). MTP joint motion was rarely observed, but it was reasonable for the runner to dorsiflex his toe in forefoot strike as a mean to increase the contact area of the forefoot. Landing with down-tilt foot/plantarflexed ankle in forefoot strike facilitated a larger ankle joint excursion and negative work done by the calf muscles to resist heel drop and to absorb impact.

Previous studies measuring EMG signals also found that forefoot strike increased ankle plantarflexor (5.10–23.53% higher in Achilles tendon tension) and decreased ankle dorsiflexor activities (54.48–75.00% lower in anterior tibialis) (Kulmala et al., 2013; Landreneau et al., 2014; Rice and Patel, 2017; Yong et al., 2014). The increased ankle joint motion and calf muscle force in forefoot strike were considered as the primary contributor to the reduced loading rate of the vertical GRF (Kulmala et al., 2013; Shih et al., 2013), and the trend of increased ankle joint loading (Rooney and Derrick, 2013). All these factors could influence the bending force on the foot arch and cause excessive arch deformation.

Excessive arch deformation was associated with plantar tissues overload and the pathologies of ligamentous injuries (Tao et al., 2010; Thordarson et al., 1995). The changes of AHI reported by our simulation was in accordance with previous studies (0.028 in running) (Hageman, 2010). MLA angular changes were also found similar to the value measured by fluoroscopy (2.3 degrees to 6.1 degrees) (Fukano and Fukubayashi, 2012). Under the non-weight bearing condition, the increased Achilles tension and MTP dorsiflexion should have elevated the foot arch due to the tightening of the plantar connective tissues (Bolgla and Malone, 2004). Instead, the plantar connective tissues underwent elongation as a result of compression from the arch top in forefoot strike running (Morales-Orcajo et al., 2018).

Plantar connective tissues are important components of the multi-layer load-bearing system of the foot arch (Kirby, 2017). The present study demonstrated the synergy among foot muscles, plantar ligaments and plantar fascia (Crary et al., 2003). Research reported that the range of plantar fascia tension was 464–922 N during walking (Chen et al., 2014; Erdemir et al., 2004; Lin et al., 2014), which was relatively high compared to our study (372 N in rearfoot strike). One possible explanation could be that the disregard of some muscle forces could transfer the total foot arch load to other arch stabilizers (Kirby, 2017). Our predicted increases of plantar fascia force in forefoot strike (18.7–109.10%) were also larger than that (9.57%) of McDonald’s (McDonald et al., 2016) because the influence of foot arch deformation was considered in the present study. In accordance with existing findings (Futrell et al., 2019; Goss and Gross, 2013; Kulmala et al., 2013; Laughton et al., 2003; Shih et al., 2013), reducing loading rate of the vertical GRF at touchdown (approximately 45% lower in forefoot strike in this study) is one of the key features that distinguishes forefoot strike from rearfoot strike. However, this changed loading rate on foot may not be the major cause of the increased fascial tension in forefoot strike running. In regardless of the landing styles, GRF does not act directly on the plantar fascia throughout the stance phase. GRF may influence the fascial loading by affecting movements of the bony structures which the fascia band was attached to, but the influence was likely small as it could be easily compounded by activities of other plantar soft tissues (Kirby, 2017). Besides, stiffness of fascia bands was reportedly not sensitive to a wide range of tensile loading rates. The different tensile force on the fascia band between rearfoot strike and forefoot strike may be more related to foot arch loading/deformation rather than GRF. Our findings were supported by another finite element study (Li et al., 2017), which reported a higher stress level and stress increase rate on the metatarsal bones in forefoot strike than rearfoot strike. A larger external force applied to the plantar forefoot in forefoot strike could further stretch the plantar fascia due to the bending strain to the foot arch (Kernozek et al., 2014, 2016).

The fatigue life of the fascia tissue was mainly determined by the maximal stress range that it sustained in the loading cycles (Carter et al., 1981). Therefore, the increased peak fascia tension in forefoot strike was likely to reduce the sustainable gait cycles before localized damage occurred. Additionally, habitual rearfoot strike runners usually increased their cadence when running in forefoot strike at the same speed (Baggaley et al., 2017). The increased loading in each cycle, together with the higher step frequency, would expose the runner to a faster pathological process if the same running regime of rearfoot strike is carried over to forefoot strike. This could be the reason that mainstream gait retraining programs emphasize the importance of a step-by-step procedure for runners to adopt forefoot strike (Huffer et al., 2017). They suggested to reduce running volume and strengthen foot muscle at the early training stage, to avoid plantar fascia overload (Cheung et al., 2015). As a mean to minimize the incidence of plantar fasciitis, runners should adjust their gait patterns with cares and also follow professional guidance.

Besides the simplification and assumption made in the modeling procedure, the single-subject design was the major limitation of our study. The problem was commonly faced by research using a theoretical approach, e.g., finite element method. The material of the ligaments and muscles were assumed linearly elastic despite that they exhibit hyperelastic or viscoelastic behavior. The approach may underestimate the joint stiffness of the model, while it remains a common simplification strategy in finite element foot model to compromise computational efficiency, in addition to the fact that the material property profile of some foot ligaments is incomplete (Morales-Orcajo et al., 2016). As foot strike patterns influenced muscle activities, measuring EMG signals of the leg muscles could enrich the inputs to the simulation and increase the accuracy of force prediction. The healthy participant recruited was assumed to represent the typical characteristics of the runner population. However, individual variances in foot geometries, gait characteristics, and running regime were not considered. A prospective study tracking the injury progression for both rearfoot strike and forefoot strike runners would provide more insight into the foot problem.

CHAPTER 6 PREDICTION ON THE PLANTAR FASCIA STRAIN OFFLOAD UPON FASCIA TAPING AND LOW-DYE TAPING DURING RUNNING

6.1. Summary of the study

Taping is commonly prescribed to treat plantar fasciitis for runners by virtue of its alleged ability to offload the plantar fascia. Reduction of fascial strain is a potential indicator of pain relief and injury prognosis. Clinical guideline for treating plantar fasciitis has consistently considered taping as the first-line intervention, in spite of that supportive evidence is limited from the mechanical standpoint.

This study aimed to investigate how different taping methods could change the loading on the plantar fascia during running using computational simulations. A finite element foot model was modified from a previous version to fit the study's purpose. Besides the bones, bulk soft tissue, foot muscles, ligaments, and tendons that were already included in assembly, the foot model was further incorporated with a solid part representing the plantar fascia. A mesh convergence test and model validation analysis were also performed to the updated foot model. A runner performed several running trials under one untaped condition and two taped conditions—Low-Dye taping and Fascia taping, which were implemented by a physiotherapist using the Kinesio tapes. The captured motion data were processed to drive a scaled musculoskeletal model and calculate segmental kinematics, foot muscle force, and joint reaction force. These variables were then input as the boundary/loading conditions for finite element analyses of running. The principal tensile strain on the plantar fascia, subtalar eversion, and navicular height during the stance phase were averaged across five trials of each condition and compared using Friedman's test.

A mesh size of 4.0 mm for the plantar fascia produced a satisfied outcome solution (< 5% changes in the fascial strain) under the loading conditions of the mid-stance of running gait. Validation of the modified foot model was accomplished as the model-predicted and the experiment-measured fascial strains were comparable under the same loading setup. Simulations of running in the three untaped/taped conditions showed that, maximal subtalar eversion did not differ among conditions (p = 0.449). Fascia taping significantly reduced maximal strains on the fascia band (p = 0.034, Kendall's W = 0.64–0.76) and increased the navicular height (p = 0.013, Kendall's W = 0.84) compared to nontaping. There were no significant differences in all outcome variables between Low-Dye taping and nontaping (p = 0.173–0.618).

In the view of controlling tissue overstrain, we believe that Fascia taping was more effective than Low-Dye taping in treating plantar fasciitis. Low-Dye taping may be better indicated for plantar fasciitis secondary to rearfoot malalignment. However, further studies would be needed to address the point.

6.2. Introduction

Taping is commonly used to treat plantar fasciitis particularly for palliative management of pain at an early pathological stage (Landorf et al., 2005). Tapes attached to the foot were opined to correct faulty biomechanics and take the strain off the plantar fascia (Podolsky and Kalichman, 2015), hereby to improve the symptoms of plantar fasciitis. Both Low-Dye taping and Fascia taping have received much research attention regarding their treatment effects on plantar fasciitis (Radford et al., 2006a; Salvioli et al., 2017; van de Water and Speksnijder, 2010). Though their clinical applications are prevalent, the reported outcomes were oftentimes conflicting. One major reason for the inconsistency is that direct comparisons among taping methods are few in terms of their mechanical functionality. It is still unclear how types of taping could change the foot biomechanics differently. In fact, current evaluations of the treatment effects are predominately based on subjective feedback from the patients (McPoil et al., 2008). The causes of symptom relief could be multi-factorial and do not necessarily reflect the changes in plantar fascia loading. Moreover, difficulties in in-vivo assessment of soft tissue strain impede research in this field. Quantitative evidence is needed to determine the extent of the offloading of the plantar fascia among different taping treatments.

In the study, we aimed to investigate loading on the plantar fascia between the taped conditions (Low-Dye taping and Fascia taping) and untaped conditions (control) during running by means of a computational foot model. Besides, the ability of taping to constrain subtalar motion and support the foot arch was also evaluated. The primary outcomes were the principal tensile strain on the plantar fascia, subtalar eversion, and navicular height during the stance phase. The hypotheses were that, during the stance phase of running, 1) the plantar fascia would be less strained under the taped conditions; 2) the maximal subtalar eversion would be reduced and; 3) the minimal navicular height would be increased by taping.

6.3. Methods

6.3.1. Subject information

A recreational runner (male, aged 30 yr., 170 cm tall, and 68 kg in mass, the same runner participating in study two) was recruited for the computer model reconstruction and gait experiment. The participant had a running experience of twelve years, average weekly mileage of 15 km, and usual pacing of 6 minutes per kilometer. He reported no musculoskeletal diseases/injuries or medical history that could influence his running performance. He was fully informed of the research procedure and signed the consent form. The experimental protocol was approved by the institution authority (IRB NO. HSEARS20170626003).

6.3.2. Experimental procedure

The experiment comprised several running trials under one untaped condition and two taped conditions (Low-Dye taping and Fascia taping). A physiotherapist implemented taping to the runner using the Kinesio tapes (Kinesio Tex Classic, Kinesio, NM, USA). The methods for Low-Dye taping and Fascia taping are illustrated in Figure 39 and Figure 40 (Hyland et al., 2006; Tsai et al., 2010).

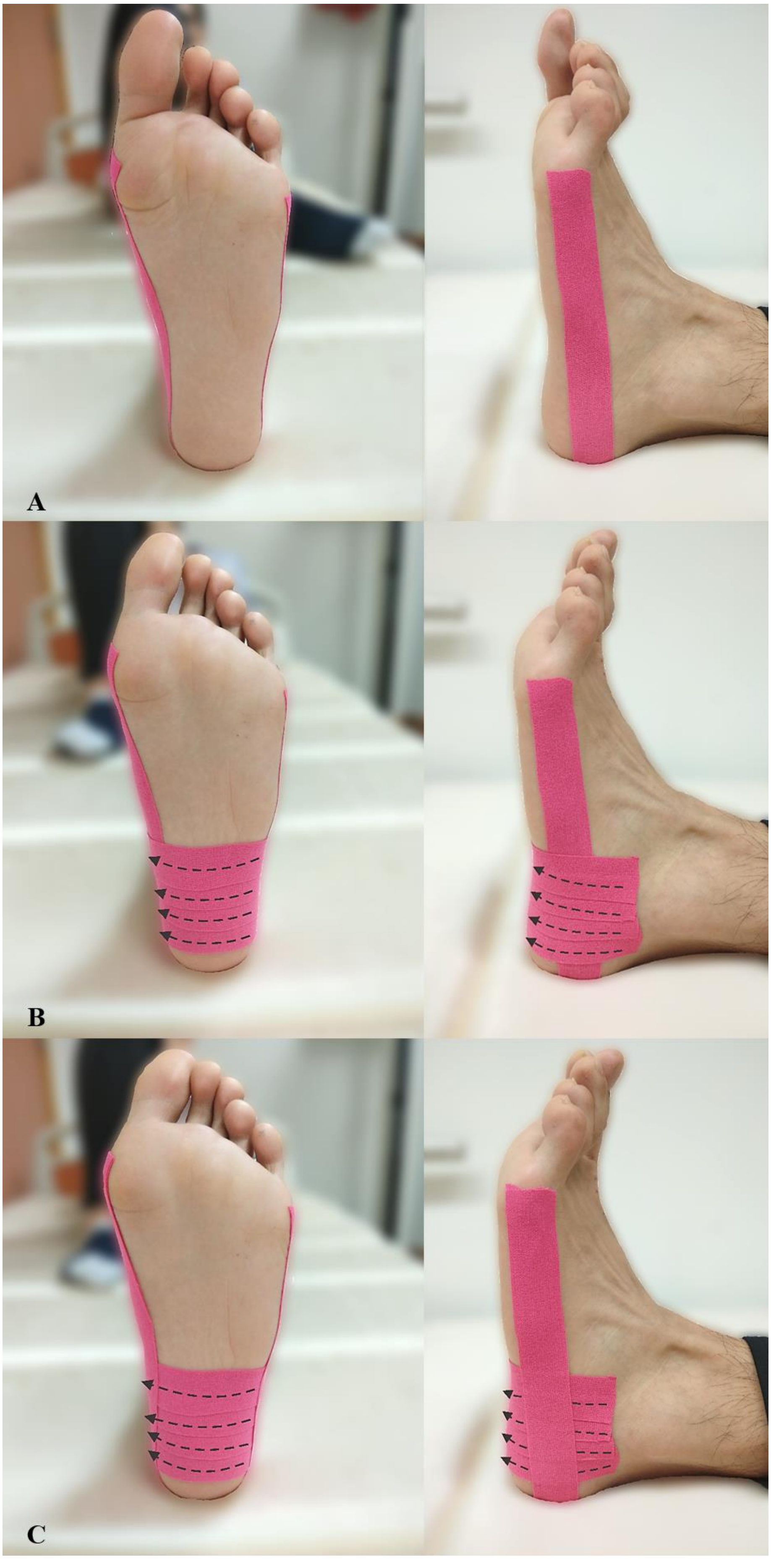

A
B
C

Figure 39. Procedures for Low-Dye taping. Dashed arrows denote the direction of stretch.

(A) The first strap, which served as the anchor tape, was applied with gentle compression. It started at the medial first metatarsal head, stretched proximally along the medial border of the foot, went around the back of the heel, and attached to the lateral fifth metatarsal head; (B) A series of stirrups were applied when the rearfoot was slightly supinated. The first stirrup started distal to the lateral malleolus, pulled the calcaneus medially with 50% stretch (50% tensile strain), and attached to just below the medial malleolus. The second, third, and forth stirrup followed the same pattern with overlap of approximately one-third of the tape width, moving in the distal direction; (C) The last strap was applied in the similar form of the first strap to secure the whole taping structure.

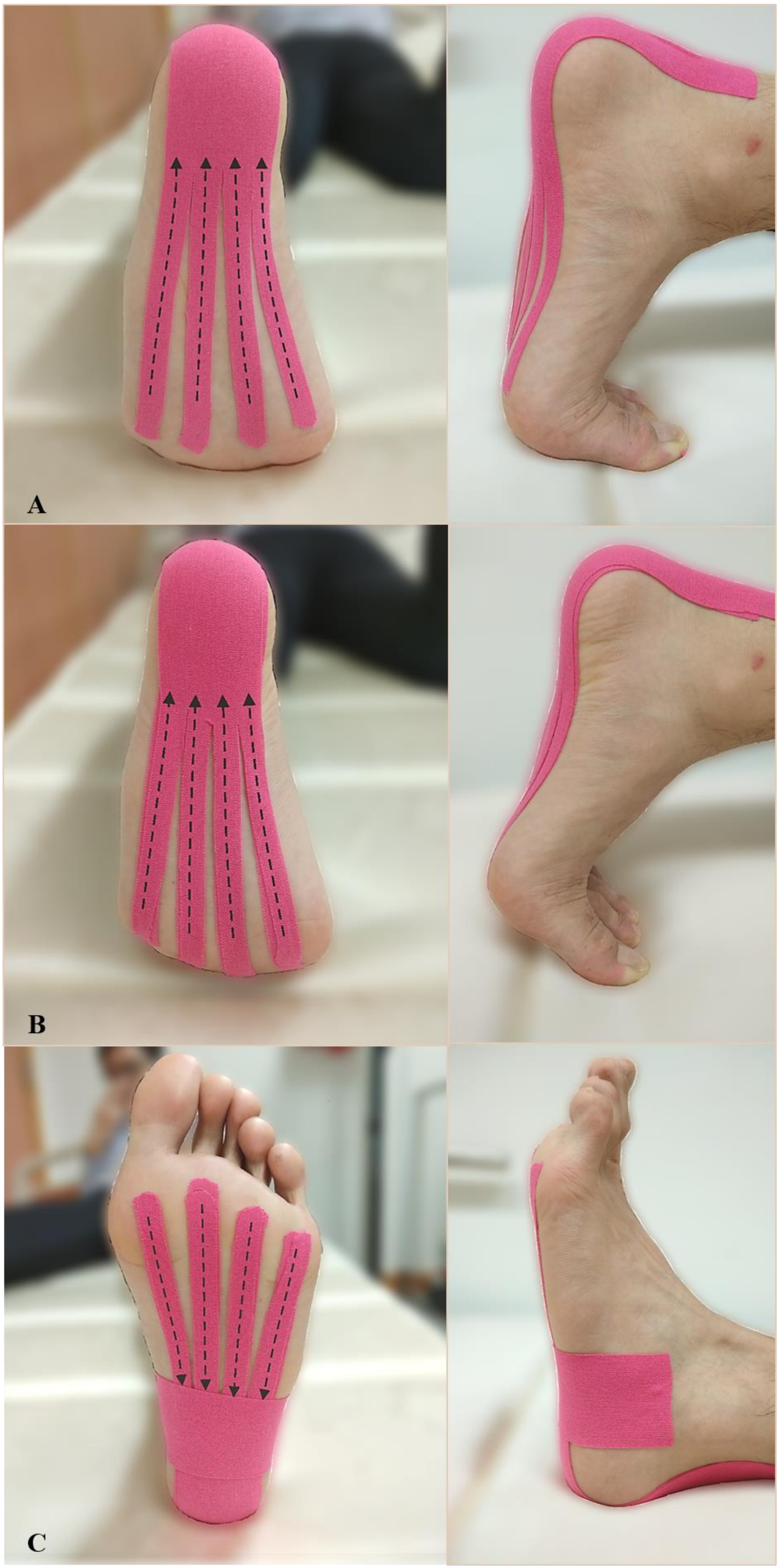

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Figure 40. Procedures for Fascia taping. Dashed arrows denote the direction of stretch.

(A) After the metatarsophalangeal joints were dorsiflexed, the first strap was adhered firmly to the posterior heel at its proximal end. The other end of the strap was cut into four slices with equal width. Each slice was applied with 50% stretch (50% tensile strain) and attached to the plantar forefoot; (B) Another strap was applied following the same pattern and overlapped the first strap; (C) The last strap was applied with gentle compression across the bases of the four slices beneath the foot and wrapped around the rearfoot.

Prior to the experiment, the participant was given ample time to warm up. The three conditions (untaped/taped) were tested in a randomized sequence with a five-minute rest interval. The participant ran barefoot to eliminate the influence of footwear. For each condition, the participant was instructed to run through the motion capture volume at his usual pacing (10 km/h). Two pairs of photoelectric cells were placed 2.6 m apart along the runway to monitor running speed (Hamill et al., 2014). Running trials were repeated until five sets of successive data were obtained for each condition. Data were considered successful when the running speed fell with 5% variance of the target value (10 km/h) and the footstep landed completely within the force platforms.

A motion capture system with eight optical-based cameras (Vicon, Oxford Metrics Ltd., Oxford, UK) and four force platforms (OR6, AMTI, Watertown, USA) were used to record marker trajectories and ground reaction force. The data were sampled at 250 Hz and 1000 Hz respectively. Thirty-six retroreflective markers were affixed to the following anatomic landmarks for motion capture: acromioclavicular joints, posterior/anterior iliac spines, greater trochanters, lateral/medial femoral epicondyles, lateral/medial malleoli, calcaneal tuberosity, the base/head of the first and fifth metatarsals, and the distal phalanx of the hallux.

6.3.3. Musculoskeletal model

The kinematic and kinetic data collected from the running trials (five trials for each condition) were input to the musculoskeletal model simulation (OpenSim, version 3.3, National Center for Simulation in Rehabilitation Research, Stanford, USA). Firstly, a generic model (John et al., 2013) with 12 rigid-body segments, 23 degrees of freedom, and 92 musculotendinous units was scaled to accommodate the mass and anthropometry of the participant. Inverse kinematics was then solved and the dynamic inconsistency was reduced by small adjustments to model mass properties (Arnold et al., 2010). Forces of the extrinsic foot muscles were estimated by the built-in Computed Muscle Control modulus (Thelen, 2003). Joint reaction force and segmental kinematics were generated by the Analyse toolkit. The above variables were exported as the boundary and loading conditions for the subsequent finite element analysis.

6.3.4. Finite element model

Geometry reconstruction

The MRI of the participant was processed to construct the geometry of the left foot model. The foot model was also used in study two of the thesis (Chen et al., 2019b). Briefly, the images were acquired by a 3.0T MRI scanner (GoldSeal Certified Signa HDxt, General Electric Company, Boston, USA) at T1 sequence, 1-mm slice interval and 0.625 mm pixel size when the participant's leg fixed at neutral position using a customized ankle-foot orthosis. Model reconstruction was performed with segmentation software (Mimics and 3-matics, Materialise, Leuven, Belgium).

The model included 20 bony segments, 26 ligaments, 9 intrinsic foot muscles, and 11 extrinsic foot muscles, as shown in Figure 41. The bulk soft tissue was modeled by a cluster of SPH particles encapsulated in a shell unit using cohesive property. The shell unit consisted of interior and exterior surfaces representing the periosteum layer (tied to the bony structures) and skin layer (in contact with the ground) respectively (Figure 41). In addition, the geometry of the plantar fascia was reconstructed as a three-dimensional solid based on the bony structures from the MRIs and the human anatomy atlas (Gray, 2011) using computer-aided design software (Solidworks v2014, Dassault Systèmes, MA, USA). From the origin to the distal end, the thickness of the fascia bands gradually reduced from 3 mm to 1.3 mm. The plantar fascia was tied to the inferior calcaneus proximally and to the five phalanges distally. The Kinesio tapes were modeled as shell units with 0.5 mm thickness based on the product catalog. Different strap designs were reproduced and tied to the skin surface in the model to mimic the two taped conditions (Figure 42). Pre-strain (+ 50% elongation) was applied on the tape straps as shown in Figure 42.

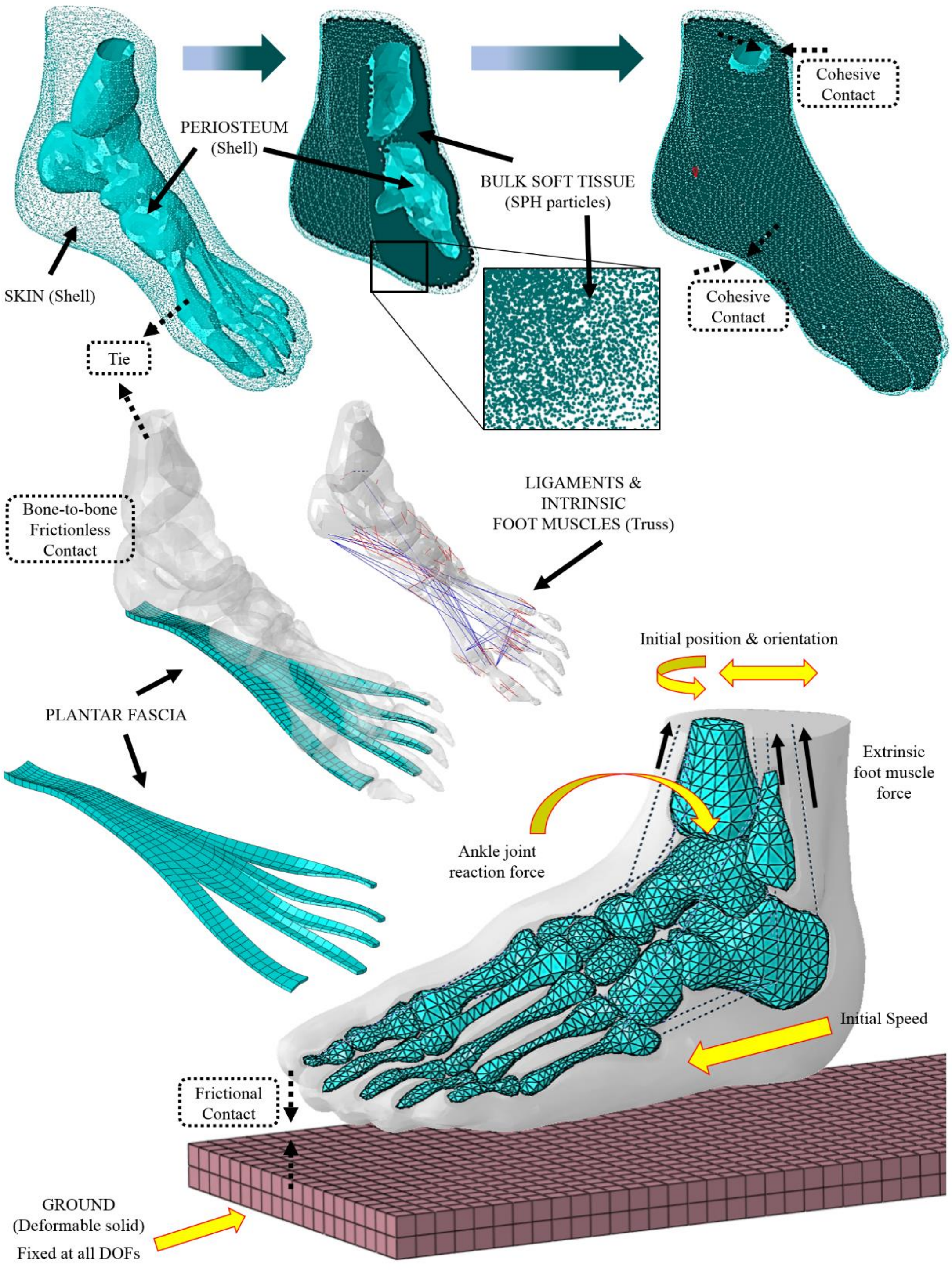

Figure 41. Setup overview of the finite element foot model and boundary conditions.

Solid black arrows denote the names of the parts. Solid maize arrows represent boundary/loading conditions. Dashed frames denote the interactions among parts. The bulk soft tissue was modeled as SPH particle and encapsulated in a shell unit that possessed an *interior periosteum layer and exterior skin layer. The internal layer of the shell was tied to the skeletal structures. The plantar foot was connected by ligaments (truss unit), intrinsic foot muscles (truss unit), and the plantar fascia (three-dimensional solid). The ground plate was fully fixed and the foot model was placed at an initial position/orientation. Three-dimensional ankle joint reaction force, extrinsic foot muscle force, and initial transitional velocity were applied to the model to drive the simulation. DOFs: degrees of freedom.*

Material Properties and Mesh Creation

The material properties of different model components were similar to those used previously (Table 5) except that the plantar fascia and Kinesio tapes were assumed hyperelastic and isotropic in this study. The updated information was listed in Table 6. The model was meshed by the finite element software, Abaqus v6.14 (Simulia, Dassault Systèmes, RI, USA). The overall mesh size was determined as 3.5 mm for the osseous and soft tissue components as supported by study two (Chen et al., 2019b). Another mesh convergence test was conducted targeting on the mesh size of the plantar fascia by simulating the mid-stance phases of a representative barefoot running trial conducted in the current study. The mesh size was reduced every 10% from 5.5 mm. The predicted strain deviation of the plantar fascia was less than 5% when the mesh size was reduced to 4.0 mm.

A frictionless contact algorithm was used for bone-to-bone interface to mimic the function of cartilages (Athanasiou et al., 1998). The interaction between the skin and the ground plate was “hard” contact with a friction coefficient of 0.6 (Zhang and Mak, 1999).

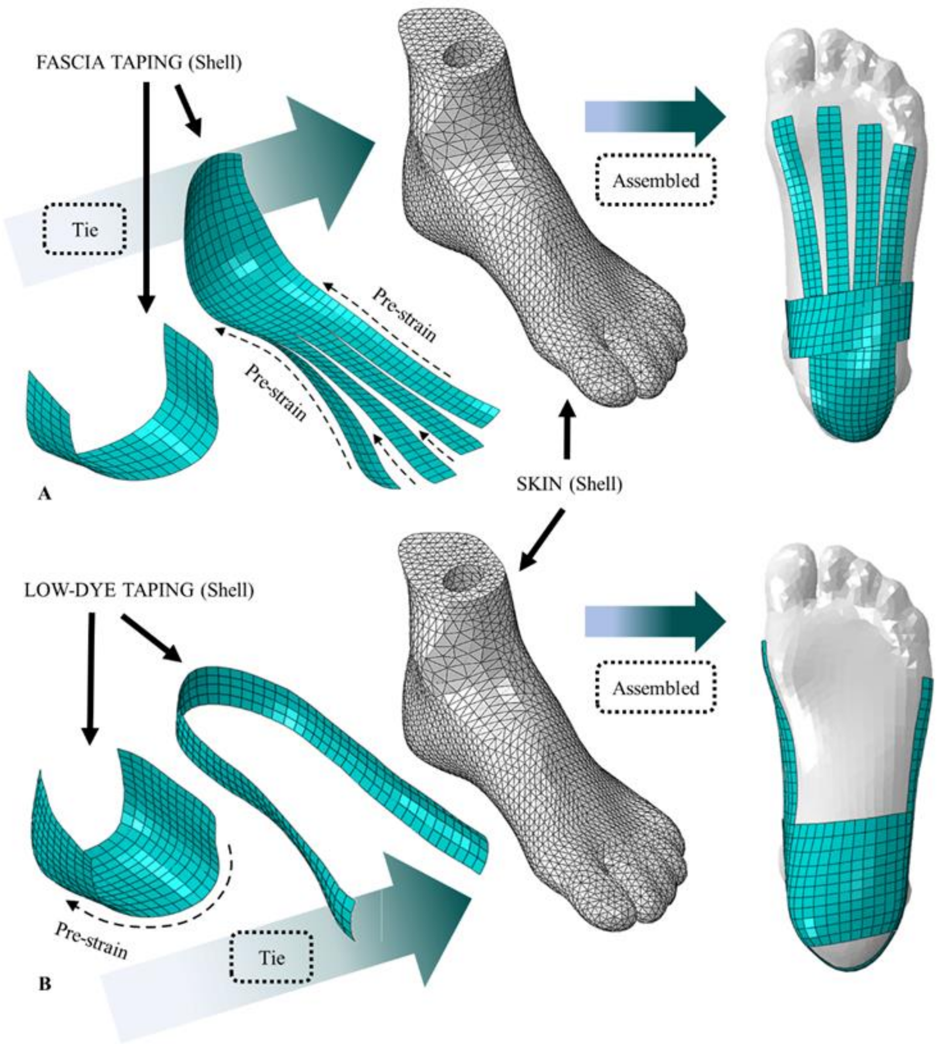

Figure 42. Modeling of (A) Fascia taping and (B) Low-Dye taping in the simulations.

Dashed arrows denote the direction assigned with a pre-strain of 50% to mimic stretch on the tapes. Solid black arrows denote the names of the parts. Shell units that represented the tape straps were tied to the skin surface.

Table 6. Element type, material property, and mesh count for the plantar fascia and Kinesio tapes.

	Element	**Material property**	**Density**	**Poisson's ratio**	**Mesh count**	**Reference**
Plantar fascia	Linear hexahedron solid (C3D8R)	Hyper-elastic (Second-order Polynomial model, C_{10}: −222.1, C_{01}: 290.97, C_{20} : −1.1257, C_{11}:4.7267, C_{02}:79.602)	1000 kg/m^3	0.4	438	Cheng et al., 2008; Kitaoka et al., 1994
Kinesio tape	Linear quadrilateral shell (S4R)	Hyper-elastic (First-order Ogden model, μ = 0.206 MPa, α = 8.88) Thickness: 0.5 mm	1150 kg/m^3	0.4	464-492	Boonkerd and Limroongreungrat, 2016

Boundary and loading conditions

The finite element model was driven by the boundary and loading conditions generated by the musculoskeletal modeling. Concentric connector force was applied to the slip-ring connectors to simulate extrinsic foot muscle force. Three-dimensional ankle joint reaction force was loaded on the ankle surface of the talus (Figure 41). All force data were input in a tabulated time-series matrix to drive the finite element analysis. The whole foot model was initially placed at the position and orientation corresponding to the instant before foot strike. A pre-defined velocity was also conferred to the whole foot model (Data of boundary and loading conditions are included in supporting files). Gravity was enabled throughout the simulation steps using the force/mass ratio of 9.8.

Simulation solver and data output

For each experimental condition, simulations of five running trials were performed by Abaqus dynamic explicit solver (version 6.14, Simulia, Dassault Systèmes, RI, USA). The principal tensile strain on the plantar fascia, subtalar eversion, and navicular height were reported and compared among the conditions. After excluding the regions that were tied to the bones, the plantar fascia was equally divided into three portions from proximal to distal. Principal tensile strain was averaged across all elements of each portion and the mean values were used for the statistical analysis. Strain contours of the plantar fascia of representative running trials were also plotted. Subtalar eversion was calculated as the angle between the longitudinal axes of the tibia and the calcaneus on the frontal plane. Longitudinal axes of the foot segments were defined by pre-labeled anatomical landmarks on the foot model based on an established method (Wu et al., 2002). Navicular height was measured as the distance between the navicular tuberosity and the plantar foot surface (Jamali et al., 2004).

6.3.5. Statistical analysis

Statistics were conducted in SPSS (Version 19.0, IBM, Armonk, USA) at a significance level of 0.05. Maximal strains of the proximal, middle, and distal plantar fascia, maximal subtalar eversion, and minimal navicular height during the running stance were compared among the three conditions. A non-parametric Friedman's test for related samples was employed to examine the significant differences. Dunn's pairwise comparison test with Bonferroni correction was carried out for pos hoc analysis (Kot et al., 2012). Effect size (Kendall's W) was calculated to demonstrate the strength of the results. The coefficients were interpreted as trivial effect (< 0.2), small effect (< 0.5), medium effect (< 0.8), and large effect (≥ 0.8) (Cohen, 1988).

6.4. Results

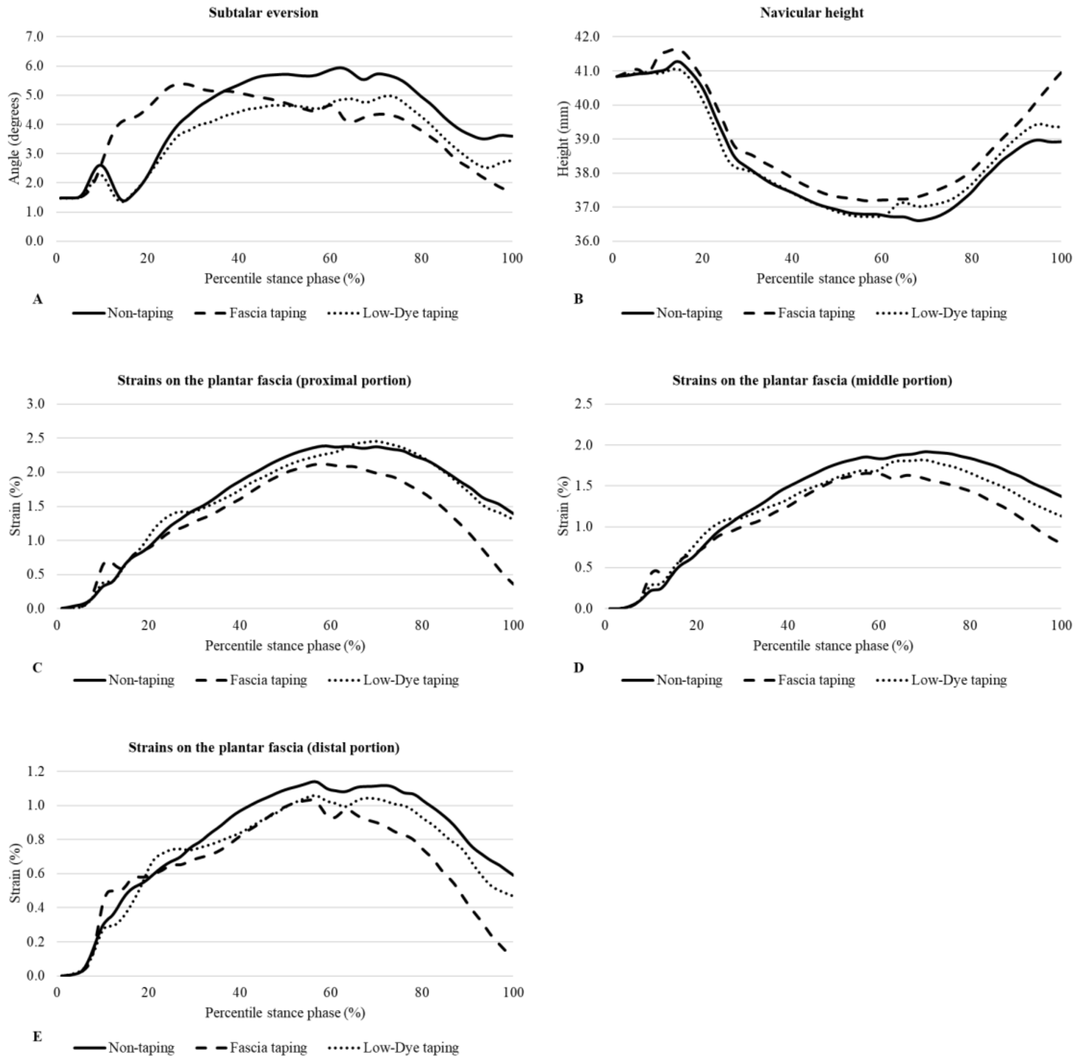

Figure 43***. Results of finite element analyses of running: (A) Subtalar eversion; (B) Navicular height; (C) Strains on the proximal plantar fascia; (D) Strains on the middle plantar fascia; (***E) Strains on the distal plantar fascia. ***All outcome variables are scaled to percentile stance phase.***

Figure 43 plot the outcome variables as a function of percentile stance phase. Friedman's test reported significant differences in the maximal strains of the plantar fascia (p = 0.022 – 0.041

for all three portions) and minimal navicular height (p = 0.015). Maximal subtalar eversion was similar among the three conditions (p = 0.449) (Table 7). Pos hoc analysis showed that Fascia taping significantly increased the navicular height (p = 0.013, Kendall's W = 0.84), reduced strains on the proximal (p = 0.034, Kendall's W = 0.76), middle (p = 0.034, Kendall's W = 0.64), and distal portions of the plantar fascia (p = 0.034, Kendall's W = 0.64) compared to non-taping (Figure 44). There were no significant differences between Low-Dye taping and non-taping (p = 0.173–0.618).

Table 7. Results of the finite element analyses (mean ± standard deviation).

		Non-taping	**Fascia taping**	**Low-Dye taping**	***p*-value**
Maximal subtalar eversion (degrees)		6.33 ± 2.32	5.43 ± 0.67	5.07 ± 0.47	0.449
Minimal navicular height (mm)		36.53 ± 0.28*	37.17 ± 0.15*	36.71 ± 0.21	**0.015**
Maximal strains on the plantar fascia (%)	Proximal	2.46 ± 0.10*	2.17 ± 0.08*	2.46 ± 0.08	**0.022**
	Middle	1.95 ± 0.10*	1.70 ± 0.08*	1.83 ± 0.10	**0.041**
	Distal	1.20 ± 0.07*	1.04 ± 0.03*	1.09 ± 0.08	**0.041**

* Significant difference in the pos hoc pairwise comparison.

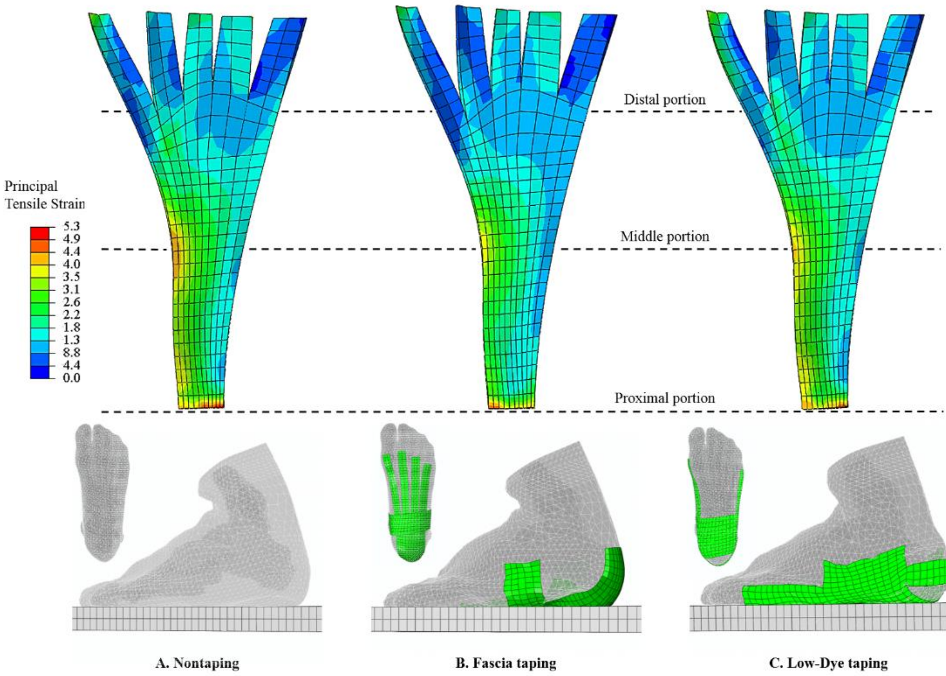

Figure 44. Strain contours of the plantar fascia at the instant of maximal strain for a representative trial of each running condition: (A) Non-taping; (B) Fascia taping; (C) Low-Dye taping.

The images are color-coded based on the distribution of principal tensile strains. Red means the highest strain and blue means the lowest. Regions of the plantar fascia that are tied to the bony segments are removed from display and the remainder is equally divided into three portions from proximal to distal. The figure shows that Fascia taping produced an apparently larger cool-tonal area, which indicates a lower strain level on average.

6.5. Discussion

The purpose of the study was to compare the plantar fascia loading between untaped and taped conditions during running using a finite element foot model. Our goal was to provide more quantitative evidence to support treating plantar fasciitis with taping. In partial accordance with our hypothesis, strains on the fascia band (from proximal to distal portions) were significantly reduced by Fascia tapping compared to non-taping. Fascia taping also elevated the navicular height during the stance phase. However, there were no significant differences between Low-Dye taping and non-taping.

In our study, the predicted peak value of subtalar eversion fell within the suggested normal range of 8 degrees to 12 degrees for recreational running (DeLeo et al., 2004; Dierks et al., 2010; McClay and Manal, 1998) and our findings suggested that subtalar eversion was not influenced by either Low-Dye or Fascia tapping. Particularly, Low-Dye tapping did not constraint the subtalar motion as what it was structurally designed for. Previous studies usually evaluated Low-Dye taping on individuals with rearfoot overpronation and they reported that it had little to no effects in restricting rearfoot motion during dynamic tasks (O'Sullivan et al., 2008; Radford et al., 2006a). Low-Dye taping may be effective in foot pronation control under static conditions (Aishwarya and Sai, 2016; Whitaker et al., 2003), but the efficacy appeared to diminish shortly after walking or running (Harradine et al., 2001; Keenan and Tanner, 2001; Moss et al., 1993; Newell, 2012). Researchers attributed the loss of pronatory control of Low-Dye taping to its insufficient strap coverage to the heel region, which exerts poor leverage to counteract excessive rearfoot motions (Keenan and Tanner, 2001). This defect of Low-Dye taping seemed to also limit its supportive capacity to the foot arch and the plantar fascia during exercises. Little evidence was showed in previous studies that Low-Dye taping could elevate navicular height (Del Rossi et al., 2004; Holmes et al., 2002; Newell, 2012) and so was it in the current study.

Changes in navicular height and plantar fascia loading are closely related (Tao et al., 2010). Both of the two outcomes variables were reduced by Fascia taping. Despite, our simulations demonstrated that running with Fascia taping only elevated the navicular height in a small extent (1.75%) compared to those reported in walking and jogging (3.29–3.66%) (Ator et al., 1991; Holmes et al., 2002). Yet, this small change in foot arch deformation appeared to yield apparent strain offload (11.58–13.05%) from the plantar fascia, which was supported by the fact that heel pain alleviation may not be necessarily facilitated by substantial improvements in arch height (Jamali et al., 2004). We believed that our prediction of fascial strain was adequately reasonable under the premise that our findings in control condition (1.20–2.46%) were comparable to those (1.51–2.73%) of the barefoot running experiment conducted by McDonald (McDonald et al., 2016). The Fascia taping reduced fascial strains by 0.16–0.29% compared to nontaping. This magnitude of changes is similar to the measurements of a cadaveric study, which reported a strain alteration of 0.4% is approximately equivalent to windlass the fascia by dorsiflexing the first metatarsophalangeal joint for around 15 degrees (Carlson et al., 2000). Interestingly, patients of plantar fasciitis had significantly less active and passive range of motion in the first metatarsophalangeal joint possibly due to the reluctance to strains on the plantar fascia (Irving et al., 2006). In this regard, we speculated that the decreased fascial strain by Fascia taping, though at a small degree, might possess clinical importance in terms of either pain relief or function improvement.

Albeit not completely supported by our findings, the satisfactory clinical outcomes of using Low-Dye tapping is undeniable (Landorf and Menz, 2008; McNeill and Silvester, 2017). Many Low-Dye taping studies to date did not perform subject screening based on their rearfoot alignment (Ha et al., 2012; Landorf et al., 2005; Van Lunen et al., 2011). Studies controlling this factor were more consistent in the outcomes—pain relief was showed in patients with rearfoot overpronation (Abd El Salam and Abd Elhafz, 2011; Hyland et al., 2006), but not in those having a neutral foot (Radford et al., 2006b). As a result, we believed that Low-Dye taping is better indicated to patients of plantar fasciitis with rearfoot overpronation (Jamali et al., 2004). In contrast, Fascia tapping could be relatively more applicable to a boarder patient population since the design targets on the plantar fascia strain directly and shall not be attenuated by rearfoot alignment. Though there is currently lacking normative strain values for treatment evaluation and injury prognosis, our findings reported positive outcomes for

Fascia taping and it could be recommended for immediate pain management (Martin et al., 2014). Plantar fascia is a dense connective tissue that heals at a slower rate than other metabolically active tissues (Sharma and Maffulli, 2005). A small reduction of the fascial strain may facilitate the healing process on an accumulation basis. To increase the success rate, taping can be used in association with other treatment methods for long-term protection of the plantar fascia (Orchard, 2012).

To accommodate the patients' conditions, tapping modalities were usually modified and could vary greatly from their original forms (Lange et al., 2004). Thus, evaluating all kinds of tapping modalities remain infeasible. However, a mixed application of more than one taping modality seemed to address multiple factors (Bartold, 2004) and provide satisfactory outcomes (Hunt et al., 2004; Lange et al., 2004; Van Tonder et al., 2018). There was a complex interaction between the tapping structure and the foot-ankle. The benefits of taping treatment shall integrate the effects of biomechanical, neurophysiological, and psychological considerations (Franettovich et al., 2008). Further investigation is warranted to explore the underlying mechanism.

Results of this study should be interpreted in light of some limitations. First, it is widely recognized that the effectiveness of taping reduces over time due to the loss of material strength and adhesive to the skin (Ator et al., 1991; Holmes et al., 2002), as clinicians usually suggest renewing the tapping every a few days or even in a shorter interval. Therefore, our findings could only confine to one-time effects. Second, the intrinsic foot muscles were modeled as string units in our foot model, which only yielded passive strength when elongated. The active part of the intrinsic foot muscle force was ignored due to the technical difficulty in tracking its contraction. Third, existing studies on plantar fascia injuries almost adopted a strain-based fatigue model to understand its pathologies, so did the current study. However, the progression of plantar fasciitis may be further explored by other fatigue models, e.g. an energy-based model. Plantar fascia is a stretchable tissue and it is subject to uniaxial loading (mainly tensile force along its longitudinal and forces in other directions are relatively small) in constant load cycles during human locomotion (Wearing et al., 2006b), which makes it easily fit into a strain-based fatigue model. The model may be reluctant to identifying a strain threshold/range that triggers injuries. Given the chronic and lengthy nature of plantar fasciitis, knowledge of energy consumption of the fascia band many give a better insight into the traumatic procedure as the model accounts for both loading magnitude and duration. Future studies are suggested to address this aspect.

CHAPTER 7 CONCLUSION AND SUGGESTIONS FOR FUTURE RESEARCH

7.1. Significance of the study

Shear wave elastography is a novice ultrasound-based technology that currently has a limited application in the plantar fascia. Elastography allows non-invasive assessments to deep human tissues while also provides robust outcomes that directly reflect changes in the material property. The present study is one of the few that not only explores the feasibility of performing elastographic measures to the plantar fascia, but also ensures the accuracy of the measures by incorporating a quality-control algorithm to the workflow of imaging processing.

The finite element foot model developed and used in the present study creatively applied the method of SPH to model the bulk soft tissue, which, to the best knowledge, is fairly rare in the study field of human movement biomechanics. SPH allows an analysis of large elementary deformation and renders simulations of intense sports, e.g., running, less problematic from the technical standpoint. The present study also shows that a finite element model integrating this modeling method could provide validated results of internal stress/strain on the soft tissues.

Running is a highly dynamic movement associated with large impacts and momentums applied to the musculoskeletal system. By using the above-mentioned foot model, the present study was able to perform a series of explicit dynamic simulations of running to better account for the forces and momentum imposed on the plantar fascia, which is a complement and advance to a field currently dominated by implicit static approaches.

The study identified a reduced plantar fascia elasticity in forefoot strikers and considered tissue overload as its primary cause. This possible explanation was verified in part by computational simulations of both running styles, wherein an increased tensile force applied on the plantar fascia was found in forefoot strike running. A plantar fascia with reduced elasticity is less resistant to strain and more vulnerable to injuries. In this regard, the study also provided quantitative evidence to support using taping treatment to offload and protect the plantar fascia during running.

7.2. Limitations

First, all of the included studies were cross-sectional with a relatively small sample size. In spite of the statistically significant differences among the experimental conditions, the causal relationship between interventions and effects in each study is inconclusive. Possible explanations to the research outcomes were discussed accordingly while a firm conclusion should be drawn upon further studies. Second, some technical limitations were involved in the process of measurements and data collection. Elastography-based imaging is still poorly adaptive to soft tissues that are too fibrous, for an example the plantar fascia at its origin near the calcaneus, which could reduce the overall qualities of the measurements. Elastography also introduces bias in the outcomes when assesses tissues that are not uniform in thickness. The finite element foot model was established based on a certain level of simplifications on the tissue geometries and material properties. Though it is a common practice in studies using a theoretical framework, this approach inevitably affects the accuracy of the prediction outcomes. Another important factor that was ignored in all studies was the intrinsic foot muscle force. Due to the close myofascial linkage of the foot muscles and plantar soft tissues, outputs from the intrinsic foot muscle could influence loading on the plantar fascia as well as its responses to the changing mechanical environment. However, observing the activities of the intrinsic foot muscles is difficult with the strength of existing technology. Until a ground-breaking method emerges, the results of studies using similar approaches should be interpreted in light of this limitation.

7.3. Conclusion

Results of the study showed that there was a trend of reduced plantar fascia elasticity in forefoot strikers compared to rearfoot strikers. On the premise of assuming similar anthropometries, running regimen, and running experiences between the two types of runners, the loss of plantar fascia elasticity in forefoot strikers was likely associated with tissue rapture caused by a repeated overload on the facial band. The conjunction was supported by the outcomes of computational simulations of running with both foot strike techniques. Forefoot strike running was estimated to impose a higher tensile force on the plantar fascia than rearfoot strike running. Though the threshold values of force for injury prediction is currently unknown, the increased loading on the plantar fascia could be a potential threat that predisposed forefoot strikers to foot problems on an accumulative basis. In the meantime, finite element analyses also demonstrated that taping treatment was effective in offloading the plantar fascia during running. For runners willing to continue training while seeking an approach of providing on-site protection to the plantar fascia, taping could be one of the options to choose.

7.4. Future direction

Future studies are expected to, on one hand, establish a structural constitutive model of the plantar fascia that incorporates fiber recruitment and accounts for material yield strength at the sub-macro level. The model can be used to detect microtears on the fascial band and indicate the risky loading range for the plantar fascia. On the other than, a controlled, longitudinal study can be conducted to observe and verify the effects of forefoot strike running as well as taping treatment on the incidence rate of plantar fasciitis.

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