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DOCTORAL (PHD) THESIS

FENGQIN FU

Enhancement of runners' performance and protection by alternative longitudinal bending stiffness of the shoes

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DECLARATION AGAINST PLAGIARISM AND CORRECT REFERENCING

I, Fengqin Fu, declare that the dissertation entitled: Enhancement of runners' performance and protection by alternative longitudinal bending stiffness of the shoes is my original work where the references given in the Reference list are used. All sections, which are transcribed or rewritten from some publications, they are correctly referenced.

Date: 2022.10.05

Fu Fengqin

(Signature)

Content

The motivation for the work

Since I graduated from postgraduate school, I have worked in the Xtep Science lab as a senior biomechanical engineer. My work is to manage product performance research to increase understanding of athletes and consumer needs and specific functional requirements to develop technologically innovative footwear, especially running shoes. The research and development (R&D) innovation center include science lab, shoe sample team, material team (sole and fabrics), and designer team. Due to my work, I am required to deeply research sports movement and their effect on human beings. This is the reason why my PhD aims to answer questions about running.

It is worthy to note that running is one of the most popular types of exercise. With improved living standards, simple sports shoes could no longer meet people's needs, and people more and more favored professional, functional running shoes. Athletic shoe optimization relied on the improvement of materials and structures. The sole was an essential aspect of the available design of a running shoe, and changing to any part of the sole would have an effect on the shoe's functioning, so the research on the sole should not be limited to the sole as a whole, but should be more refined. In addition, as part of the sole of a running shoe, the midsole had a critical role in the athletic performance of running. Running performance means characteristics of running ability or running economic. Material hardness, stiffness, tear strength, other physical indicators, and the shape of the structure, area, and other factors would affect the runner's sports performance. Recently, runners debated running shoes' longitudinal bending stiffness (LBS). LBS was the moment required to bend around the metatarsophalangeal joint area of an athletic shoe, which was influenced by the material and structure of the sole and was a vital test index in athletic shoe research. The motivation was double, providing suggestions for athletic shoe optimization and enhancing the competitiveness of the laboratory. Thus, I started to focus on the LBS.

At the beginning of my research, the R&D innovation team made test shoes according to my requirements. Besides, the company’s running club has runners from all over the country, totaling over 800,000. It had runners in different situations, like runners of different running levels, genders, regions, running injury experiences, etc. In other words, this study could recruit runners based on needs. They had found suggestions and solutions for manufacturing combining the LBS of running shoes and the physiological characteristics, needs, and wearing habits. Male and female runners had distinct features in running biomechanics and wearing patterns (e.g., women had had the experience of wearing high heels since ancient times). Thus, it was necessary to combine their physiological and biomechanical characteristics and perhaps consider their wearing habits when studying running shoes for female runners. A pair of shoes should meet consumers' requirements for functionality and their needs for subjective feelings such as comfort and fit of running shoes. Through personal testing, the impact of different shoes on people's emotional feelings could be understood. At the same time, it was essential to understand the effect of such changes on human subjective evaluation. Thus, combining sports biomechanical tests, finite element simulations, and subjective tests, gait characteristics during running, were studied by changing the controllable factors of midsole materials.

Research Objectives

Based on the motivation, I targeted to answer three important questions with regard to shoe design. Therefore, my objectives are the followings:

My first objective is to investigate whether a forefoot carbon-fiber plate, inserted into the midsole, can alter physiological properties such as plantarflexion angular velocity, power, etc. in order to achieve higher running performance. It is also my question How much can the shape of the plate alter the running performance?

My second objective is to identify those factors or parameters, which have the most influence on gait movement, and to implement them in our shoe design methodology. In this objective, I need to carry out gait analysis on a group of people who regularly wear high-heel shoes and on another group who are inexperienced in these sorts of shoes. By means of experiments, the common and different parameters should be separated and highlighted.

My third objective is to determine the structure and material of midsole, according to the characteristic of a group of people who regularly wear high-heel shoes. What kind of structure of midsole is benefit to motion control? Which material is better for improving running performance? Last but most important, how can combine reducing risk of injuries and running economic with the regard to shoe design?

Methodology of the dissertation

In this doctoral work, the complex method is used to address problems involving experimental measurement, finite element simulation, and subjective perception methods. Firstly, the thesis begins with the experimental measurement and finite element simulation to investigate the independent effect of the construction of the forefoot carbon-fiber plate inserted into the midsole on running. For the finite element simulation, the outsole, midsole, and two kinds of carbon fiber plate had been modeled in 3D based on an industrial 2D shoe design drawing with Rhino 6 Computer-Aided Design (CAD) software (Robert McNeel & Assoc, Seattle, WA, USA). All of the solid parts were assembled into a whole sole model, then imported into the FE package ABAQUS (Dassault Systemes Simulia Corp, Johnston, RI, USA) to develop the numerical model. A Vicon motion system with 10 cameras (Vantage 5, Vicon, Metrics Ltd., Oxford, UK) was used to capture kinematic data, the floor force plates (combined dimensions 270 × 60 cm, 1000 Hz (AMTI, Watertown, MA, USA) was used to collect kinetic data during experimental.

Secondly, a force platform (Kistler, Switzerland) and the 8-camera Vicon motion analysis system (Oxford Metrics Ltd., Oxford, UK) were used to investigate the differences in lower limb kinematics and kinetics between experienced (EW) and inexperienced (IEW) moderate high-heel wearers during jogging and running.

Thirdly, we got the mechanical data about shoes by an impact tester and flexion tester (Brentwood, NH, USA) before experimental measurement. A 10-camera motion analysis system (Vantage 5, Vicon, Metrics Ltd., Oxford, UK) and the floor force plates (AMTI, Watertown, MA, USA) to capture biomechanical data. Also, fifteen cm visual analogue scale (VAS) was carried out to apply for running footwear assessment.

1 Literature review

1.1 Running

Run regularly for fitness and maintain a healthy lifestyle, and many cities in Western societies had recreational running events [1, 2]. In recent years, China has also launched a "marathon fever," the number of urban marathons held in China and the number of participants was also increasing year by year. According to official statistics, the number of marathons increased about nine times in these five years from 2010 to 2015 [3]. The ground reaction force was responsible for converting the biological energy provided by muscles into kinetic energy, which the body needs to move forward. Hence, the movement technology of the lower limbs was an essential part of the running movement technology. A reasonable technical action could help the human body minimize energy loss and convert the biological energy generated by muscle contraction into kinetic energy for forwarding movement. An in-depth study of the characteristics of the human running lower limb work could be a more comprehensive understanding of the technical characteristics of the running process, for improving the running technology of the public runners and reducing the risk of injury caused by running was very important.

1.1.1 Anatomical structure and activity characteristics of foot joints

The foot joints include the ankle, intertarsal, metatarsophalangeal, intermetatarsal, metatarsophalangeal, and interphalangeal joints. The ankle joint was the most mobile and functional, and the metatarsophalangeal joint was the most associated with forwarding propulsion. The forefoot consists of five metatarsal heads and the proximal phalangeal base of each toe, collectively referred to as the metatarsophalangeal joint (MTPJ) which has two axes of motion. The sagittal plane around the coronal axis was flexion (plantarflexion) and extension (dorsiflexion). In the horizontal plane around the sagittal axis, a small range of adduction (phalanges together) and abduction (phalanges apart). Flexed and extended were together during movement (**Figure 1**).

An accurate assessment of joint mobility helped the therapist determine treatment cycles, assess treatment status, and evaluate treatment outcomes. The MTPJ played a vital role as the final part of the movement when running, jumping, and other activities off the ground [4]. The human anatomy showed that the active extension (dorsiflexion) of the MTPJ could reach 50-60°, a passive extension could reach 90°, and active flexion could reach 45-50° under normal conditions [5]. In addition, the range of motion of the MTPJ was up to 51.9° of active extension, 67.6° of passive stretching, and 39.1° of active flexion [6].

The mobility of the MTPJ could reach 31.5° and 22.6° during the running and jumping phases [4]. But the MTPJ was not exposed when wearing shoes for a range of daily activities, so runners must consider the impact of shoes. The rotation of the MTPJ by joint flexion and extension of the first to fifth MTPJs was not a straight line, so the different definitions of the axis of rotation during the study would also cause different results of the relevant parameters. Smith et al. [5] defined the joint rotation axis of the MTPJ differently when studying the effect of LBS on the MTPJ, defining the line connecting the centers of the first and second MTPJs as the first part and the line connecting the centers of the second and fifth MTPJs as the second part. This way of definition could more accurately describe the motion of the MTPJ and obtain more precise kinematic parameters. Studies had found that increasing LBS improves athletes' running efficiency by between 1% and 2% [6, 7].

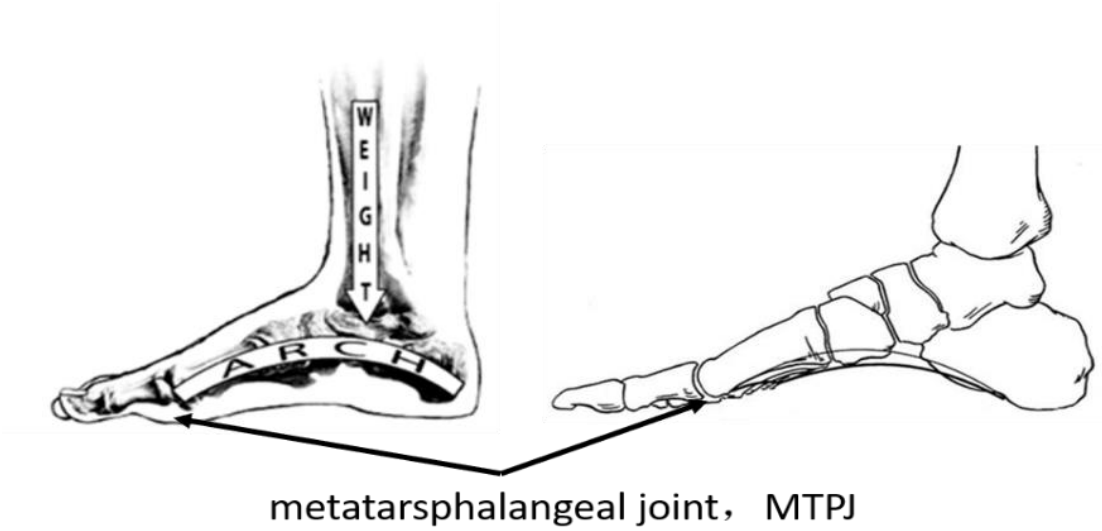


Figure 1. The position of metatarsophalangeal joint (MTPJ)

1.1.2 Overview of running on biomechanical

The measured parameters of running biomechanics included kinematic, kinetic, and biological parameters. The measurement of kinematic parameters contained temporal parameters of motion, spatial parameters, and period division, such as the center of gravity velocity, joint angular velocity, joint angle, motion trajectory, etc. The measurement of kinetic parameters contained ground force and reaction force, joint moment, collective power, rotational inertia, etc. Biological parameters included anthropometry, electromyography, and other physical factors [8]. All in all, running performance can be improved by decreasing the eccentric work performed by muscles and the associated mechanical energy dissipated at a joint.

The kinematics of running

Kinematics was the basis for studying the biomechanics of running, for example, through kinematic analysis studies that could provide a rational joint motion pattern for the lower extremity movements of a single running cycle. The movement of the lower limbs during running was mainly flexion and extension and included the rise and fall of the center of gravity, the degree of tilt of the trunk, and the change of the ground angle. Other biomechanical characteristics could be studied and analyzed by measuring their kinematic parameters for the technical movements during running. Kinematics only examines the laws of the velocity and position of an object over time. The cause of changes in the body's position and state of motion was not discussed.

The hip joint consists of the femoral head and acetabulum and was a multiaxial ball and socket joint, the most significant and most stable joint in the body. The joint capsule was relatively tight and challenging because the femoral head was deeply embedded in the acetabular fossa. The hip joint was capable of flexion, extension, extension, extension retraction, internal rotation, external rotation, and circular motion in three axes restricted by ligaments. The range of motion in each axis around the hip was much less than that of the shoulder was also a ball and socket joint, which has excellent stability to accommodate weight-bearing and perform the functions of walking, running, and jumping. In the sagittal plane, the hip joint could perform flexion and extension movements, with flexion ranging from 0° to 140° and extension ranging from 0° to 15°. In the frontal plane, the hip joint could perform adduction and abduction movements, with adduction going from 0° to 25° and abduction ranging slightly more from 0° to 30°. The hip joint could accomplish internal and external rotation movements in the horizontal plane. The angular range of internal and external rotation movement was not the same under different flexion angles of the hip joint. When the hip joint flexes, the maximum internal rotation could reach 70°, and the entire external process could get 90°. When the hip joint become less flexed or straightened, the hip joint's internal and external angle naturally decreases due to the limitation of soft tissues.

When the foot was about to touch down, the hip joint was about 25° to 30°. The hip joint was gradually flexed and extended along with the ankle and knee joints when the body's center of gravity exceeds the support surface. When the foot was about to toe-off, the angle between thigh and ground would increase by 22°, and the angle of the hip joint would increase by 20°. The faster the running speed was, the stronger the backpedaling force would be, and the angle of the hip joint would increase by more than 5° [9]. There was a correspondence between the degree of human muscle recruitment and the hip flexion angle during running. For example, the relationship between the hip flexion angle and the degree of muscle activation of the gluteus medius in the supine position has been viewed differently by scholars in related fields [8,10, 11]. From the research of Michael [10], the gluteus medius was activated at a moderate level when the subjects performed the clam exercise with hip flexion angles of 30° and 60°. It has been shown that the activation of the gluteus medius was affected by the increase in hip flexion angle, which led to a significant decrease in the activation of the vastus lateralis muscle. The final results showed a better activation of the gluteus medius and lower activation of the vastus medialis while the hip was in flexion and abduction [11]. Other studies had reported that abduction with internal hip rotation increased gluteus medius activation more significantly than external rotation [12].

The knee joint was the greatest and most complicated joint in the lower limb. It was both a talocrural and elliptical joint. The knee joint was comprised of the femoro-tibial joint and the femoro-patellar joint. The femorotibial joint was consisted of the medial and lateral condyles of the femur opposite the medial and lateral condyles of the tibia, forming an elliptical joint. The femoral-patellar joint consisted of the femur's patellar surface that meets the patella's articular surface, including the gliding joint. The knee joint could only be flexed and extended. Still, due to the unique characteristics of the knee joint structure and the soft tissues near the joint, internally and externally rotated within a specific range of angles. In the sagittal plane, the knee has the greatest range of motion in flexion, reaching 0° to 140°. In the frontal plane, the tibial and peroneal collateral ligaments become tense during knee extension and relax during knee flexion, resulting in essentially no mobility in the frontal plane when the knee extends. When the knee flexes to 30°, the abduction and adduction angles were smaller, averaging about 11° in total. During jogging, the knee joint ranges 90° in the sagittal plane, 30° in the frontal plane, and 40° in the horizontal plane throughout the stance phase. The lateral femoris, vastus medialis, and vastus medialis were the main muscles that control knee flexion and extension. The insufficient strength of the main muscles would affect the knee joint's function flexion-extension the sagittal plane dominating the motion. In addition to the importance of the hip joint mentioned above, the knee joint was also an essential part of running exercise research. The knee joint angle differed at touchdown due to different landing patterns during running. The forefoot landing increased knee flexion and a more vertical lower leg than heel landing movement patterns. In contrast, runners who landed on their heels had a lower joint angle of the knee at touchdown [12].

The ankle joint was capable of dorsiflexion (extension) and plantarflexion (flexion) in the sagittal plane. The ankle joint, also known as the talofibular joint, was consisted of the lower tibia, lower fibula, and talofibular carriage. It was a near uniaxial flexion joint with a variable axis of rotation during dorsiflexion or plantarflexion. The anterior part of the wider carriage was embedded in the joint socket when dorsiflexed, making the ankle joint more stable. The foot could make slight lateral directional movements. The joint was not durable enough because the narrower posterior part of the slide enters the joint socket. Therefore, most ankle joint sprains occurred in the state of plantarflexion. The maximum range of motion of the ankle joint was the movement of dorsiflexion and plantarflexion in the sagittal plane. The content of the activity of inversion and eversion in the frontal plane, and internal and external rotation in the horizontal plane were relatively small. Whether professional athletes or runners, most people were used to landing heel first during jogging. At the moment of landing, the ankle joint would be plantarflexed about 5°, then the knee joint would be bent, the lower leg would be continued to move forward relative to the foot, and the ankle joint would be dorsiflexed about 20°. In the middle of the support phase of running, the ankle joint dorsiflexion reached its maximum level, and the knee joint flexion would also be at its maximum angle. After the foot toe-off from the ground, the ankle joint would be at its maximum plantarflexion angle, approximately 70° [9, 11].

Ground reaction force (GRF)

It was well known that GRF was a crucial indicator in studying lower extremity dynamics in running sports techniques. In the running process, the human foot, as the end link of the human lower limb, would directly contact the ground and thus generate ground support reaction force. In running kinetics studies, researchers usually used a three-dimensional (3D) force measurement system to collect the 3D GRF and related indexes during the running support phase of subjects. Studies had shown a significant correlation between the first peak force and running injuries [13]. The GRF in the front and rear directions was also divided into braking and propulsion. When the foot starts to land until the body's center of gravity exceeds the support surface, the lower limb would produce braking resistance at this stage, and the running speed would naturally be disturbed and slowed down. After the body's center of gravity exceeds the support surface, the lower extremities would appear again to promote running to continue to travel. Therefore, in the study of running technical movements, it was necessary to reduce the braking force and increase the propulsive force to improve the running efficiency. The internal and external GRF could determine the trajectory of the runner in the running process. It would lead to the runner's running rotation being a curved march when the inner and outward direction of the force was too much on the runner, thus consuming too much runner's physical energy, slowing down the running speed, reduce running efficiency [14]. Studies had shown that during walking, the lower extremities to withstand the GRF could reach up to 1.5 times the body weight of the human body. The lower extremities to resist the GRF may reach 2 to 3 times the body weight during running [15]. When a person lands in a jumping motion, the GRF on the lower limb would be seven times the bodyweight [16].

The impact of the ground during running was affected by many factors, such as the body's mass, soft tissues, running speed, the center of gravity at the moment of touchdown, and so on [17]. Some scholars in the study of different landing methods of runners had found that the runner using the rear foot landing in the heel landing, the vertical GRF would immediately produce a peak impact force. The first impact peak was about two times the runner's weight. The second peak in the vertical direction of the GRF usually occurred late in the stance phase, in the range of 60% to 75%, and the second peak lasted longer about 200 ms [17, 19]. The study also indicated that the deceleration of the trunk and swing leg during running and the braking of the support leg determined the magnitude of the second peak force [17].

1.1.3 Injuries and Gender differences in lower extremity mechanics during running

A fatigue run was considered exhaustion or a distance of more than 3000 meters, considered a long-distance run [18]. According to the definition of injury and follow-up survey, the proportion of runners injured was between 11% and 85% [21-23]. The lower limbs and feet were continuously subjected to high-frequency stress in the running, which was very likely to cause overwork injury to the lower limbs and feet. The most common site of injury was the lower extremity. Common injuries included Achilles tendonitis, plantar fasciitis, tibial pain, fatigue fractures, and patellofemoral joint pain (runner's knee) [6, 19]. It was reported that fatigue caused running kinematics and dynamics changes that could increase the risk of injury [20-22]. These changes included improved impact acceleration, increased trunk tilt, and ankle eversion changes in maximum knee extension angle and knee flexion angle [21]. Most kinematic data in the long-duration running included speed, stride frequency, stride length, touchdown time, and time to vacate. There was no significant stride frequency or stride length during running. An increase in step frequency and reduction in stride length was linked to reducing the risk of injury [13,23, 24].

Some sports injury studies had also involved hip motion. Some studies had suggested that proximal activity may be associated with knee injuries. Suppose the proximal joint was not stable enough. In that case, it may cause more hip pronation in the lower extremity, which leads to increased knee eversion angle and ultimately increases the risk of lower extremity injury in runners. Some studies had reported that external and internal rotation of the hip during the stance phase may potentially affect the kinematics of the entire lower extremity. It was also reported that the internal angle on the hip joint was correlated with knee injuries [2, 25], which was higher in the female group than that of the male [26]. Excessive hip pronation may increase the tension on the supporting phase of the iliotibial bundle during running. This increased tension may damage the iliotibial pile during repetitive running [2]. The hip joint's angle change and the knee joint's angle change were inseparable, and the two compensate for each other to support the running movement. Excessive internal and internal rotation of the hip joint would cause the center of the knee joint to move inward relative to the foot. When the foot was fixed on the ground, the inward movement of the knee joint would cause the tibia to abduct and the foot to rotate forward, finally causing the knee joint to increase in the angle of eversion. It was suggested that increased patellofemoral joint stress in runners with anterior knee pain may be caused by an increase in dynamic Q angle due to excessive hip adduction [27]. Studies had shown that hip pronation was a significant factor in dynamic knee eversion. Excessive hip pronation also pulled on the soft tissues limited knee eversion (e.g., anterior cruciate ligament, medial patellar ligament, etc.) [28].

The knee joint was one of the joints with the most cushioning in the lower extremities of the human body, and studies on running injuries had shown that the most common injury style was on the knee joint. The physical function gradually decreased when fatigue sets in, which led to a series of biomechanical changing, such as an increasing knee flexion angle [29]. In terms of running economy, relevant studies had shown a significant positive correlation between running economy and knee flexion angle at the moment of toe-off [30]. The reason may be that the increase of knee flexion angle makes the energy consumption during flexion decrease, and the lower limb rotation inertia decreases. In contrast, the lower limb extensor muscle group was in the correct initial part and power position, increasing the pedaling force [31]. Lenhart et al. [32] showed that increased knee flexion angle increased the moment of mobilization of the knee extensor group, which correlated with higher patellofemoral joint pressure during landing. Dierks et al. [33, 34] reported that distance runners with anterior knee pain exhibited smaller maximum knee flexion angles.

In contrast, a survey by Kulmala et al. [35] indicated that a smaller maximum knee flexion angle during the support phase of running healthy runners with forefoot landing (forefoot) induced patellofemoral joint contact forces of 16%. From the view of runners, an increase in ankle eversion angle increased the risk of sports injury. In one study, the ankle eversion angle was 20°. The tibial internal rotation angle was 13° and 3° in two subjects, suggesting that more significant tibial internal rotation may significantly affect lower extremity knee injuries. In contrast, the same knee eversion angle does not necessarily lead to the same tibial internal rotation angle [36]. Scholars had different views on the correlation between ankle dorsiflexion motion and lower extremity force lines and injuries. When ankle dorsiflexion was limited, it reduced the proximal joint's mobility in the sagittal plane, such as the knee joint [37]. There was a potential relationship between knee and ankle kinematics when ankle dorsiflexion was limited. Internal displacement of the knee may increase the risk of Anterior Cruciate Ligament (ACL) injury [38]. During the stance phase of running, the tibia's movement would be affected relative to the talus when the ankle dorsiflexion was restricted, which led to the anterior rotation of the subtalar joint compensating and was associated with anterior knee pain injury [39]. Kuhman et al. [40] compared the kinematic parameters of college cross-country runners with and without damage, showing no significant difference in ankle dorsiflexion angle.

For lower extremity joints kinematics, reduced hip motion associated with trunk position was used to reduce the risk of injury [41]. Fatigue resulted in a series of kinematic changes, such as a significant increase in mid-lateral acceleration, a significant decrease in stride regularity, heel lift, a significant reduction in knee rotation speed during the swing, and decreased knee posterior displacement speed [27, 42, 48]. These kinematic changes ultimately impacted performance. There was a significant fatigue decrease in vertical GRF, whereas one study found no difference in GRF during a fixed speed circle. Those studies that found substantial vertical GRF also reported significant reductions in propulsion [43, 44]. Girard et al. [43] also found that braking force decreased when speed decreased during self-selected running speed but not during constant speed runs. Seventy-six percent of knee pain was found in women. In a prospective study of patellofemoral pain, most patellofemoral problems were found in young women [45]. It was twice to suffer specific running injuries in female runners than that of male runners [31]. It was related to the fact that male and female runners had significantly different physiological characteristics and sports running posture. A greater hip internal rotation, vertical GRF, accessible vertical torque, peak hip flexion angle, and negative work were displayed in females than that in men [31]. Women had a greater ratio of hip-width to femur length, which resulted in greater hip internal rotation. Women exhibiting higher Q angles would increase lateral quadriceps pull on the patella. It would exacerbate patellar tenderness or recurrent lateral patellar subluxation conditions, which induced a higher incidence of patellofemoral joint pain [45].

In summary, in the form of national fitness, more and more young people like to participate in running, and injuries would reduce the fun of sports. Repeated injuries would lead to temporary or even permanent stop running. Therefore, the research on sports running shoes and the design for gender differences between men and women played a crucial role in promoting running sports development. However, there is no experimental evidence supporting this idea and future studies will need to investigate the influence of racing running shoes with a stiff plate on long-distance running performance for female runners.

1.2 Research on running shoes

1.2.1 Running shoe characteristics

The body's energy expenditure consisted of three central elements: the maintenance of the primary metabolism, the energy consumed during exercise, and the thermogenetic energy of things. The impact of running shoes on energy metabolism mainly affected the energy consumed during exercise. Today's running shoes were broadly classified into cushioning, motion control, and a neutral type between these two categories, depending on the functional requirements. Modern running shoes relied on the shoe last to determine the shape and fit of the shoe. The outermost outsole mainly played the function of anti-slip and wear resistance, while the upper and other designs played the role of beauty and breathability. As the most technologically advanced part of running shoes, the midsole could be protected, controlled, or cushioned by different materials [46].

In one study, subjects were asked to wear three running shoes with different heel camber (one with no heel camber. one with 16° heel camber. and one with negative heel camber and curvature) for a running test. The study results showed that the ankle eversion angle decreased as the lateral inclination of the running shoe decreased during the support phase of running. In contrast, the change in the lateral tip of the running shoe didn’t change the three-dimensional angle of the ankle and knee joints before the subject touched the ground

[47]. The research of Nicolas [53] found that the knee joint's flexion angle decreased while the dorsiflexion angle of the ankle joint increased during the support phase among barefoot runners. In a study by Roberts and Birch [48], the kinematic parameters at the moment of touchdown were compared between barefoot and MBT shoes. The results showed that the ankle eversion and ankle dorsiflexion angles at the touchdown the moment of touchdown were significantly lower barefoot than that in shoes.

The energy cost during running would be affected by mass, cushioning, and LBS of the shoes. Lighter shoes reduced the energy cost by reducing the inertia of the swinging foot, which was translated directly into faster athletic performance. Barefoot running was the most appropriate in terms of shoe mass but not in terms of energy, which required more extraordinary muscle work to cushion the impact of the foot against the ground [49]. The kill point required to be observed in energy recovery was the return of energy at the suitable time at the appropriate frequency in the proper location [50]. Cushioning-related materials of running shoes were generally not suitable energy return materials. The heel's location of maximum possible energy storage was not the location where the return energy could be effectively utilized. Because the forefoot was suitable for returning energy while the rearfoot was for storing energy, the energy generated was consumed, stored, or returned. Before leaving the ground running, the rear foot rolled to the forefoot only dorsiflexion without plantar flexion. In contrast, the forefoot could only be destroyed or returned. Therefore, reducing exertion could improve energy return and enhance athletic performance. On this basis, the researchers suggested that minimizing energy loss positively affected athletic shoe performance compared to maximizing energy return.

As the MTPJ was an energy-absorbing joint, early studies had focused on how to minimize the energy release of the MTPJ during exercise. Willwacher et al. [4] found increased LBS of the running shoe changing time like ground contact time and push-off time. Kram and Taylor [51] suggested that faster pace and shorter contact times required faster cross-bridge cycling and more adenosine triphosphate, and the energy expended by muscle contraction during running was inversely proportional to contact time. The angular impulse of the ankle, knee and hip joints increased slightly when the LBS level exceeded the critical LBS, while the angular stimulation of the MTPJ decreased significantly [52]. Running to save energy could reduce energy metabolism by reducing the lower limb joints' work. A previous study reported two different running strategies that could be used to maintain steady running. One group showed increased ankle torque to compensate for the more extended force arm, while the other group showed increased stance time with no significant change in ankle torque [53]. These results supported the idea that complementary changes in lower extremity joint torque were required to maintain stable running in stiffer shoes. The metabolic cost of running may vary due to the different muscle groups involved in joint moment-driven changes. The MTPJ was a joint that mainly does centrifugal flexion and absorbs energy with little positive work during running [54-56], so minimizing energy loss at the MTPJ was essential for the research of sports shoes and the development of top sports equipment.

1.3 A widespread application to increase the LBS of shoes

As a midsole characteristic, LBS refers to the moment of force required to bend the forefoot of a shoe at the MTPJ per unit turn, which has received less attention than other sneaker characteristics such as outsole traction and midsole cushioning. In terms of athletic ability, wearing higher LBS athletic shoes could improve the performance of sprinting, jumping, side-cutting, and other sports. In terms of injury, many forefoot injuries may occur during sports activities due to a large number of bones, muscles, and ligaments in it. The most customary injuries at MTPJ included metatarsal stress fractures, metatarsalgia, bunions, and sprains of the first MTPJ (often referred to as turf toe). In recent years, increased LBS has been considered a treatment for these MTPJ injuries and a preventive method to reduce the risk of these injuries in athletes or reduce forefoot extension to treat injuries such as turf toe [63, 64].

Studies had shown that the LBS of shoes directly affected the performance of various sports such as running, jumping, sprinting, and multidirectional movements. Lam et al. [57] analyzed the effect of different part LBS changes by changing the LBS of other parts on vertical jump and sprint in order to simulate basketball game conditions. The study results showed that participants wearing stiffer shoes (i.e., with carbon fiber panels added to the medial and lateral plates) improved their average running vertical jump by 1.7 cm and their athletic performance by 2.9%. In contrast, there was no difference in vertical jump performance in the study by Worobets et al. [58], only a moderate effect of LBS was found on sprint and lateral cut performance. It was probably related to the fact that Worobets and Wannop [58] changed the LBS of the shoe by cutting out a small vertical portion of the shoe outsole on the outside of the shoe, which may cause structural damage to their prototype shoes, affecting jumping performance. The increased LBS of shoes also improves fatigued athletes' reverse squat jump performance [59]. Tinoco arranged 12 athletes according to their performance in the Counter Movement Jump (CMJ), with the odd-numbered and even-numbered groups ranked. The full sprint of 20 meters was repeated eight times to make them tired. Finally, it was found that the jump performance of the athletes with softer midsole decreased significantly by 16.1% after fatigue training. Meanwhile, the athletes with stiff midsole only reduced by 9%. It was suggested that increasing the LBS of shoes under fatigue conditions could improve athletic performance. Tinoco also studied the effect of LBS on the performance of a multi-directional sprint test with lateral braking and cutting motion using a specific route and found that an increase in LBS also improved athletic performance. Worobets and Wannop [23] showed similar results in a study with 20 basketball players. They found that the performance of timed eager exercises increased by 1.7% when the bending LBS was increased by 50% (0.22-0.33Nm/deg).

Moreover, it could improve the athletic performance of jumping movements, such as increasing the height of running touch, squat jump, bounce dump jump, and implementing multi-directional sprint movements [4, 60, 67]. The study indicated that in the fatigue scheme of running and jumping combination, the height difference of CMJ before and after the fatigue of shoes with high LBS was 9% lower than that of shoes with low LBS [59]. It meant that the motion performance of shoes with high LBS in the fatigue state was reduced less. With recent research, the LBS of running shoes has received more and more attention, as it has the potential to affect both injury and performance. According to previous studies, changing the LBS of the sole within a suitable range improves not only the performance of running and sprinting but also the running economy to some extent [52, 55, 56].

The results discussed above indicated that increasing the LBS of the shoes could improve athletic performance. In the experiments of Roy et al. [61] and Stefanyshyn et al. [7], running economy, collective energy, and electromyography data were collected from 13 subjects. It was found that when the subjects ran on a stiffer midsole, there were approximately 1% energy savings compared to the control midsole. Compared to the control shoes, oxygen consumption decreased more in the heavier subjects with a stiff midsole. Energy absorption at the MTPJ was not significantly different in the stiffer midsole than in the control shoes of it. A stiffer shoe with natural flexion of the MTPJ was beneficial to a reduction in running energy loss [58].

In brief, it would be reduced the energy loss performed at the metatarsophalangeal when running with a stiff running shoe (increasing the LBS of running shoes). However, the role of carbon fibre plates embedded in running footwear midsoles on running injuries remains unclear, and the the mechanism of LBS is not yet fully known in details today. So it’s important to better understand the mechanism and quantify the LBS‘s impact factors.

1.4 The influence of LBS on forefoot area on running performance from the view of biomechanical

Some scholars now focused on the metatarsophalangeal joint's motion mechanism during sports [51, 61]. Krell et al. [62] collected kinematic data from a large number of Olympic athletes in the 100-meter run and found that the angle at the moment of touchdown and the peak extension velocity on the MTPJ were significantly correlated with their race performance. Stefanyshyn's team asked athletes to wear sneakers with different LBSs for a 40m sprint to change the flexion and extension of the MTPJ and found that the athletes' performance varied considerably [56]. Some domestic scholars also found that the MTPJ was constrained and could not walk as freely as usual in special hard-soled shoes, which showed a reduced range of motion. Still, to complete the walking movement, other joints would show corresponding compensatory signs, among which the ankle and knee joints contributed more to the adjustment of gait [20]. Some scholars had found that there was a correlation between LBS and running economy. It could improve running economy and save energy when the LBS of running shoes was appropriate, while a large or small LBS was less economical [63]. In a study by Qiujie Li et al. [64], it was shown that the LBS of running shoes showed a significant "U" shaped relationship with the mechanical negative work at the MTPJ during jogging. Also, in comparing different LBS of sports shoes, it was found that the center of pressure and the direction of pressure in the human body was shifted after the intervention with additional LBS of sports shoes. The reason for that was the degree of ankle valgus, hip adduction and pelvic tilt decreased as the LBS of the shoe sole increased. it was affected the internal load of the hip, knee, and ankle joints, and even caused injury. A study in which patients with plantar fasciitis wore athletic shoes with three different midfoot stiffnesses found that the angle of midfoot bending was highly correlated with the LBS of the different shoes (R=0.8839) [65]. Nigg et al. [50] also demonstrated that increased the LBS of shoes induced an increased positive work at the MTPJ. It was similar to the findings of Toon et al. [14] and Willwacher et al. [54]. Willwacher et al. [54] emphasized that the net joint work was systematically shifted positively in the rigid condition compared to the control condition. Supplementary information showed an increase in net MTPJ work and a decrease in negative work in the high LBS condition among ten subjects. The author also suggested that these energy rebounds originated from the tendon structure, the sole material energy rebound, and the muscle [54].

In contrast, there was no energy return at the MTPJ was found by Stefanyshy et al. [63]. In other words, no increase in positive work of the MTPJ was found, possibly due to differences in control conditions, such as different materials of the shoe, toe spring, and so on. However, it has also been found that changes in LBS could also had an effect on the hip, knee, and ankle joints. Several kinds of research indicated that increasing the LBS of running footwear may significantly reduce energy loss at the MTPJ [55, 66-68]. The MTPJ might cause the energy loss changes because of an increased LBS and increased peak plantarflexion moment [52, 67, 69].

Metatarsal stress fractures were microfractures of the metatarsal bone due to periodic submaximal loading. However, the most common mechanism of injury was associated with a sudden increase in training volume in athletes, the LBS of footwear. From the points of indirect studies, the LBS of the shoes may influence the occurrence of metatarsal stress fractures by modulating the peak pressure acting on various regions of the foot [77] to reduce the chance of metatarsal stress fractures. It may also play a role in these types of injuries by directly changing the position of the load acting on the foot or by affecting the muscle tissue surrounding the foot. The midsole of race running shoes inserted carbon fiber plate increased the LBS of shoes and affected the strain force at the MTPJ. There was evidence that fatigue of the thumb muscles increased the pressure on the second metatarsal, which may lead to an increased risk of stress fractures [64].

These findings suggested that increasing the LBS of shoes induced some lower limb changes biomechanical, particularly in the MTPJ, enhancing running performance while reducing the risk of injury to the MTPJ.

2 Optimize the carbon fiber plate design without notably alter the LBS of shoe

2.1 Introduction

Most studies had changed the LBS of running shoes by adding carbon fiber plates or adding other stiff materials to the midsole to improve LBS [56]. It was notable that the track shoes inserted with carbon fiber plates, such as the Nike Vaporfly 4% (VF) shoe combined both advances in midsole thickness and LBS to reduce energy loss by about 4% for runners, which was contributed to improving running performance [53, 70]. The full-length embedded carbon fiber plate to the midsole would increase the LBS of the shoe [71, 72], reducing the running economy by about 1% [73]. The VF made runners trend more to midfoot or forefoot strike and has high requirements for the runner's muscle strength because of its high rearfoot thickness and the strong propulsion structure of the forefoot [74]. The previous research on the foot strike patterns demonstrated that the rearfoot strike pattern was mainly used among the prolonged runners in road races, with percentages ranging from 74.9% of runners in a professional half-marathon race, to over 90% of amateur runners in marathon distance events [14, 74-76].

The Xtep innovation R&D center thus created a pair of racing shoes that reduced the thickness of the midsole but retained the curved carbon fiber plate to meet the needs of marathon runners of different levels. According to the pilot work from the Xtep lab, marathon runners felt too hard on the forefoot area if they continued to run after 30 km when wearing running shoes with a full carbon fiber plate. As mentioned above, the carbon plate embedded in the midsole would increase the LBS of track running shoes while also increasing the overall hardness of the sole, which transferred the center of pressure under the foot forward [54]. In addition, the marathon was a long-distance repetitive sport that would lead to foot muscle fatigue, and fatigue of the thumb muscles of the foot increases the pressure on the second metatarsal bone, which may lead to an increased risk of stress fracture [63].

In summary, it was valuable to pay attention to the plantar pressure of MTPJ without changing the performance of running when designing a pair of running shoes inserted carbon fiber plate. It was noted that the effects of the forefoot construction of the carbon fiber plate had not been investigated, and it could likely be very hard to investigate the pressure distribution on the plate or midsole through human trials [85]. So, manufacturers need to do further research about the construction of the forefoot plate such as adjusting the full forefoot plate construction (FFC) to segmented forefoot plate construction (SFC). In recent years, finite element (FE) methods had been commonly applied in biomechanical research of the lower extremity due to their ability to process the complex geometry structures for both static and dynamic analysis [77, 78]. This study aimed to research the effect of the construction of the forefoot plate combined with the running biomechanics and FE simulation. Based on previous literature, it was hypothesized that (1) the SFC model has a lower LBS than the FFC model, which would increase the angle of MTPJ dorsiflexion and potentially increase the amount of energy loss at the MTPJ. (2) the SFC model has a lower maximum pressure on the forefoot area than the FFC model.

2.2 Methods

2.2.1 Finite Element Simulation

In this study, the outsole, midsole, and two kinds of carbon fiber plate had been modeled in 3D based on an industrial 2D shoe design drawing with Rhino 6 Computer-Aided Design (CAD) software (Robert McNeel & Assoc, Seattle, WA, USA). The carbon plate built-in midsole divides the midsole into two layers. The thickness of the upper midsole of the forefoot accounts for about 60%, and the thickness of the upper midsole of the heel accounts for about 36%. A meshing of the shoe has been done with ABAQUS software by this CAD model (Dassault Systemes Simulia Corp, Johnston, RI, USA) that the discretization was 2.7 mm **(Figure 2)**. All of the solid parts were assembled into a whole sole model, then imported into the FE package ABAQUS (Dassault Systemes Simulia Corp, Johnston, RI, USA) to develop the numerical model. To simulate the flexion mechanical test, the sole model was first positioned on two rigid plates, corresponding to the virtual flexing machine: fixed and flexing one. The sole was camped to the fixed plate by applying a 900 N to toe clamp at a 70%-foot length (heel to toe) while the heel end was on the flexion plate (**Figure 3**), and the flex angle was 45 degrees. The coefficient of friction between the sole and plates was 0.6. the coefficient is from some paper and simulation results have good correlation with experimental results [181]. In this study, the midsole was made of Polyetherblockamide foam (Pebax®, UBESTA, Yubu Xingchan Co., Ltd, Yubu, Japan). thus, to determine mechanical properties for finite element analysis, this material was tested at quasi-static rates by using a universal material test machine. Compression tests were performed according to the ASTM-575 standard using cylinder specimens (diameter: 28.6 mm. thickness: 12.5 mm) at a 10 mm/min speed. The specimen density was 0.12 g/cm^3. Force-displacement data were obtained from the quasistatic tests and converted to stress-strain data using the sample dimensions. The Ogden hyper foam material model was chosen to represent the non-linear response of the Pebax® foam obtained from the experiments. This model describes a compressible and nonlinearly elastic behavior. The hyper foam material constants for Pebax® foam were μ = 0.28, α = 5.177, Poisson's ratio = 0.125. To determine the mechanical properties of the reinforced carbon fiber plate, a three-point bending test was carried out using the material testing machine with a speed of 1 mm/min. The test samples were prepared according to ASTM-D790: 1 mm thickness, 18 mm width, and 80 mm length strips were cut from an original plate with the help of an electrical power saw. The specimen density was 1.1 g/cm^3. Young's modulus (E) was obtained from the mechanical test = 33,000 MPa and Poisson's ratio = 0.4. The sole, made of foam, was discretized using tetrahedral elements with an average size equal to 2.7 mm. The carbon fiber plate was also discretized, using tetrahedral elements with an average size equal to 1 mm. A convergence study has been performed to confirm if the mesh density was acceptable. Finally, the simulation was performed in Abaqus using the Dynamic Explicit solver. Peak torque (Nm), stiffness (Nm/deg) and energy return (%), contact pressure on the plates (MPa) were calculated.

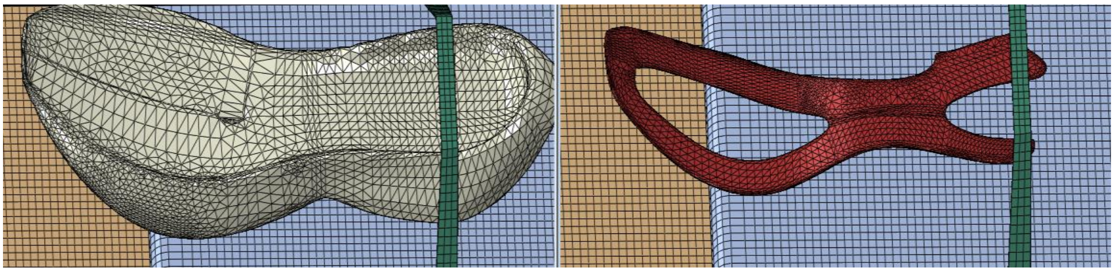


Figure 2. Meshing of the shoe and the forefoot plate.

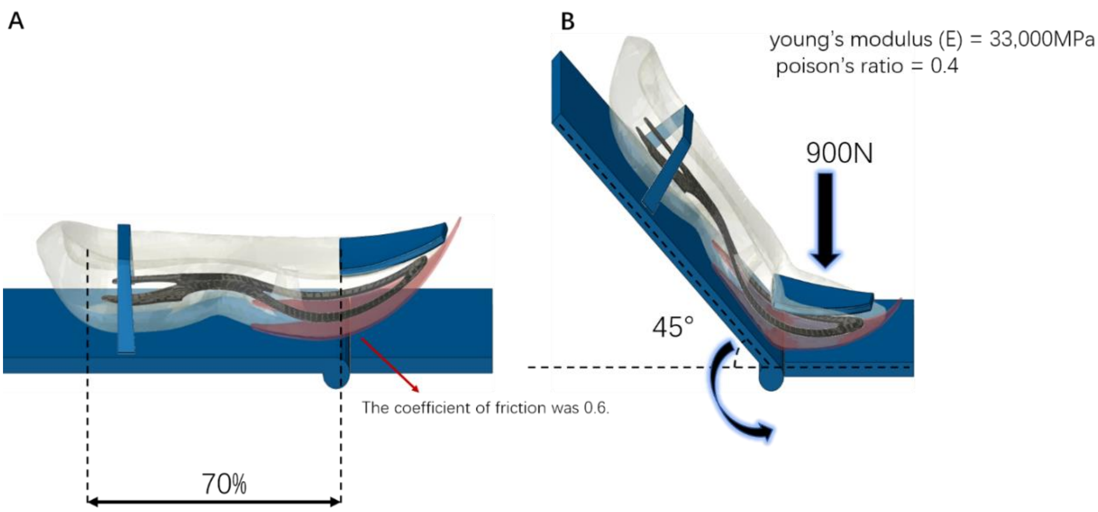


Figure 3. Finite element model: (A) Initial position (B) Conditions imposed

2.2.2 Biomechanical Data Collection

Participants performed eight valid right foot rearfoot strike running trials per testing shoe on a 145 m concrete indoor running loop (**Figure 4**). We hoped that this pair of shoes could be suitable for different types of runners, not only for professionals but also for amateur runners, so we chose a fast and slow speed for all tests. A valid trial was within the specified velocity range (fast speed: 4.81 ± 0.32 m/s, slow speed: 3.97 ± 0.19 m/s) and made up of the whole right foot contacting the force plate area. Before, data collection participants warmed up for about five minutes and were acquainted with the target speed and shoe conditions by running 2 laps in each shoe condition. Upon failing to match the required speed in the first two laps, further familiarization laps were performed as necessary. For GRF and 3D kinematic measurements, participants ran across a set up of three consecutive and flush into the floor force plates (combined dimensions 270 × 60 cm, 1000 Hz (AMTI, Watertown, MA, USA)) in each shoe condition (**Figure 5**). The test sequence of shoes was randomized for each participant. The two-timing gates 8 m far from the middle force plate were used to record the running speed (Smart speed, Burbank, CA, USA) set 8 m apart, centering the middle force plate. Right leg kinematics were collected at 250 Hz and were collected using a 10-camera motion analysis system in a capture volume of 4.0 × 1.0 × 1.5 m (Vantage 5, Vicon, Metrics Ltd., Oxford, UK). The marker set was according to the calibrated anatomical systems technique [79]. The right thigh, the right shank, the right foot (forefoot and rearfoot) were defined as segments by attaching retro-reflective markers of fourteen millimeters in diameter on the skin of the right and left anterior superior iliac spine (ASIS), the right and the left posterior superior iliac spine (PSIS), the right greater trochanter, the medial and lateral epicondyle of the femur, the medial and lateral malleolus, as well as attached to the shoe, representing the first and fifth metatarsal heads and second toe. Three marker tracking clusters were attached to the lateral side of the thigh and the lateral side of the lower leg [80]. The extra reflective markers were added to the distal, proximal heel, and lateral rearfoot, respectively, and were defined as shoe-mounted tracking markers [81] (**Figure 6**). A static trial was conducted before data collection. All study procedures about biomechanical data collection were similar to the published paper [82], both performed in the Xtep science lab. In the trial, valid data could be used when the first impact peak and shoe ground angle more than zero appeared. We used the Vicon Nexus 2.7 and Visual3D (C-Motion, Germantown, MD, USA) to process the collected experimental data. A fourth-order low pass Butterworth filter was used with a cut-off frequency of 100

Hz (kinetic) and 10 Hz (kinematic) [82]. The XYZ Cardan sequence was used to calculate lower limbs' kinematic and kinetic data, in which X represents flexion-extension, Y represents abduction–adduction, and Z represents internal-external rotation [83]. The angle, the angular velocity, the GRF, and the work of the hip, the knee, the ankle, and the MTPJ of the right lower limb were measured during the stance phase using Visual3D (C-Motion, Germantown, MD, USA).

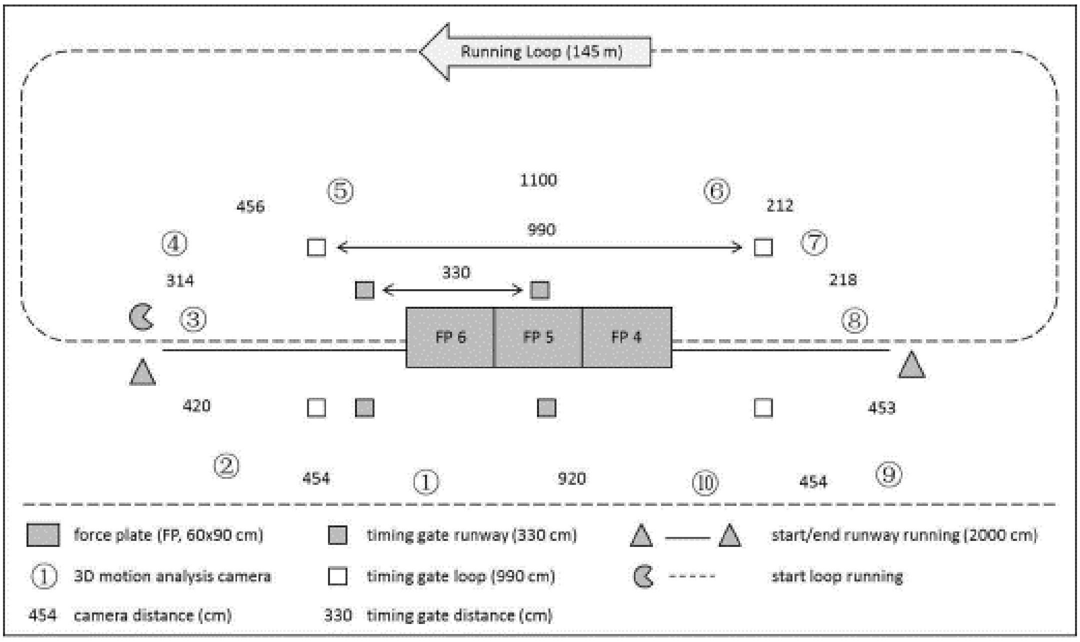


Figure 4. Laboratory space for runway running and loop running data collection methodologies.

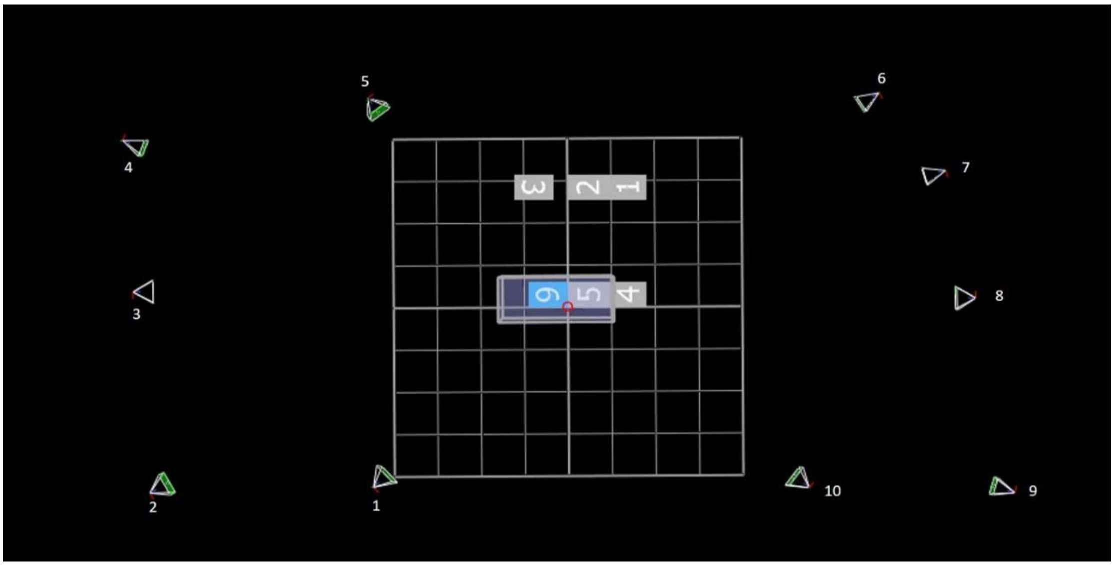


Figure 5. Laboratory space, force plate, camera position, and the red circle indicate capture volume origin for runway running.

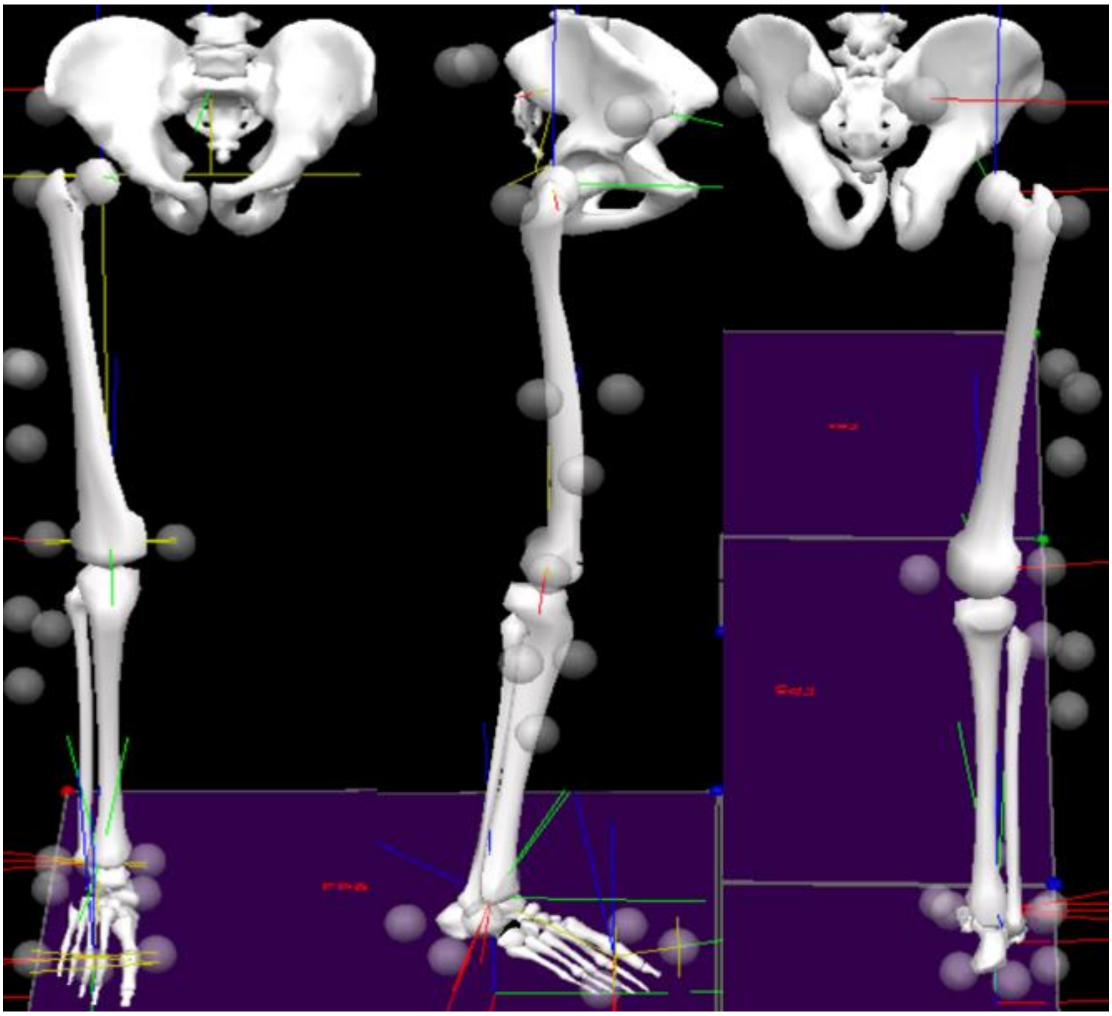


Figure 6. Right leg lower extremity marker model

Figure 7. Brannock Device

2.2.3 Participants and Experiment footwear

Fifteen male runners (mean (SD) age: 34.93 (10.25) years, height: 1.70 (0.05) m, weight: 61.47 (45.59) kg, BMI: 21.22 (1.77) kg/m2) joined in this research. All participants were recruited from the Xiamen running club and identified themselves as rearfoot strike runners. Participants were free from injury for at least six months before this study. All participants had been confirmed in foot size (EU 41 ± 0.5) by the Brannock Device (The Brannock Device Co., Syracuse, NY, USA) (**Figure 7**) before the official test.

There were two kinds of experimental footwear used in this research: (rearfoot height: 26 mm, forefoot height: 18 mm, offset: 8 mm, rearfoot width: 76.5 mm, forefoot width: 102 mm, midsole material: foam in hardness of 50 asker C, outsole material: rubber in hardness: 62 asker A, differing in their construction of carbon fiber plate (SFC: 1 mm thick carbon fiber plate with segmented forefoot plate construction, FFC: 1 mm thick carbon fiber plate with complete forefoot plate construction) inserted in the midsole. The inner length of the carbon plate is 223mm, the outer length is 219mm, the widest spacing of the forefoot hole is 74mm, the maximum distance between the inner and outer sides of the carbon plate of the forefoot is 19mm and 17mm respectively, the radian of the bottom of the middle waist of the carbon plate is 23.1 °, and the radian of the bottom of the forefoot is 34.6 °. The dimension data of the broken forefoot carbon plate is 16.3mm wide and 6.5mm length (**Figure 8**).

Mechanical flexion measurements fixed the forefoot area in the location of 70%-foot length (heel to toe), then bending at 45 degrees was performed by applying a dynamic shoe flexor device (Brentwood, NH, USA) **(Figure 9)** to measure the shoe LBS and energy return [84], last, measuring the force on the forefoot area by a pressure sensor.

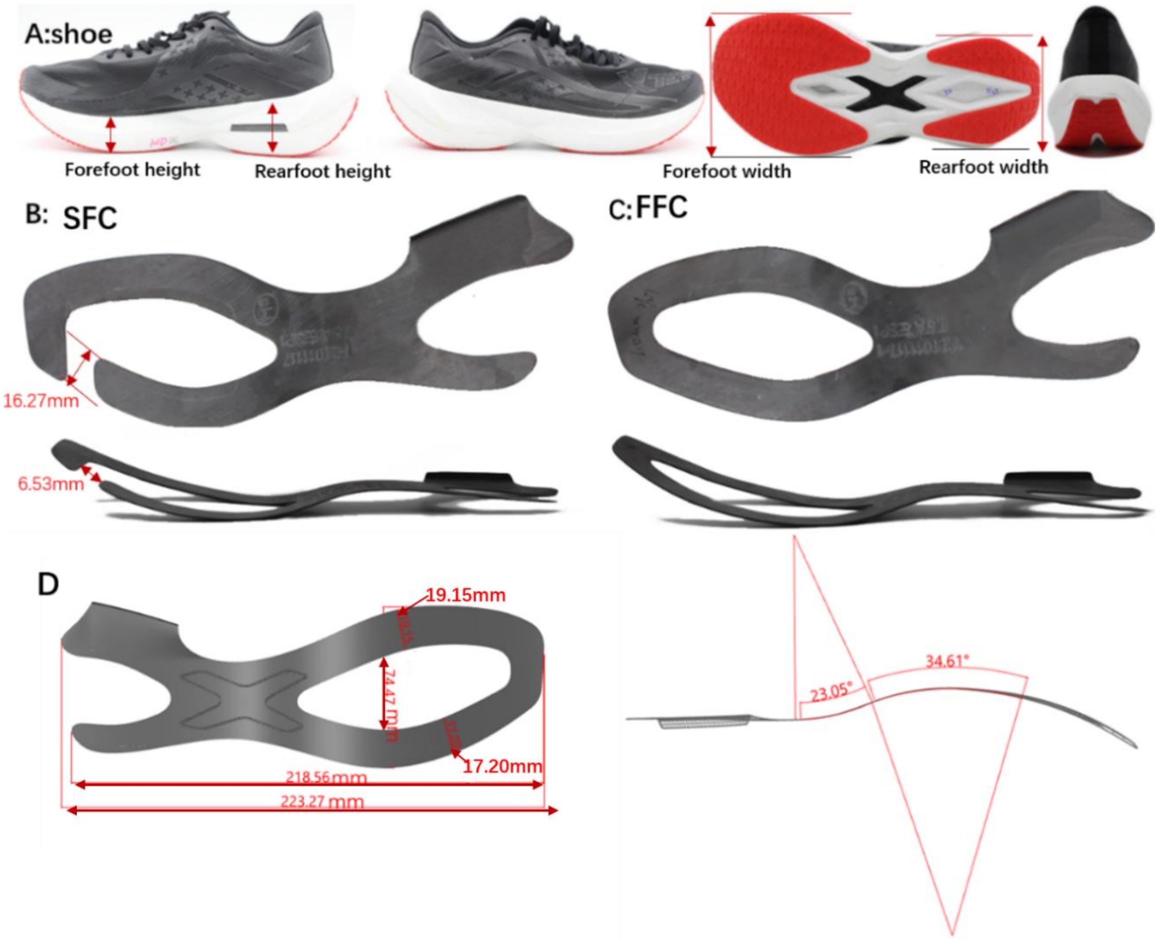


Figure 8. (**A**): Experiment shoe (Forefoot height: vertical thickness at 12% of external length, Rearfoot height: vertical thickness at 75% of external length, offset: offset = rearfoot height − forefoot height). (**B**): the forefoot area of carbon fiber plate (carbon fiber plate was made up of 63 % carbon fiber and 37% epoxy resin fiber) was designed to a segment construction inserted to midsole (SFC), (**C**): the forefoot area of carbon fiber plate was designed to a full construction inserted to midsole (FFC)). (**D**): the information about the geometry and dimensions of the carbon plates.

Figure 9. dynamic shoe flexor device

2.2.4 Statistical analysis

Statistics were processed by Statistical Product and Service Solutions (SPSS) (24, IBM Corp., Armonk, NY, USA). Shapiro–Wilk tests were adopted in this study. A 2 $\times$2 (factors: running speed, running shoes) within-subjects factorial repeated-measures analysis of variance (RM ANOVA) was selected to evaluate the main effects and the interaction of these factors on the biomechanical variables. Statistical Alpha levels were set to 0.05. The alpha levels were adjusted to < 0.003 according to post-hoc pairwise comparisons with a Bonferroni correction when variables showed a significant main or interaction effect. Partial eta squared estimates (ηp^2) were calculated for statistically significant variables.

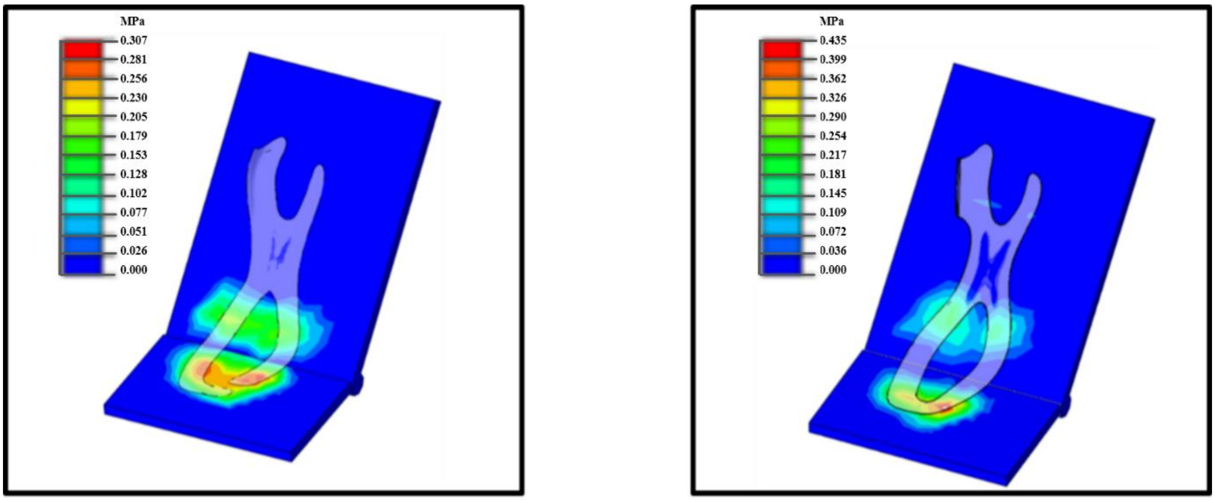
A Statistical Parametric Mapping (SPM) technique [85, 86] assessed the main effects of ‘running shoes’ and ‘running speed ‘factors and their interaction, and SPM tests were calculated with the SPM1D v0.4 for MATLAB (www.spm1d.org (accessed on 03.01.2021), [85]). The significance level was set at 0.05 for all statistical tests.

2.3 Results and Discussion

The FE simulation showed that the maximum pressure on the forefoot of the SFC (0.307 MPa) was lower than that on the FFC (0.435 MPa) (**Figure 10**), the maximum strain on the forefoot of the SFC (0.440) was lower than on the FFC (0.560) (**Figure 11**). Still, the results from the forefoot flexion scores and bending simulation indicated no effects between SFC and FFC in LBS and energy return (**Table 2.1**).

Table 2.1. The forefoot flexion scores for experimental shoes with the same insole and the FE simulation variables for experimental carbon fiber plate.

Measurement Method	Variable	Experimental Shoe Condition	
		SFC	FFC
Weight(g)		184.17	187.92
Forefoot flexion	Peak torque (Nm)	16.50	15.44
	Stiffness (Nm/deg)	0.370	0.369
	Energy return (%)	33.97	34.66
FE simulation	Peak torque (Nm)	13.54	13.68
	Stiffness (Nm/deg)	0.301	0.304
	Energy return (%)	64.08	64.82



（a-SFC）　　　　（b-FFC）

Figure 10. The pressure on the SFC and FFC during bending.

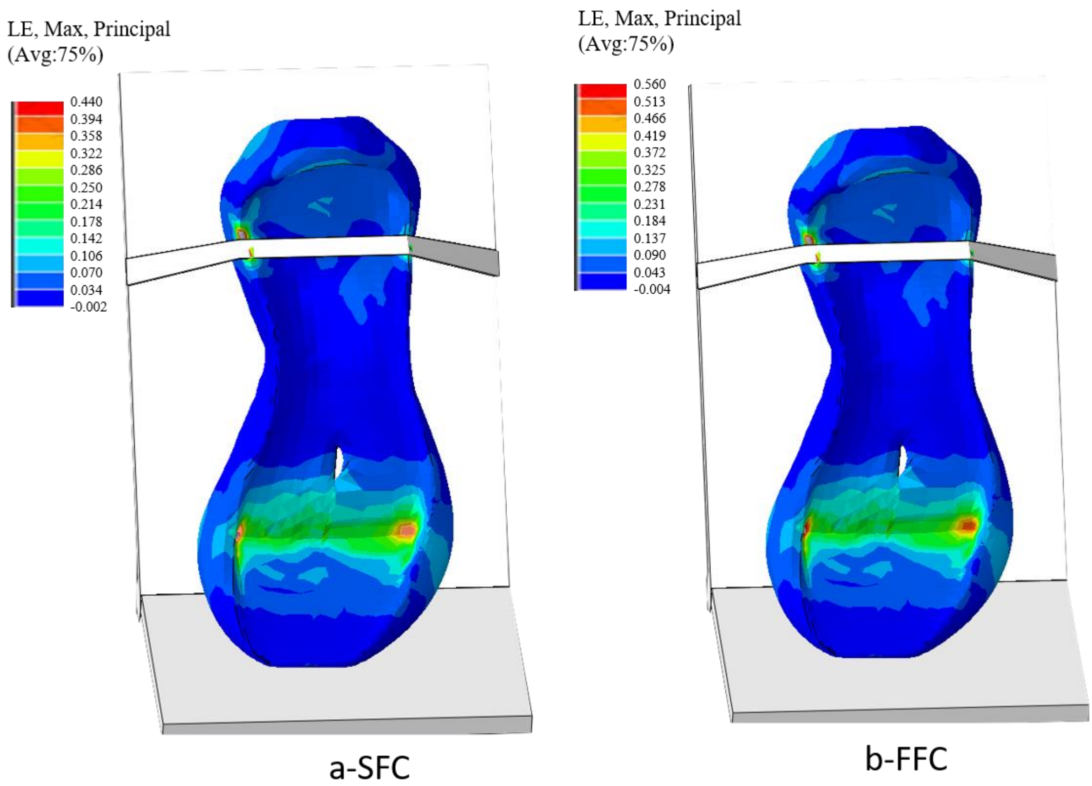


Figure 11. The strain on the SFC and FFC during bending.

The result showed that the vertical and anteroposterior GRF (**Figure 12a**), ankle, knee, and hip range of motion (**Figure 13**), the moment at each lower limb joint (**Figure 14**), MTPJ, and shoe slap velocity (**Figure 12b**), positive and negative work at each lower limb joint (**Figure 15**) of faster speed (4.81± 0.32 m/s), were bigger than with the slower speed (3.97 ± 0.19 m/s) in both experimental shoes ($p < 0.05$).

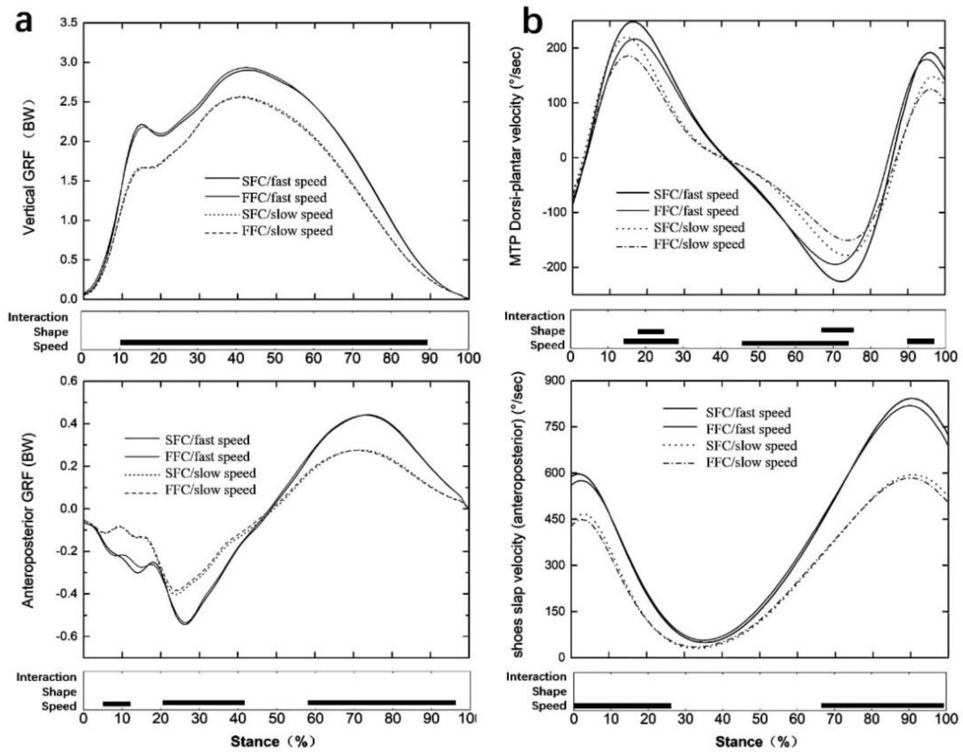


Figure 12. Vertical GRF time (**a**) and anteroposterior GRF time (**b**) (weight−normalized) (shoe slap velocity showed the shoe ground velocity).

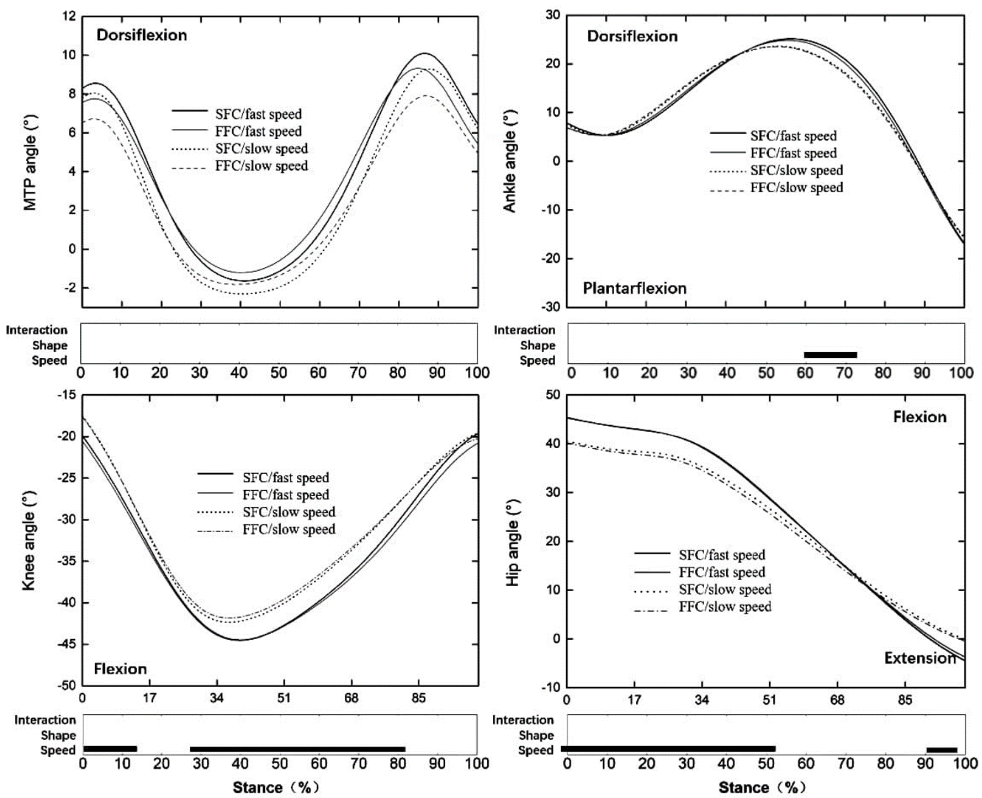


Figure 13. Lower limb joint angles time−normalized.

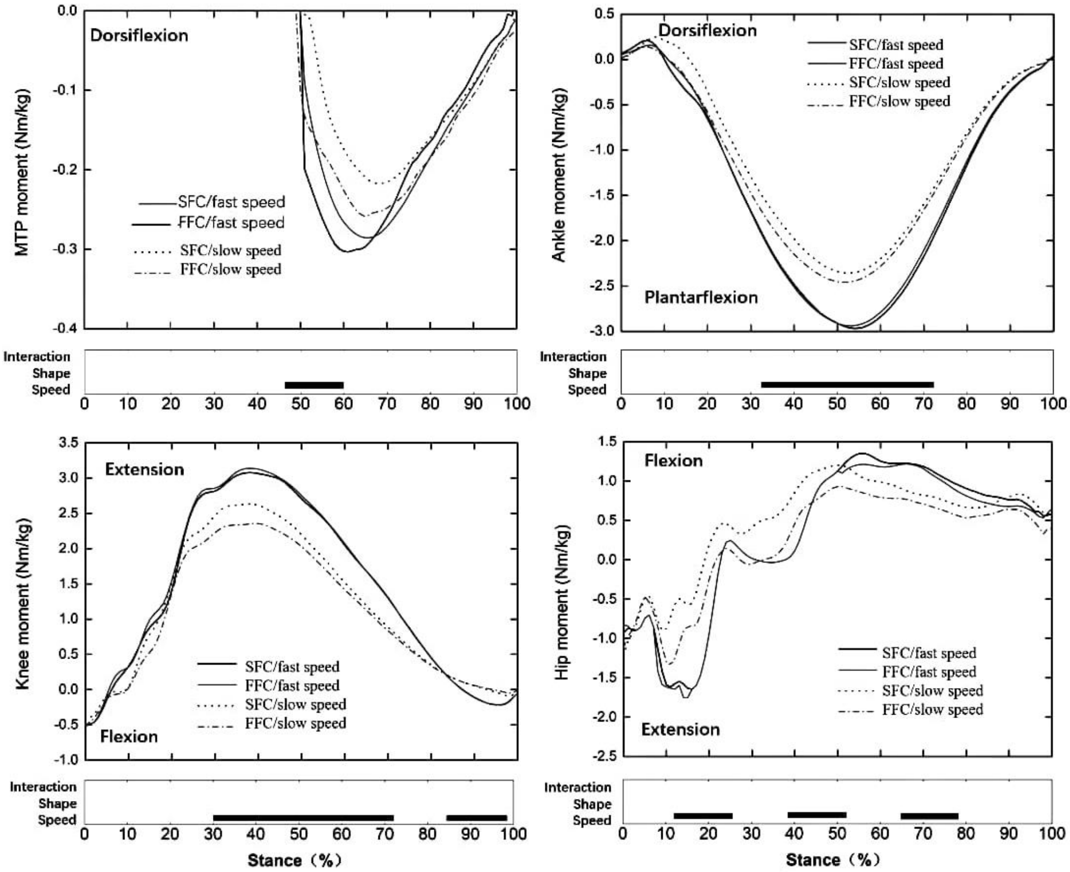


Figure14. Lower limb joint moment time and weight−normalized. Note: The significant main effects of the interaction, the location, and the speed were highlighted (black horizontal bars at the bottom of the figure) during the stance phase of running.

As for the effect of the construction of the carbon fiber plate, the SFC induced more MTPJ Dorsi-plantar velocity from 18% to 26% ($p < 0.05$) and 67% to 78% of the stance phase ($p < 0.05$) compared to the FFC (**Figure 12b**). The positive joint work at the knee joint ($p = 0.038$, $\eta p^2 = 0.178$) was larger for the SFC compared to the FFC (**Figure 15**). There were no significant differences between SFC and FFC at ground contact time, breaking phase time, and some variables for MTPJ, such as MTPJ negative work, MTPJ Dorsi-plantar range of motion, and so on (**Table 2.2**). In addition, there was no effect from the interaction between the construction of the carbon fiber plate and the running speed.

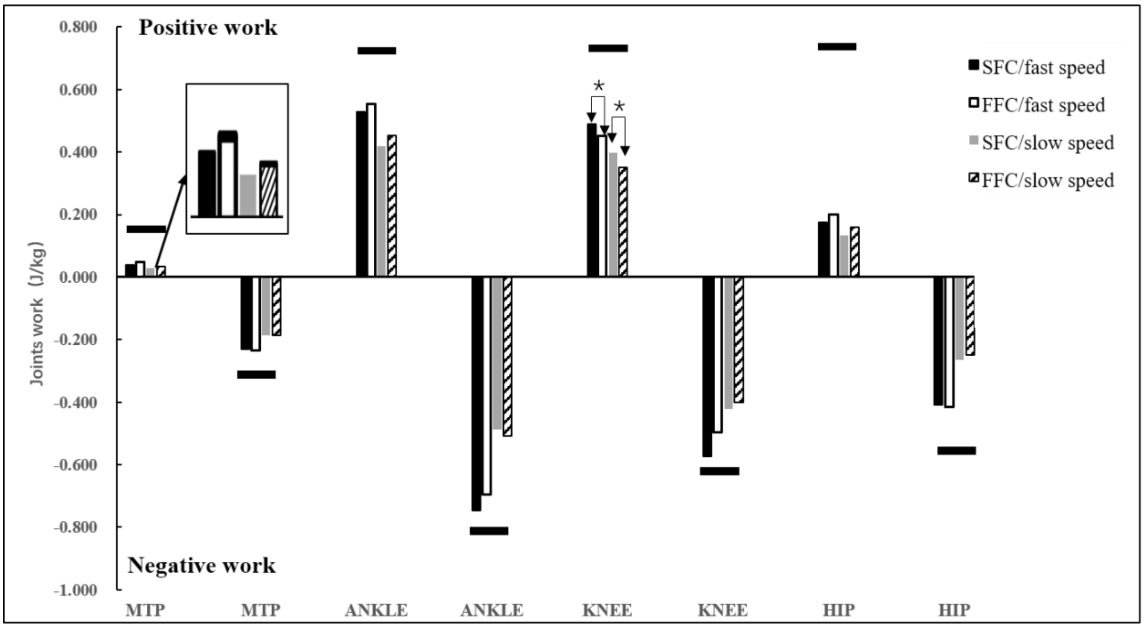


Figure 15. Joints work and showed significant main effects of the interaction. The black horizontal bars showed significant main effects of speed, and the * showed significant main effects of the construction of the carbon fiber plate.

Table 2.2 The biomechanical variables for MTPJ and spatiotemporal parameters between SFC and FFC.

Variables	SFC(F)	SFC(S)	FFC(F)	FFC(S)	Main effect (construction) P	Main effect (speed) P	Interaction effect P	Effect size (ηp^2) F	effect size (ηp^2) S
ground contact time (ms)	185.13±20.13	231.42±11.44	184.69±22.80	227.63±16.48	0.636	0.000	0.147	0.034	0.094
breaking phase time (ms)	92.552±0.01	117.88±0.00	91.495±0.020	113.696±0.01	0.964	0.009	0.557	0.234	0.121
propulsion phase time (ms)	92.583±1.547	114.532±2.30	93.196±2.182	115.012±2.30	0.487	0.004	0.219	0.052	0.059
MTPJ plantarflexion velocity (sagittal) (°/sec)	255.33±29.04	201.55±32.69	212.60±39.14	171.64±34.26	0.015	0.001	0.216	0.341	0.287
MTPJ negative work (J/kg)	0.052±0.006	0.049±0.008	0.042±0.007	0.040±0.01	0.123	0.553	0.339	0.176	0.073
MTPJ dorsiflexion angle at toe-off (°)	7.29±2.54	6.81±2.16	5.76±2.14	5.35±2.09	0.478	0.248	0.363	0.173	0.131
MTPJ range of motion(sagittal) (°)	13.46±1.50	12.82±2.04	12.09±1.84	11.24±2.27	0.135	0.092	0.107	0.254	0.191
MTPJ dorsiflexion angle (sagittal) at pp (°)	7.63±3.03	4.61±2.69	6.81±2.80	3.88±2.33	0.453	0.002	0.410	0.179	0.142
MTPJ dorsiflexion angle (sagittal) maximum (°)	11.81±3.11	10.92±2.89	10.85±2.70	9.59±2.13	p = 0.297	p = 0.296	p = 0.123	0.116	0.114

Discussion

This research aimed to investigate the effect of the construction of a carbon fiber plate. In contrast to our first hypothesis, differences in the construction of the carbon fiber plate did local effects were observed in shoe LBS during bending (**Tables 2.1 and 2.2**). This was inconsistent with the previous result which has shown that cutting the carbon fiber plate would reduce the shoe LBS due to the mechanical behavior-changing of the shoe midsole [87]. What's more, the results of mechanical and finite element analysis showed that changing the construction of the carbon fiber plate on the forefoot area could also not affect the energy return. This could be due to the adjustment has noticeable difference in its local environment. These were some differences between experimental data and simulation results, even though the model could display realistic trends in general, it overestimated the energy return of the midsole measured in the flexion machine, and the reasons causing the overestimation were discussed below. In the analysis, the material models applied were a simplistic representation of the complex behavior of each material in response to loading. For example, the modeled carbon plate did not have defined viscoelastic properties and thus no means for energy dissipation. In the analysis, the representativeness of the loading conditions used also has a degree of uncertainty. Firstly, the energy return was calculated for a single trial, and the model output was compared to average values measured across 65 trials. The accuracy of the computed results was also dependent on the force applied to the toe clamp to hold the sole during flexion. During flexion test motion of the toe, the clamp was observed, while for the simulation, when flexion was applied, fixation was fixed. The interaction between the footwear parts and the footwear flexion machine was another area that contains several significant simplifications. The friction coefficient used in the analysis for the whole model was taken to be equal to 0.6. While not investigated, it was hypothesized that improved results could also be achieved by using a different friction coefficient between other parts instead of the same coefficient used in the current methodology. Finally, the exclusion of the outsole, insole, and upper, from the FE model, could have resulted in an overestimated energy return of the sole. Even though the methodology applied has reported limitations, there were similarities between the results predicted with the analysis and those measured from flexion tests, with similar trends in the peak torque of the soles observed. While not perfect, the model was still considered valuable as a comparative tool, to evaluate the peak torque, energy return, sole stiffness, stresses, and strains that might occur in future footwear designs. Nevertheless, changing the construction of the carbon fiber plate derivatized several biomechanical changes during running. For example, SFC increased the peak of the MTPJ plantarflexion velocity and the positive work at the knee compared to FFC.

From biomechanical perspectives, the MTPJ joint was a possible target area for the application of improving running performance. There was an increased peak of MTPJ plantarflexion velocity with the SFC compared to FFC (**Figure 12b**). The major factor was that the shape of the forefoot carbon fiber plate was changed. In addition, some studies had shown that carbon fiber plates had the ability to store and return elastic energy, which might work as a torsional spring as the MTPJ joint underwent rotational deformation during the ground contact in the running. Running is a rolling process with a certain internal rotation state when toe off. Cutting the carbon fiber plate would change the location of the forefoot carbon fiber plate which affect this function of the torsional spring. Cutting the carbon fiber plate would weaken this function of the torsional spring which was presented by more positive work performed at the MTPJ [54, 67, 88-90]. There was a redistribution of positive lower limb joint work from the knee to the MTPJ, increasing the midsole bending stiffness [90]. There was more positive work at the knee joint with the SFC compared to FFC in this study. The main factor was that the midsole bending stiffness deformation of experimental shoes was insufficient to lead to the work redistribution on the lower limb joint.

In line with our second hypothesis, the maximum pressure on the forefoot area of the plate was lower with the SFC compared to the FFC during the bending simulation (**Figure 10**). In other words, it would reduce the maximum pressure by about 29.4% on the midsole each step when adjusting the FFC to SFC. This suggests that it was of importance to take the construction of the carbon fiber plate into account when footwear manufacturers plan to design a marathon shoe because the racing shoes embedded in the carbon fiber plate would bend probably between 30,000 and 40,000 times during a prolonged run such as a full marathon.

Two kinds of running speed (fast speed: 4.81 ± 0.32 m/s, slow speed: 3.97 ± 0.19 m/s) induced those significant changes in this study which were in line with those previously observed [131]. The results showed that the fast speed significantly increased vertical and propulsive GRF, increased ankle, knee, and hip joint range of motion, and increased moment and work in all lower limb joints, giving more MTPJ angular velocity and shoe slap velocity compared to the slow speed. There was no effect of the interaction between the construction of the carbon fiber plate and the running speed (**Figures 12–14**).

The study showed that adjusting the full forefoot plate construction to segmented forefoot plate construction induced some biomechanics changes, such as more MTPJ plantarflexion angular velocity and more positive work at the knee joint. Still, it did not affect the work at the MTPJ. In addition, the results in finite element simulation provided practical evidence for footwear manufacturers that could be beneficial from a long-distance running perspective by reducing the maximum pressure on the midsole without significantly affecting the LBS.

2.4 Conclusion

In this paper, to investigate the independent effect of the construction of the forefoot carbon-fiber plate inserted into the midsole on running biomechanics and finite element simulation, fifteen male marathon runners were arranged to run across a runway with embedded force plates at two specific running speeds (fast-speed: 4.81 ± 0.32 m/s, slow-speed: 3.97 ± 0.19 m/s) with two different experimental shoes (a segmented forefoot plate construction (SFC), and a full forefoot plate construction (FFC)), simulating the different pressure distributions, energy return, and the LBS in the forefoot region between the SFC and FFC inserted to midsole. Kinetics and joint mechanics were analyzed. The results showed that the footwear with SFC significantly increased the peak MTPJ plantarflexion velocity and positive work at the knee joint compared to the footwear with FFC. The results of finite element simulation showed a reduced maximum pressure on the midsole, meanwhile, not significantly affected was the LBS and energy return with the SFC compared to the FFC. The results could be used to design marathon running shoes because changing the full carbon fiber plate to a segment carbon fiber plate induced some biomechanical transformation but did not significantly affect the running performance. What was more, reducing the peak pressure of the carbon plate to the midsole by cutting the forefoot area of the carbon fiber plate could be beneficial from a long-distance running perspective for manufacturers.

3 Lower limb mechanics during moderate high-heel jogging and running in different experienced wearers

3.1 Introduction

Most studies on improving sports performance by increasing the LBS of carbon running shoes had focused on male runners. Still, there were apparent differences between male and female runners in terms of physiological characteristics and biomechanical performance during running. Thus, it was necessary to understand their physiological and biomechanical factors and consider their wearing habits before studying running shoes for female runners. The high-heel design has been remaining one of the dominant features of women's footwear. Social and fashion customs encourage the continued use of high-heel shoes [91]. A retired professional social dancer (Tennesseeran) finished the entire marathon wearing high heels to set a world record. Despite detrimental effects on the musculoskeletal system, such as lower back pain, ankle sprains, foot pain, hallux eversion, and increased predisposition toward degenerative knee osteoarthritis [92, 93].

From the points of biomechanical, forcing the ankle to a plantar-flexed state, high-heel shoes with narrow supporting bases led to a series of kinematic and kinetic changes in lower limbs during walking or running. Changes in spatiotemporal parameters had been well documented. Unanimous results revealed that an increase in heel height contributed to slower self-selected walking speeds and shorter strides with generally unchanged cadence [94-96]. Most studies on high-heeled gait concern changes in the knee joint. Compared with barefoot or flat shoe conditions, walking in high heels has been believed to increase knee flexion during the stance phase and at heel strike as compensatory mechanisms attenuating GRF [97, 98]. According to the notion that a larger knee abduction moment was associated with the development of knee osteoarthritis, the increase of peak knee abduction moment as the result of increasing heel height had been highlighted [96, 98, 99]. Studies concerning alternations in ankle joint during high-heeled walking concluded that the risk of lateral ankle sprain would increase as heel height increased with increasing plantarflexion and inversion

[100, 101]. Barkema et al. [96] found that systematically increased peak ankle eversion moment during the late stance phase with increased heel height. As to the hip joint, increased flexion and abduction moment were verified as assistants to attenuate GRF during the first half of the stance phase and compensate for the reduced ankle plantarflexion moment during push-off [95, 98]. Ebbeling et al. [102] reported that the first and second maximal vertical GRF increased as heel height increased during walking at a fixed speed. Research by Stefanyshyn et al. [100] obtained similar conclusions from habitual subjects. This study also indicated that the threshold for the impact force and maximal vertical loading rate increased as heel height increased [100]. It was also reported that when heel height increased from 7.6 cm to 8.5 cm, both impact force and loading rate decreased, which would be an injury prevention strategy employed by high-heel wearers [103]. In general, kinematic and kinetic alternations of the lower limb in the studies mentioned above were responses to high heels at the walking level. In contrast, studies instructing subjects to walk at a fixed speed showed no significant differences in spatiotemporal parameters, joint angles, and GRF between experienced and inexperienced high-heeled wearers [98, 102]. One reason for significant differences may be that the fixed speeds were greater than the preferred speeds in high-heeled walking. A previous study demonstrated that the maximal vertical GRF increased linearly from 1.2 body weight (BW) to approximately 2.5 BW during walking and running, respectively [104]. The joint motion of the lower limb during walking also significantly differs from that during jogging and running.

There were few studies concerning the effects of high-heeled jogging and running on lower limb mechanics. Gu, Sun, Li, Graham, and Ren [105] noted that the motion range of knee abduction-adduction and hip flexion-extension increased significantly as heel height increased during jogging which may induce high loading force in knee joints. All these studies recruited habitual high-heeled wearers as subjects. Accordingly, whether these changes were immediate or chronic adaptations had not been fully explored. Traditional wearers of high-heel shoes were reported to experience long-term adaptations in muscle-tendon architecture, such as increased Achilles tendon stiffness and tendon hypertrophy [106], shortening of gastrocnemius medialis fascicles [94]. These adaptations could shift the stretch distribution of muscle-tendon units away from tendinous tissues toward muscle fascicles during walking, potentially altering neural activation patterns and decreasing efficiency [107]. It was noted that biomechanical adaptations varied between experienced and inexperienced high-heel wearers, including increased knee flexion and exaggerated upper trunk rotations of the latter during the stance phase when walking at a preferred speed [108]. Other differences such as increased abductor and reduced internal rotator moment at the hip, reduced ranges of flexion-extension and adduction-abduction at the knee, increased external rotation, pronator, and external rotator moments at the ankle were also observed in the self-selected walking task wearing 7.3 cm high-heel shoes [92]. However, whether these changes in inexperienced moderate high-heel wearers were the same or even worse and whether the increased speed of running has extended effects remain unclear.

It was worth noting that, the number of participants in the half marathon has increased from 300,000 in 1990 to nearly 2 million in 2013, and over 60% of them were female runners [109]. Among female runners divided into experienced (EW) and inexperienced (IEW) high-heel wearers, biomechanical differences between these two types of runners in jogging and running in high heels were investigated to provide a reference when designing running shoes for different kinds of women based on the habit of wearing in the future.

The purpose of this study was to clarify differences in lower limb kinematics and GRF between EW and IEW during moderate high-heel jogging and running. It was hypothesized that EW would show faster self-select speeds of jogging and running than IEW. EW would decrease joint instability while increasing GRF compared to IEW. Changes of lower limb joints (range of motions and peak angles) and GRF would increase as speed increased for all wearers.

3.2 Methods

3.2.1 Experiment protocol and procedure

A force platform (Kistler, Switzerland) was fixed in the middle of the walkway and utilized to collect GRF at the frequency of 1000 Hz. The 8-camera Vicon motion analysis system

(Oxford Metrics Ltd., Oxford, UK) was used to capture lower limb kinematics at the frequency of 200 Hz. Subjects completed jogging and running tasks separately at self-selected speed along a 10-meter-walkway. The experimental shoe with 4.5 cm height heels (**Figure 16**) was commercially available.

Subjects were given enough time to familiarize themselves with the experimental environment and adjust their gait to ensure the right foot stepped onto the force plate completely and naturally before data collection. IEW also performed some progressive training, learning to jog at progressively increasing speeds at which they feel comfortable and safe. Subjects were required to wear tight-fitting pants and 16 reflective markers (diameter: 14 mm) were attached with adhesive on the left and right lower limbs, respectively. The marker locations included: anterior-superior iliac spine, posterior-superior iliac spine, lateral mid-thigh, lateral knee, lateral mid-shank, lateral malleolus, second metatarsal head, and calcaneus. Each subject undertook five jogging and five running trials. About 5 mins were given for subjects to have a rest between jogging and running sessions.

Figure 16. The experimental shoe with 4.5 cm heel.

3.2.2 Participants

Eleven experienced female wearers of moderate high-heel shoes (EW: age: 24.2 ± 1.2 years. height: 160 ± 2.2 cm. mass: 51.6 ± 2.6 kg) and eleven matched controls (IEW. age: 23.7 ± 1.3 years. height: 162.3 ± 2.3 cm. mass: 52.6 ± 4.5 kg) participated in this test with informed written consent, as approved by the Ethics Committee of Ningbo University. All the subjects were informed of the objectives, requirements, and experimental procedures. None of the subjects suffered from any musculoskeletal pathology that might affect regular jogging and running. All subjects were right foot dominant with the European shoe size of 37. EW had worn shoes with narrow heels of 3–6 cm height with a minimum of three times per week, 6 h per day for at least two years, IEW wore high-heeled shoes less than twice per month.

3.2.3 Data analysis

One gait cycle was defined as the duration from the initial contact of one foot to the subsequent contact of the same foot. Data of joint changes in three planes were time normalized to 0-100% of the gait cycle. The right-side motion during one gait cycle was analyzed for all kinematic and kinetic variables. A vertical GRF threshold of 20 N was used for the identification of heel strike events. Gait speed was calculated as the anterior-superior displacement of the right anterior-superior iliac spine marker dividing the corresponding time. Stride length (SL) was calculated as the anterior-posterior displacement of the suitable heel marker during two consecutive heel strike events. Stride frequency (SF) was computed as the inverse of one gait cycle duration. Stance phase percentage (ST/GC) was the percentage of stance phase in one gait cycle. Joint range of motion (ROM) during the stance phase was also computed for each joint in three planes. GRF was normalized to body weight (BW). The loading rate was calculated as the slope of the vertical GRF between 20% and 80% of the period from heel-strike to impact force.

3.2.4 Statistical analysis

ANOVA analysis with post-hoc Bonferroni correction was performed to assess the effect of wearing experience combined with gait speed (jogging and running) on spatiotemporal parameters including SL, SF, and ST/GC, joints ROM and peak angles. GRF and vertical average loading rate (VALR). Statistical results were considered significant if $p < 0.05$. In addition, linear regression analyses were performed to determine the correlation between speed and SL as well as speed and SF for both EW and IEW. All statistical analyses were performed using Stata 12.0 (Stata Corp, College station, TX).

3.3 Results and Discussion

3.3.1 Spatiotemporal characteristics

Differences in spatiotemporal characteristics were displayed in **Table 3.1**. There were no significant differences in each corresponding speed between EW and IEW. Compared with jogging, the speed of running increased significantly for EW and IEW. There were significant differences in SL, SF, and ST/GC. SL of EW was substantially larger than that of IEW, while SF showed to be smaller. Differences in the effect of speed on SL and SF were also significant. Compared with jogging, EW exhibited significantly larger SL while IEW exhibited significantly increased SL and SF during running. Furthermore, data from regression analyses showed significant differences in the correlation between speed and SL as well as speed and SF between the two groups (**Figure 17**). For IEW, the speed had a significant correlation with SL at the jogging level and SL and SF at the running level (**Tables 3.2 and table 3.3**). for EW, the speed had a significant correlation with SF at the jogging level and a significant correlation with SL at running level. IEW showed significantly larger ST/GC than EW at each corresponding speed level (**Tables 3.1**). The significance of the speed effect on ST/GC was shown in IEW. As speed increased, ST/GC increased significantly.

Table 3.1 Spatiotemporal parameters and kinematic parameters during stance phase.

	Parameters	EW jog	EW Run	IEW jog	IEW Run
Spatio-temporal	Speed (m/s)	2.50±0.14 [a,c]	3.05±0.14[d]	2.24±0.26f	2.84±0.29
	SL (m)	1.86±0.06[a,b]	1.86±0.06[d,e]	1.49±0.20f	1.79±0.16
	SF (steps/min)	82.43±3.48[b,c]	85.84±3.39[d,e]	90.74±2.92f	96.16±3.00
	ST/GC	0.35±0.02[b]	0.33±0.02[d,e]	0.42±0.03f	0.37±0.01
Sagittal plane ROM	Ankle	39.40±4.44[b]	36.16±2.42[d,e]	47.88±2.59	43.89±3.70
	Knee	30.37±2.11[b]	30.97±0.86[d]	29.90±2.67	30.16±1.79
	Hip	39.22±3.73[a]	46.12±3.88[d]	39.99±6.06	44.29±5.19
Frontal plane ROM	Ankle	4.90±0.48[b,c]	5.76±0.46	6.66±0.26	6.30±0.44
	Knee	7.23±2.17 [b,c]	9.19±1.15	11.27±1.20	11.04±1.63
	Hip	8.70±1.56	8.72±1.56	8.80±0.93	10.45±1.78
Transverse plane ROM	Ankle	21.38±2.08	22.32±3.06	22.67±1.05	21.57±0.83
	Knee	16.91±2.21	16.26±1.72[e]	18.34±1.08	19.97±1.26
	Hip	12.94±2.34[c]	16.69±3.53	17.74±3.82	19.56±4.40
Sagittal plane Peak	Ankle	12.86±2.10	10.64±0.86	12.94±1.88	10.73±1.02
	Knee	39.47±1.80[b,c]	42.73±2.13	45.01±2.04	44.16±2.07
	Hip	27.70±2.82[a]	36.02±2.94[d,e]	27.69±4.00	29.15±4.10
Frontal plane Peak	Ankle	5.51±0.40[a,b,c]	6.80±0.23[e]	7.51±0.43	7.73±0.33
	Knee	4.57±0.60[c]	5.84±0.69	5.16±0.58f	7.12±0.89
	Hip	6.80±0.89[b,c]	7.73±1.01[d,e]	12.62±1.23	13.37±2.07
Transverse plane Peak	Ankle	-23.58±1.05 [a,c]	-26.82±1.90	-26.29±1.06	-26.73±0.55
	Knee	12.13±2.19 [a,c]	15.95±1.62	15.44±1.52	15.88±0.99
	Hip	15.34±1.53	16.91±1.56	14.69±0.95	14.72±0.99
Sagittal plane Touchdown angle	Ankle	-10.95±2.15 [b,c]	-9.97±0.85	-14.34±2.31	-13.63±0.72
	Knee	18.72±5.87[c]	24.06±3.42	23.39±2.22	26.34±1.47
	Hip	27.54±2.84[a]	35.99±2.96	27.61±3.92	29.09±4.10

[a] Significant difference between EW jog and EW run.

[b] Significant difference between EW jog and IEW jog.

[c] Significant difference between EW jog and IEW run.

[d] Significant difference between EW run and IEW jog.

[e] Significant difference between EW run and IEW run.

[f] Significant difference between IEW jog and IEW run.

3.3.2 Joint kinematics

Figure 18a –18c shows comparisons of three-dimensional joint changes at the ankle, knee, and hip during one gait cycle between EW and IEW. Peak angles in the planes during the stance phase showed clear differences between EW and IEW while speed effects were not obvious within IEW. In the sagittal plane, peak knee flexion of EW during jogging was significantly smaller than IEW during jogging and running. At the hip, peak flexion of EW during running was significantly larger. In the frontal plane, peak ankle inversion of EW during jogging decreased, with $p < 0.05$. Also, compared with IEW during running, EW during running showed significantly smaller peak inversion. A significant increase of peak knee flexion in this plane existed in IEW running compared with jogging and EW jogging. Peak hip flexion was shown to be significantly smaller in EW. In the transverse plane, EW showed smaller ankle and knee peak rotation during jogging than running and IEW running. Larger hip peak rotation was found in EW during running in comparison with IEW during jogging and running. Ankle ROM of EW in the sagittal plane was significantly smaller in comparison with that of IEW. Knee ROM of EW was larger than that of IEW during jogging with significance. EW during running showed a significant increase in knee and hip ROMs compared with IEW jogging. Hip ROM during the running of EW was also shown to increase than IEW jogging. In the frontal plane, ankle and knee ROMs of EW during jogging showed a significant decrease in comparison with IEW jogging and running.

There were no effects of whether speed or wearing experience on hip ROM in this plane. In the transverse plane, ROM at the knee of EW was significantly smaller than that of IEW running. At the hip, EW only showed a significant decrease during jogging than IEW running. Speed had no apparent effects on joints ROM of IEW. At touchdown, the ankle of IEW was at a more plantar-flexed position. A significant increase in knee flexion at initial contact was found in the comparison between EW jogging and IEW running. Hip flexion of EW during running was significantly larger than that of IEW at both speeds. Also significant, hip flexion of EW during running increased more than during jogging.

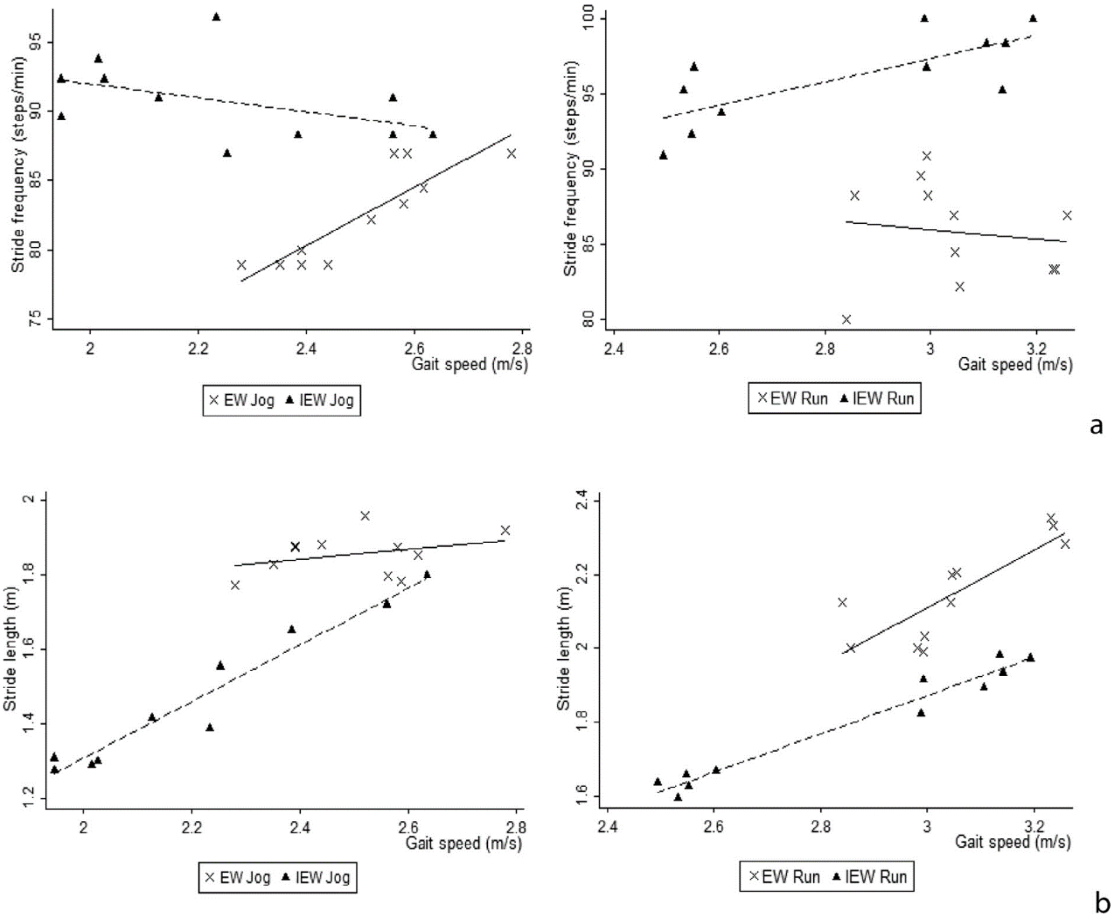


Figure 17. (a) Regression analyses (Stride frequency was plotted versus speed. Each sign represents the average for each subject for a specific condition). (b) Regression analyses (Stride length was plotted versus gait speed. Each sign represents the average for each subject for a given situation).

Table 3.2 Results of linear regression for SF.

Variables	R^2		Prob > F	
	EW	IEW	EW	IEW
SF Jog	0.761	0.197	0.001	0.171
SF Run	0.017	0.568	0.701	0.007

Table 3.3 Results of linear regression for SL.

Variables	R^2		Prob > F	
	EW	IEW	EW	IEW
SL Jog	0.112	0.953	0.314	0.000
SL Run	0.677	0.962	0.002	0.000

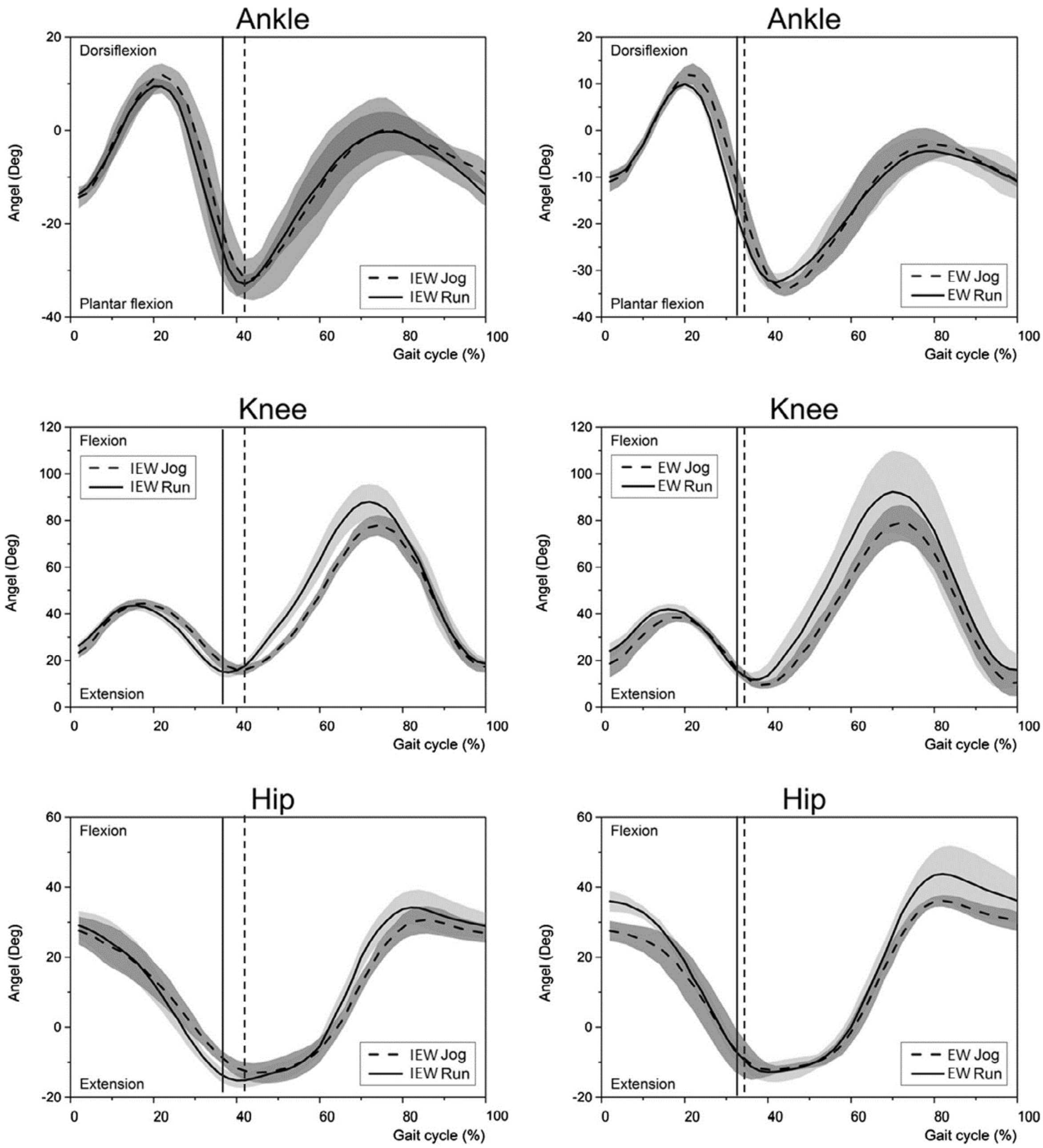


Figure 18a. Changes of joint angles in the sagittal plane (solid and dash straight lines represent the transition from stance phase to swing phase during running and jogging, respectively).

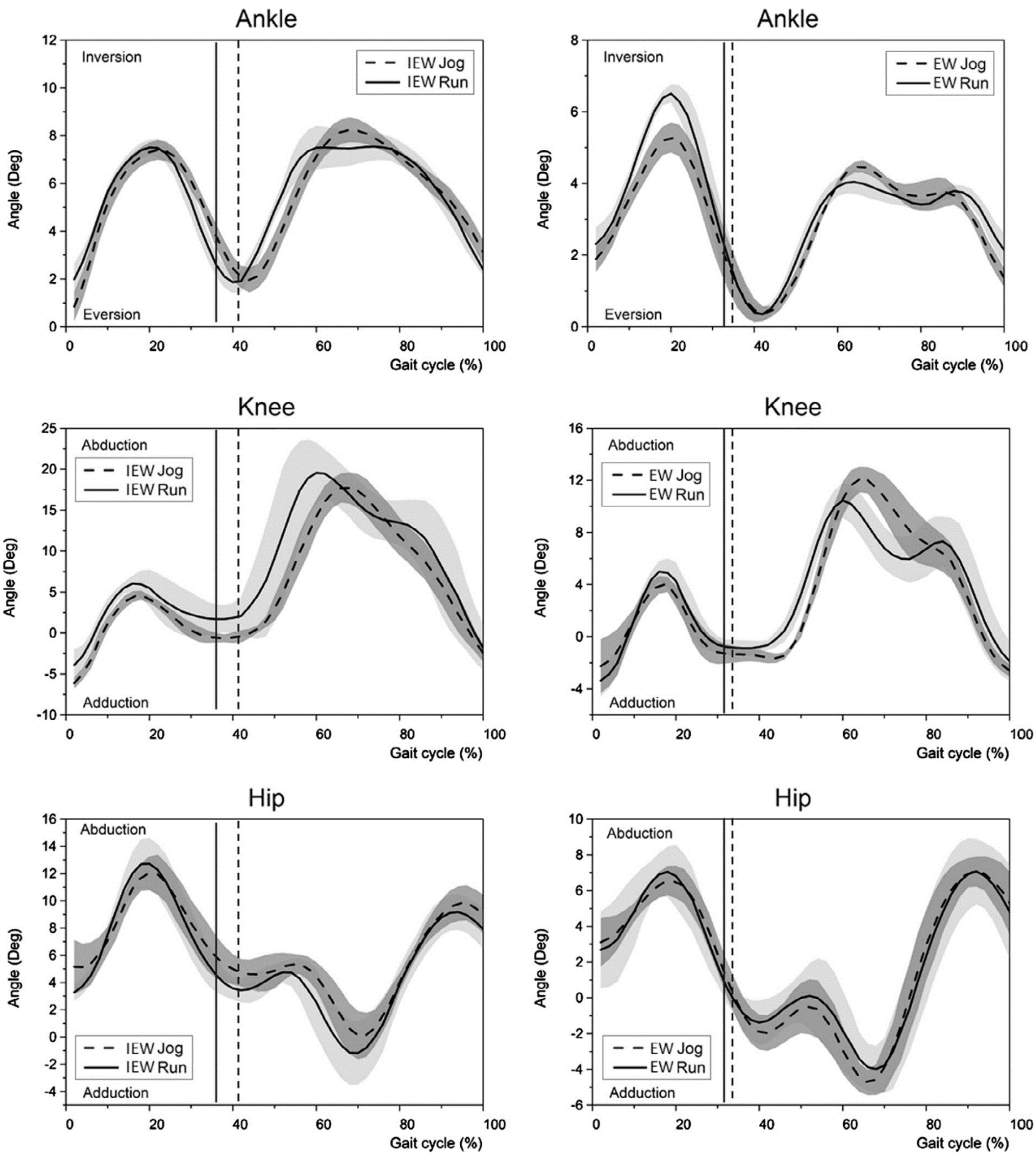


Figure 18b. Changes of joint angles in the frontal plane (solid and dash straight lines represent the transition from stance phase to swing phase during running and jogging, respectively).

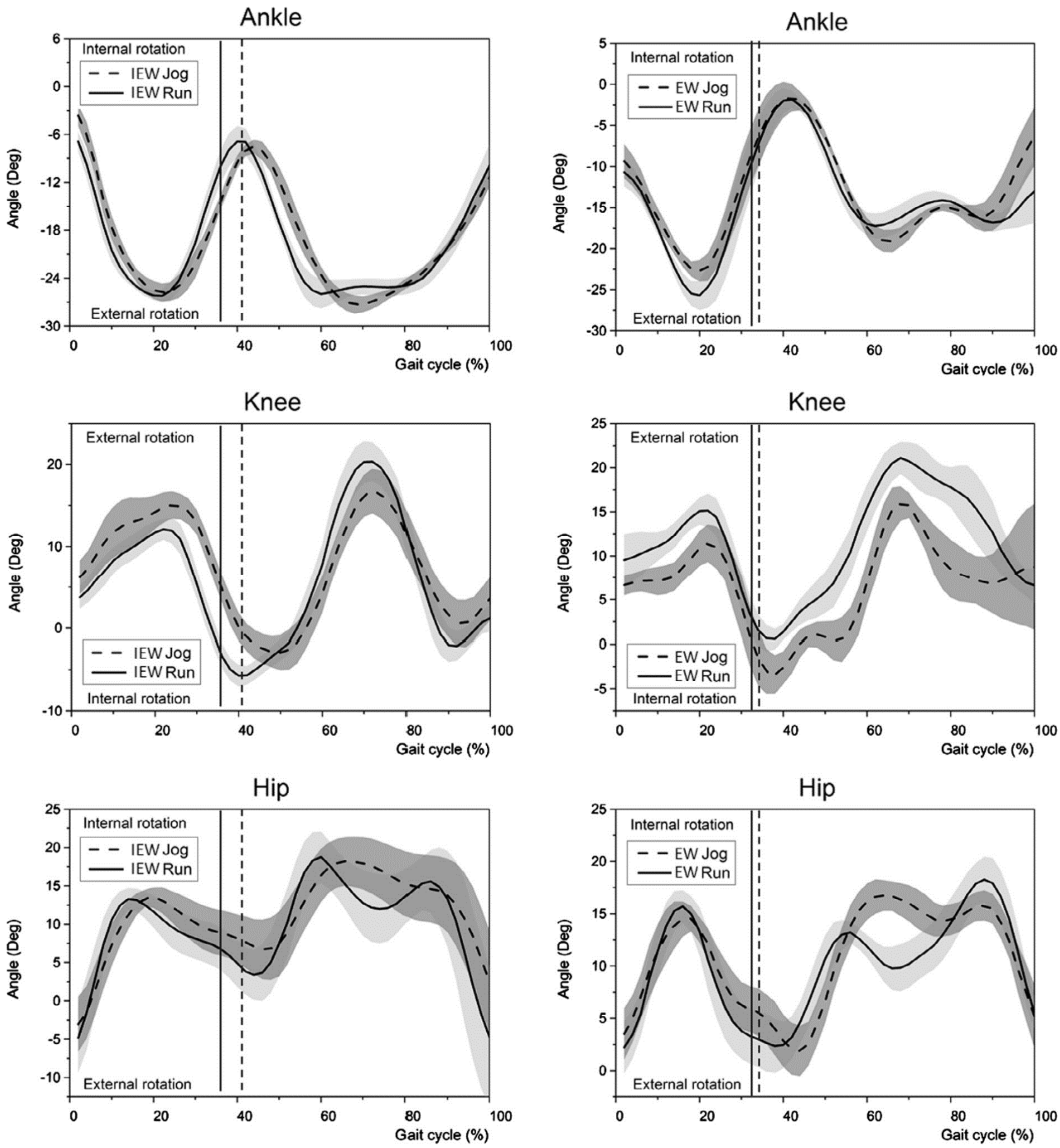


Figure 18c. Changes of joint angles in the transverse plane (solid and dash straight lines represent the transition from stance phase to swing phase during running and jogging, respectively).

3.3.3 GRF and VALR

Table 3.4 and **Figures 19, and 20** summarize the main GRF and VALR characteristics of EW and IEW. Impact force showed no significant differences between wearing experience or speeds. The maximal vertical GRF of EW during jogging was significantly larger than that of IEW jogging. EW during running showed significantly larger maximal vertical GRF compared with IEW jogging and running.

Table 3.4

GRF and VALR characteristics of EW and IEW wearers during jogging and running

	EW		IEW	
	Jog	Run	Jog	Run
VALR (BW/s)	62.40 ± 10.46[a]	102.66 ± 4.99[c,d]	62.17 ± 14.38	78.15 ± 17.00
Impact force (BW%)	1.19 ± 0.18	1.35 ± 0.39	1.24 ± 0.24	1.52 ± 0.28
Maximal vertical GRF (BW%)	2.42 ± 0.12[b]	2.51 ± 0.14[c,d]	2.05 ± 0.24	2.27 ± 0.12

[a] Significant difference between EW jog and EW run.

[b] Significant difference between EW jog and IEW jog.

[c] Significant difference between EW run and IEW jog.

[d] Significant difference between EW run and IEW run.

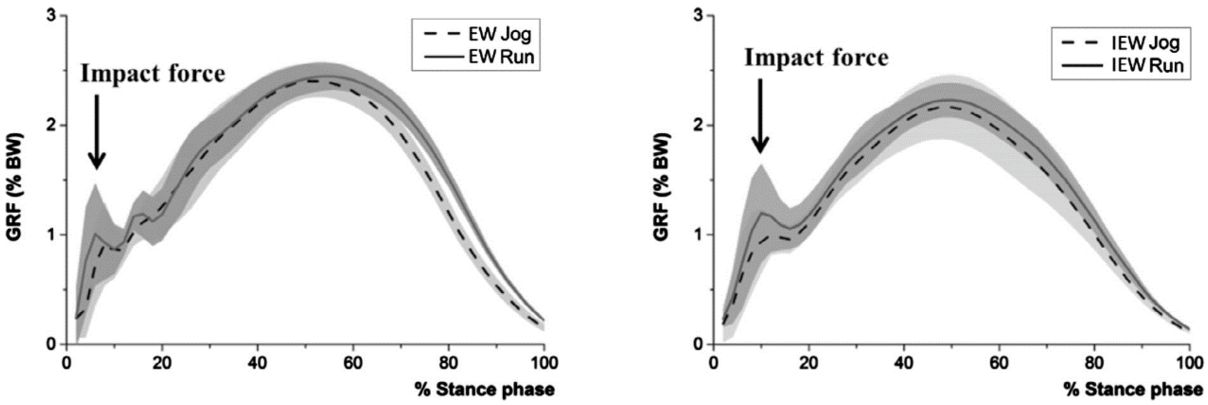


Figure 19. The GRF of EW (left) and IEW (right) wearers during jogging and running.

Figure 20. The VALR of EW and IEW during jogging and running.

Discussion

This study identified the long-term effects of wearing moderate high heel shoes on kinematic and kinetic changes in terms of spatiotemporal parameters including SL, SF, and ST/GC. kinematics including joint motion characteristics of the ankle, knee, and hip. kinetics including impact force and maximal vertical GRF, and VALR during jogging and running.

Most studies had ignored the possible importance of high-heeled wearing experience [110]. Different from the first hypothesis, EW and IEW showed comparable preferred speeds during jogging and running. However, they performed different SL and SF for a certain speed. Statistical data showed that IEW adopted higher SF with shorter SL no matter during jogging or running. A similar result has been reported that elder women in China with bound feet took more steps and shorter strides than those with normal feet when walking at a similar speed [111]. Chien, Lu, and Liu [92] noted that in comparison with habitual wearers, inexperienced controls showed less stable body balance during walking. This increased SF with decreased SL may be a strategy adopted by IEW to compensate for reduced local dynamic stability through enhancing medial-lateral and backward margins of stability, respectively [112]. However, a limitation of this study was that there was no evidence to prove whether EW possesses better body balance than IEW. The next step was to examine the influence of long-term use of high heels on balance control in jogging and running with parameters that could quantify gait stability such as short-term Lyapunov exponent, margins of stability, COM-COP inclination angles, and the rate of inclination angle changes [113, 114, 92].

As the speed increased from jogging to running, the SL of both groups improved significantly, while SF showed a significant increase in IEW. Therefore, it could be speculated that EW adapts speed increases with increasing SL while IEW adapts speed increases with both SL and SF. Further, speed showed a linear correlation with SL in IEW and SF in EW during jogging. As to running, speed showed a linear correlation with SL and SF in IEW and SL in EW. For IEW, the speed had a significant correlation with SL at the jogging level and SL and SF at the running level, there was no correlation with SF at the jogging level. For EW, the speed had a significant correlation with SF at the jogging level and a significant correlation with SL at running level. These findings from regression analysis suggest that EW adopted an SF-dominant strategy while IEW adopted an SL-dominant strategy in regulating jogging speed. With speed increasing to running, SL and SF concurrently played a role in regulating speed for IEW. Compared with SF, SL showed a more obvious effect with a higher R2. For EW, the increasing running speed was mainly acquired by the increase of SL. The overlong stride during high-heeled running may potentially cause falls or musculoskeletal injuries. Consistent with the concept that contact time decreases with running speeds increase during normal shod or barefoot running [115, 116], results from IEW in this study also revealed significantly decreased ST/GC as speed increases. In addition, IEW showed longer stance phase duration in one gait cycle at both speeds, indicating a cautious gait style to deal with body instability.

One might expect greater joint ROMs with faster speed, particularly while running on high heels, however, there was, in fact, no obvious effect of speed on joint ROMs within each group except for the ROM of hip flexion-extension. Hip ROM in the sagittal plane of EW increased significantly during running in comparison with that of EW and IEW during jogging due to the larger peak flexion. Significant change in hip ROM was also observed in the transverse plane. EW presented significantly smaller ROM during jogging than IEW running. Similarly, knee ROM of EW in the frontal plane during jogging was shown to be smaller. In the transverse plane, the knee ROM of EW during running was significantly smaller. These reduced motions in hip and knee joints of EW may be associated with larger leg stiffness after long-term use of high heels to maintain stability. However, it has been suggested that exaggerated or insufficient stiffness throughout the lower limb would predispose individuals to at high risk of injury [117, 118]. Too little stiffness may allow for excessive joint motion leading to soft tissue injury. Conversely, extreme stiffness would lead to increased peak forces, loading rates, and shock. Knowledge of these indicates that individuals with different high-heeled wearing experiences should consciously modify their lower limb stiffness [117]. EW showed a significant increase in knee ROM in the sagittal plane at both speeds compared with IEW jogging. The larger knee motion mainly results from extension during push-off, which could help to lengthen stride. At the ankle, the mobile ankle in the sagittal plane of IEW serves as a less effective lever for applying muscle force to the ground, which requires greater muscle work to achieve similar mechanical output generated by triceps sure during the propulsive period [119]. This was a potential factor exacerbating muscle fatigue and strain injury, particularly IEW. Relatively, EW showed better control of joint motions at both speeds.

Results on peak joint angles support the hypothesis that EW showed limited peak angle except for peak hip flexion and internal rotation during running. However, the speed effect was not obvious within IEW. In the sagittal plane, EW showed significantly larger hip peak flexion during running, which may help to lower the centre of mass to enhance body balance as speed increases [120]. Larger hip flexion at initial contact was also found in EW during running. This has been reported to be a compensatory mechanism to attenuate increases in impact force to prevent injury [97, 121]. In opposition to the statement that habitual wearers had much greater increases in knee flexion during the stance phase of walking than the inexperienced, this study showed significantly smaller peak knee flexion in EW during jogging [108]. Although larger knee flexion during walking facilitates maintaining balance, extended flexion of IEW during jogging and running may lead to excessive knee extensor moment and rectus femoris activity, both of which were causes of knee overload [97-100, 122-124]. The 7% increased extensor moment and 14% increased knee abduction moment were observed respectively in young and elderly women when walking on shoes with a 3.8 cm heel height compared with flat shoes [124]. Research also proved that large quadriceps forces induced by increased knee flexion would increase proximal anterior tibial shear force, a major factor of anterior cruciate ligament strain [98, 125]. In addition, increased knee flexion has been correlated with the warding moving of the centre of mass over the larger and more stable forepart of the foot, possibly resulting in larger forefoot plantar pressure [126]. Gu, Sun, et al. [105, 127] found a significant increase of plantar pressure in the first metatarsal region with heel height increasing during jogging. Knee flexion of IEW at initial contact increased significantly during running than that of EW jogging. Increasing knee flexion was common in high-heeled gait to reduce imbalance [110]. As to the ankle, EW showed significantly smaller ankle plantarflexion at touchdown, which was prone to landing with the isolated narrow heel.

Overall, joint motions in the frontal plane were more subtle than in the sagittal plane. At the hip, IEW showed significantly larger peak abduction at both speeds compared with EW, which possibly induces larger medial-lateral excursion of the centre of mass leading to body instability. IEW showed significantly increased peak abduction at the knee during running than both groups during jogging. Previous study has revealed that increased knee abduction with heel height rising during jogging may increase knee joint loading [127]. It has been calculated that a 1% increase in knee abduction moment increases the risk of progression of osteoarthritis by 6.46 times [99]. At the ankle, EW showed less peak ankle inversion than IEW. One possible explanation for the reduced inversion was that pronator activity increased after long-term use of high heels [92]. Compare with inexperienced controls, experienced subjects showed larger ankle pronator moment in the walking test [92]. Ankle peak inversion increased significantly as speed increased in EW. Coupled with a plantar-flexed joint position, a larger inversion angle put wearers at high risk of lateral ankle sprain [128]. Motions in the transverse plane were more flexible at the ankle compared to the hip and knee. EW showed significantly less peak ankle and knee external rotation during jogging than both groups during running. These results indicated that regardless of wearing experience, wearers were caught in a high incidence of ankle sprain and stress fracture during high-heeled running. Joint instability of IEW could be attributed to weak control ability and that of EW during running was potentially caused by landing with the narrow heel. Another clear consequence of EW's more obvious heel strike pattern was reflected by the significantly larger hip peak rotation during running compared with IEW during jogging and running. This may partly reduce excessive knee rotation during running for EW.

During the swing phase, EW also showed different gait kinematics from IEW. In the sagittal plane, the maximum hip and knee flexion increased as speed increased, which was linked with larger stride length [120]. EW showed larger maximum hip and knee flexion compared with IEW. These were in line with the outcome of the largest stride length exhibited in EW during running. Additionally, larger hip and knee flexion during the swing phase could contribute to avoiding tripping caused by the low minimum distance between the toe and the ground [129]. On account of less maximum flexion of the hip and knee, IEW showed slightly less maximum ankle plantar flexion. However, knowledge of toe clearance excursion and the timing of the minimum toe clearance in one gait cycle during high-heeled jogging or running remain absent. Moreover, angle-time curves of EW in the sagittal plane showed an apparent change of hip extension during the second half of the swing phase, which related to better avoiding excessive deceleration that would occur at touchdown if the foot was too far ahead of the centre of mass [120]. In the frontal and transverse planes, differences may be caused by a compromised neuromuscular control system which was mediated by central mechanisms [130]. Previous research has demonstrated that alternating joint position sense may lead to kinematic changes during gait [131]. Based on this, larger ankle inversion and external rotation of IEW throughout the swing phase during jogging and running could be explained. It might be due to different joint position senses under a passive plantar-flexed state while wearing high heels. Similar results were also observed in chronic ankle instability subjects during jogging compared to healthy controls [132, 133]. It was important to investigate if joint motions during the swing phase had influences on kinetics and kinematics during the stance phase. As reported previously, kinematic alternations at mid-swing could decrease impact peak and loading rate during running with sports shoes [134].

GRF and VALR had both been reported as factors leading to running injuries [135, 136]. Impact transients were sudden forces with high rates and magnitudes of loading that travel rapidly up the body and thus may contribute to the high incidence of running-related injuries, especially tibial stress fractures, patellofemoral pain, Achilles' tendinopathy, and plantar fasciitis [137, 138]. To our best knowledge, this was the first work to investigate the difference in GRF during moderate high-heeled jogging and running among different experienced wearers. For EW, the GRF-time curve was characterized by an initial sharp peak immediately followed by a second peak during shock absorption, particularly in running. The obvious second peak during running was likely to be attributed to an immediate and evident slap of the forefoot on the ground followed by a heel strike. The ‘‘double-shock’ force in running may aggravate joint injuries. In contrast, IEW showed relatively easy and fluent rollover of the foot from the heel to forefoot contacting with the ground during jogging and running, resulting in one spark peak at heel strike. Impact force showed no differences across four conditions. However, Maximal vertical GRF of EW during jogging was shown to be significantly larger than that of IEW at the same speed. Moreover, EW also showed significantly larger Maximal vertical GRF during running compared with IEW at both speeds. The increased GRF found in EW appears to contribute to increasing plantar flexor and pronator moments at the ankle [92], which helps reduce ankle instability during push-off. Another obvious difference between EW and IEW was found in VALR. EW showed significantly larger VALR during running, which was largely due to faster reaching to impact force. This may relate to the higher acceleration of the foot at touchdown with a heel, suggesting that the negative effect caused by the strike pattern exceeds the positive effect of the larger knee and hip flexion of EW in attenuating impact shock. It has been widely documented that impact force with a fast-increasing rate would create a robust shock wave during a heel strike, which was then transmitted up to joints and the musculoskeletal system [139] potentially causing lower limb soft tissues damage and back-pain complaints [140] and eventually leading to degenerative joint disorders [141].

This study provides a basis for evaluation of lower limb mechanics in moderate high-heeled jogging and running based on wearing experience and also provides information on moderate high-heel shoe design for individuals with different wearing experiences. Our findings along with those from Luximon et al. [142] suggested that the small supporting base, especially at heel strike, was a major factor of reduced stability for habitual high-heeled wearers. Despite wearing experience, however, running on moderate-high heels increases the risks of knee osteoarthritis and ankle sprain. The heel geometry including height [100, 127], base size [142], and even angle between the sole and heel, etc. should be integrated into a high-heel shoe designed for individuals with different wearing experiences.

A key finding of this study was that compared with IEW, EW showed reduced joint ROM during the stance phase to prevent excessive joint loading except for motions of the knee and hip in the sagittal plane that respectively aid in propulsion and load attenuation instead. Moderate high-heel shoes placed IEW at a greater risk of joint and soft tissue injury with generally larger peak angles during the stance phase. However, the effect of these conservative control strategies adopted by EW was partially lost during running in comparison with jogging. As speed increased from jogging to running, EW mainly relied on increasing SL, leading to landing with narrow high heels, which consequently resulted in an extremely high loading rate. From a kinetic perspective, EW also tended to have a bone-on-bone injury in high-heeled running. In conclusion, moderate high-heeled wearers who had to run, whether regularly or occasionally, were putting themselves at high risk of lower limb damage. Wearers must control joint stability and strike patterns consciously.

In summary, EW regulated their stride speed through stride length, and the maximum vertical GRF and vertical load loading rate were increased when running at a fast speed, which would increase the risk of injury to the musculoskeletal system of the lower limbs. This was crucial because it was essential to provide running shoes with more cushioning for EW runners. IEW regulated their stride speed through stride length and stride frequency, which helped compensate for the reduced balance caused by inexperience in wearing. Still, the risk of knee osteoarthritis and ankle sprain was higher due to the greater joint mobility. Compared to EW, it should pay more attention to stability when designing a pair of running shoes for IEW runners.

3.4 Conclusion

The aim of this study was to investigate the differences in lower limb kinematics and kinetics between experienced (EW) and inexperienced (IEW) moderate high-heel wearers during jogging and running. Eleven experienced female wearers of moderate high-heel shoes and eleven matched controls participated in jogging and running tests. A Vicon motion analysis system was used to capture kinematic data and a Kistler force platform was used to collect GRF. There were no significant differences in jogging and running speed, respectively. Compared with IEW, EW adopted a larger stride length (SL) with a lower stride frequency (SF) at each corresponding speed. EW enlarged SL significantly during running while IEW increased both SL and SF significantly. Kinematic data showed that IEW had a generally larger joint range of motion (ROM) and peak angles during the stance phase. The speed effect was not apparent within IEW. EW exhibited a significantly increased maximal vertical GRF (Maximal vertical GRF) and vertical average loading rate (VALR) during running, which was potentially caused by an overlong stride. These suggest that both EW and IEW were at high risk of joint injuries when running on moderate-high heels. It was crucial for wearers to do some running on moderate-high heels to control joint stability and balance SL and SF consciously.

4 Effect of the high offset running shoes with a special structure of midsole on the lower limb biomechanical

4.1 Introduction

From the opinion of running performance, Healey and Hoogkamer [87] highlighted that there was no significant effect on the energy savings in the VF by decreasing the LBS, which indicated that the function of the plate in the 4% energy savings was the limitation. In addition, there was a new effect on running mechanics that the influence of the curved carbon fiber plate inserted into the midsole worked as a 'teeter-totter' [143]. The plate stiffens the MTPJ and works as a lever to decrease the work rate at the ankle [144]. The principle was that the point of application of the GRF moves anteriorly and towards the front end of the curved carbon fiber plate during the second half of ground contact [143]. In other words, carbon plates altered the position as well as the magnitude of ground reaction forces [54], potentially leading to optimized musculoskeletal performance of the athlete and running involving reduced energy loss in the metatarsophalangeal joint [4, 59]. It was not yet understood whether the midsole material [73, 145], midsole construction [146], or shape of the carbon-fiber plate [143, 147] contribute more to these 'racing running shoes. Some researchers argued that increasing the LBS of shoes could also increase the material of the midsole [148]. In addition, it was well known that gender differences in lower extremity structure and running models were noticeable [26]. A greater active hip internal rotation, vertical ground reaction force (GRF), accessible vertical torque, peak hip flexion angle, and negative work were displayed in females than that of men [26]. Women had a greater ratio of hip-width to femur length, which resulted in greater hip internal rotation. In addition, women exhibiting higher Q angles would increase lateral quadriceps pull on the patella. It would exacerbate patellar tenderness or recurrent lateral patellar subluxation conditions, which induced a higher incidence of patellofemoral joint pain [45]. Tendon stiffness might be associated with the regular use of high-heeled shoes in some females, leading to higher hypertrophy and shortening of the Achilles tendon, a higher pre-activation amplitude of the peroneal muscle greater gluteus maximus muscle activation [106]. Therefore, it was essential to design running shoes according to female athletes' biomechanics and body structure characteristics.

From the perspective of running injuries, Seventy-six percent of knee pain was found in women. It had been recorded that the number of traditionally shod female runners who landed with a rearfoot strike (RFS) was more than 80% [14, 75]. Commonly running-related injuries such as tibial stress fractures, patellofemoral pain, and plantar fasciitis were linked to the high loading rates and impact transients during rearfoot striking [24, 149, 150]. It was stated that female runners would be twice as likely to sustain certain running injuries like the above sports injuries [151]. Therefore, it was imperative to improve the cushioning of running shoes related to loading rates and impact transients during rearfoot striking. Running shoe manufacturers were focusing on cushioning while making shoes for female runners to diminish running injuries [152]. It was previously proved that reducing the impact forces by wearing a more cushioned shoe may release stress on musculoskeletal tissue [153]. It was worth mentioning that Burns and Tam [155] introduced the midsole thickness as the main footwear characteristic that improves running performance and cushioning. Studies had also been conducted to compare how different thicknesses of soles affect lower limb muscle activity [163]. Experimenters investigated the difference between various thicknesses and barefoot during walking. They showed that an increased lower limb muscle activity, earlier work on the peroneus longus muscle, which controls the foot valgus on the lateral side of the lower leg, and an increased moment of the subtalar joint were found in athletic shoes than that of the barefoot condition [154]. Increasing the midsole thickness could protonate the runner's effective leg length, such as the VF which has a 31 mm heel height [155]. It could decrease energy loss for the runner by increasing an 8 mm effective leg length [72, 156]. Besides, some researchers also figured out that the effect of midsole thickness was about 1% for running economy [157].

The heel-to-toe drop would increase with the increased thickness of the heel material. Footwear such as the elements of the midsole and heel-toe drop (HTD) had been considered in studies of young athletes that could influence a runner’s performance, particularly in cushioning [14, 158-160]. Recently, the HTD as a key feature in shoe design has been linked to the risk of running injury [152]. Several authors from the biomechanics view had researched the effect of different HTDs. Richert et al [161] found that a 4mm HTD induced a higher vertical loading rate compared to 8mm and 12mm HTD. What's more, the lower limb biomechanics performance of a 4mm HTD wasn't similar to barefoot running. During the investigation, there was no specific adaptation in spatiotemporal variables and kinematics between the three kinds of shoes (0mm HTD, 6mm HTD, 10mm HTD) during the investigation [162]. Gabriel Gijon-Nogueron [163] found that heel-to-toe drop (4 and 12mm) does not directly affect the spatiotemporal parameters of the running cycle in female runners. It was still being debated that the height of the heel over 45mm was associated with some gait troubles, such as postural disorders and changing spatiotemporal parameters because it would modify the muscle balance up to muscle overuse and strain injuries [96, 97, 164, 165]. It was reported in [166], where it was essential to affect the center of pressure that the height of the rearfoot had to be more than 2.5cm. The vertical loading rate and the associated transient peak reduced as the shoe drops growing [160]. Above all, few shoe manufacturers had made running shoes with an over 12mm HTD and also had little research about it according to female runners' characteristics. It was worth researching whether increasing the HTD by more than 12mm could significantly reduce impact force or not. S. Hessas [167] asked the volunteer to test in barefoot or equipped with three stiffness with the same lift height of 20 mm. They found that it was no significant influence of material stiffness anterior-posterior displacement of the center of pressure and metatarsal pressures, but obviously affected the peak pressure on the calcaneus. Furthermore, it was insufficient to combine investigating the effect of different HTDs on lower limb biomechanical and perceptual sensitivity. The center of gravity of the running shoe would become higher with the increase of the heel thickness, which would affect the dynamic stability of the running, so it was necessary to think about improving the stability of the heel running shoe by adjusting the midsole heel material or structure.

This study was targeted to investigate how changing the lower limb of biomechanics when wearing a pair of special running shoes (IRS) with a 16mm HTD. The shoes had three layers of the midsole (up and lower layers were for cushioning, and the middle layer was for support) and investigate the biomechanics difference between IRS and normal running shoes (NRS). According to the previous literature, it was hypothesized that (1) a reduced plantarflexion at touchdown and lower vertical loading rate would be found in the IRS model than that of NRS. (2) The IRS would increase the joints moment in the sagittal plane compared to the NRS.

4.2 Methods

4.2.1 Biomechanical Data Collection

Participants performed eight valid right foot rearfoot strike running trials per shoe condition on a 145 m concrete indoor running loop. Data collection methodology was carried out as in the previous research [82]. A valid trial was one within the specified velocity range (3.6 m/s ± 5%) and made up of the whole right foot contacting the force plate area. Before, data collection participants warmed up for about five minutes and were acquainted with the target speed and shoe conditions by running 2 laps in each shoe condition. Upon failing to match the required speed in the first two laps, further familiarization laps were performed as necessary. GRF and 3D kinematic measurements were carried out as the previous methodology.

4.2.2 Subjective Perception

Testing took place simultaneously with biomechanical data collection, with participants filling in on the questionnaire immediately after completing the eight successful trials required for the respective shoe condition. Runners assessed six perception variables (shoe weight, fit, arch support, cushioning, stability, over preference) on a questionnaire that had been repeatedly highlighted in some papers [168, 169]. Fifteen cm visual analogue scale (VAS) was carried out, these had been previously applied for running footwear assessment [168, 170]. Participants were shown and explained both variables prior to each shoe condition during the initial familiarization and data collection.

4.2.3 Participant and Experiment footwear

Fifteen female runners who were used to wearing high heels [mean (SD) 39.00 (10.09) years, 1.58 (3.37) m, 50.34 (3.24) kg, 20.14 (1.28) kg/m2] joined this research. All of the participants were recruited from the Xiamen running club and identified themselves as rearfoot strike runners. All participants had been confirmed in foot size (EU 37 ± 0.5) by the Brannock Device (The Brannock Device Co., Sy-racuse, NY, USA) before the official test. Participants were free from injury for at least six months before this study.

There were two kinds of experimental footwear shoes: IRS, Normal running shoes: NRS, which differ in their offset, mechanical midsole hardness, rearfoot im-pact, and forefoot flexion properties used in this research **(Figure 21)**.

Mechanical impact measurement took the final five impacts from 30 repetitive im-pacts by an impact tester (Brentwood, NH, USA) on the experimental shoes with a drop height of 5.0 mm and a drop mass of 8.5 kg [171]. Mechanical flexion measurements fixed the forefoot area in the location of 70%-foot length (heel to toe) **(Figure 22)**, then bending at 45 degrees was performed by applying a dynamic shoe flexor device (Brentwood, NH, USA) to measure the shoe LBS and energy return [84]. All shoe conditions featured an Xtep Softpad Lite HD foam insole with a forefoot and rearfoot thickness of 5.0 mm. All characteristics of shoes were shown in **Table 4.1**.

Figure 21. Picture of the IRS prototype used during running (A), and NRS (B).

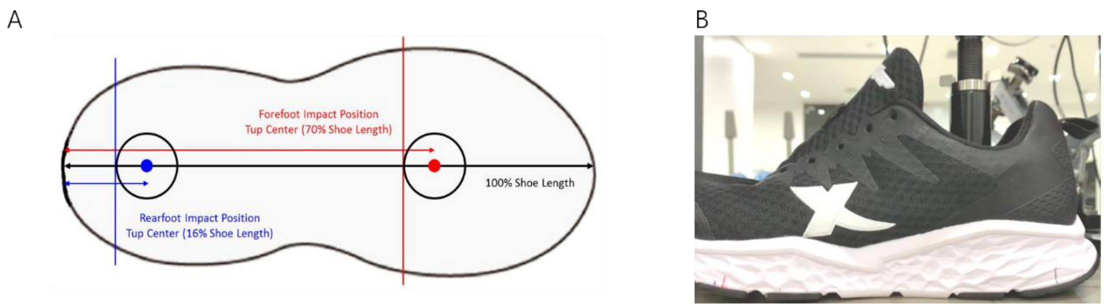


Figure 22. A： Xtep procedure of shoe sample dimension measurement for impact testing, B: Schematic diagram of the impact test apparatus.

Table 4.1 the characteristics of experimental shoe condition.

Measurement method	Characteristics	Running Shoe models	
		IRS	NRS
Basic information	Mass(g)	260.9	205.0
	Rearfoot thickness(mm)	32	18
	Forefoot thickness(mm)	16	11.5
	Offset (mm)	16	6.5
	Rearfoot width (mm)	81.67	80.6
	Forefoot width (mm)	95.6	100
	Midsole material	EVA	EVA
	Midsole hardness (Asker C)	Up to 40C middle 50C under 40C	55C
	Outsole material	Rubber	Rubber
	Outsole hardness (Asker)	62A	62A
Rearfoot impact	Peak acceleration(g)	9.9	13.7
	Energy return (%)	56.61	64.53
Forefoot flexion	Peak torque (Nm)	13.49	9.79
	Stiffness (Nm/deg)	0.307	0.169
	Energy feedback (%)	24.78	27.14

4.2.4 Data analysis and Statistical analysis

In the trial, valid data could be used when the first impact peak and shoe ground angle more than zero appeared. We used the Vicon Nexus 2.7 and Visual3D (C-Motion, Germantown, MD, USA) to process the collected experimental data. A fourth-order low pass Butterworth filter was used with 100 Hz (kinetic) and 10 Hz (kinematic) cut-off frequency. The angle, the angular velocity, the GRF, and the work of the hip, the knee, the ankle, and the MTP joints of the right lower limb were measured during the stance phase using Visual3D (C-Motion, Germantown, MD, USA). The XYZ Cardan sequence was used to calculate lower limbs' kinematic and kinetic data, in which X represents flexion-extension, Y represents abduction–adduction, and Z represents internal-external rotation. SL was calculated as the anterior-posterior displacement of the right heel marker during two consecutive heel-strike events. The loading rate was calculated as the slope of the vertical GRF between 20% and 80% of the period from heel-strike to impact force. All vertical GRF variables were calculated based on the recommendations by Ueda et al. [172].

For 0D parameters includes spatiotemporal parameters, average loading rate1, peak loading rate1, peak vertical force 1, joints moment, and some kinematics parameters. Shapiro–Wilk tests were adopted in this study for normality distribution. Permutation non-parametric tests were chosen with Matlab (The Mathworks, Naticks, MA) when the null hypothesis of the normality test was rejected. Paired t-tests were applied when appropriate. Statistics 0D parametric tests were processed by SPSS (24, IBM Corp., Armonk, NY, USA). Effect sizes (Cohen's d) were displayed for all statistical tests (0.2< Cohen's d<0.5 = small effect, 0.5< Cohen's d <0.8 = medium effect, Cohen's d >0.8 = large effect).SPM technique [85, 86] was used to assess the time series parameters such as one-dimensional (1D) kinematic and force trajectories.

SPM T-paired were performed on shoe effects for every 1D parameter [173]. SPM tests were calculated with the SPM1D v0.4 for MATLAB (www.spm1d.org,[85]). The statistical significance alpha levels were set to <.05 for all statistical tests.

4.3 Results and Discussion

Running speed was 3.59±0.27,3.61±0.31 for IRS and NRS (P=.673). Shapiro-Wilk tests revealed that 100% of biomechanical variables and 100% of perception variables were normally distributed (both $p > .05$).

4.3.1 Kinematics variables

Concerning joint angles at touchdown, the ankle of IRS was at a more dorsiflexed position (p=.023) (**Table 4.2**), with no significant change of knee, hip flexion, and ankle inversion at the beginning of contact ground were reported than that of the NRS.

Peak MTPJ dorsiflexion angle and peak MTPJ dorsiflexion velocity of IRS during running were significantly smaller than that of NRS (all $p<.001$) (**Table 4.2**). In the frontal plane, there was no significant difference between IRS and NRS regarding peak ankle eversion angle, and peak ankle eversion velocity. MTPJ range of motion (ROM) of IRS in the sagittal plane was significantly smaller in comparison with that of NRS($p<.001$) (**Table 4.2**), there were no effects of wearing experience on the ankle, knee, and hip ROM in this plane. Ankle ROM (In-eversion) showed no obvious difference between IRS and NRS.

The SPM analysis showed a significantly higher ankle dorsiflexion angle for IRS compared to NRS between 0% and 4% of the stance time ($p< .05$). There was a smaller internal rotation angle for IRS than that of NRS from 0% to 6% and 63% to 72% of stance time (both $p< .05$). No significant angle differences between shoe conditions were found around the hip joint (**Figure 23**).

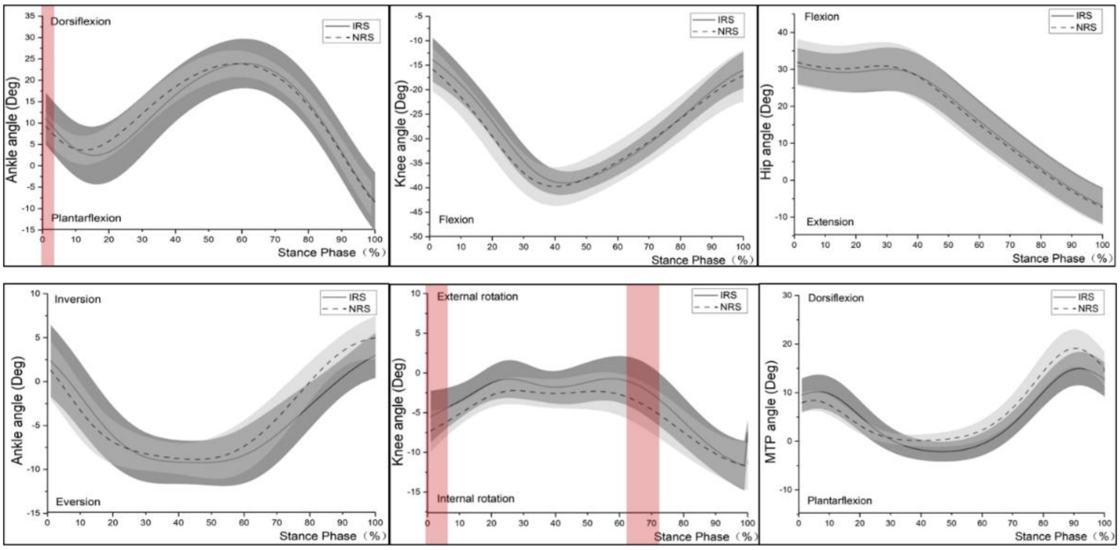


Figure 23. Lower limb joint angles time-normalized. Note: The red horizontal bars within the figure during corresponding periods represent significant shoe effects (SPM T-paired) between IRS and NRS.

Table 4.2. Mean values (±SD) for the main 0D parameters in IRS and NRS.

Variables	IRS	NRS	P	Cohen's d
Contact time (ms)	205.9±18.1	204.5±17.5	0.079	0.177
Braking phase (ms)	117.3±16.5	108.1±9.4	0.019*	0.264
Push-off phase (ms)	90.9±15.5	96.5±11.8	0.061	0.061
Step frequency	187.8±9.2	185.7±9.1	0.410	0.529
Step length(m)	2.18±0.07	2.17±0.06	0.400	0.476
Average loading rate 1 (BW/s)	78.4±20.6	99.9±24.4	0.005*	0.606
Peak loading rate 1 (BW/s)	106.7±35.6	169.4±32.2	0.000*	0.850
Peak vertical force 1 (BW)	1.93±0.27	2.12±0.27	0.002*	0.548
Time to peak vertical force 1 (ms)	39.1±11.0	28.9±5.0	0.001*	0.560
Peak braking force (BW)	0.42±0.08	0.46±0.07	0.003*	0.362
MTPJ peak plantarflexion moment (Nm/kg)	1.56±0.27	1.80±0.37	0.047*	0.298
MTPJ peak dorsiflexion angle (°)	16.2±5.5	20.6±3.8	0.002*	0.273
MTPJ ROM in the sagittal plane (°)	18.0±2.7	19.7±3.0	0.000*	0.551
MTPJ peak dorsiflexion velocity (°/sec)	352.9±48.2	396.0±55.5	0.000*	0.448
MTPJ negative work in the sagittal plane (J/kg)	0.06±0.02	0.07±0.03	0.383	0.071
MTPJ positive work in the sagittal plane(J/kg)	0.004±0.001	0.005±0.002	0.172	0.133
Ankle negative work in the sagittal plane (J/kg)	0.44±0.07	0.46±0.08	0.225	0.267
Ankle positive work in the sagittal plane(J/kg)	0.46±0.07	0.46±0.07	0.267	0.225
Peak ankle plantarflexion moment (Nm/kg)	2.36±0.27	2.21±0.24	0.196	0.187
Peak knee flexion moment (Nm/kg)	2.84±0.31	2.69±0.44	0.168	0.128
Ankle dorsiflexion angle at contact (°)	11.8±5.2	9.4±3.7	0.023*	0.546
Peak ankle eversion angle (°)	11.0±4.6	10.2±3.3	0.296	0.097
Peak ankle eversion velocity (°/sec)	299.9±92.9	292.5±51.8	0.557	0.557

** Showed significant effect between IRS and NRS. GRF were normalized to body weight (BW).*

4.3.2 Kinetics

There was no significant effect between IRS and NRS on contact time and push-off phase, yet the braking phase and the time of peak vertical force 1 of IRS were found longer than that of NRS (p=.019. p=.001). It induced a certainly lower average vertical loading rate 1(95%CI [IRS:67.97 to 88.83. NRS:87.55 to 112.25]), peak vertical loading rate 1(95%CI [IRS:88.68 to 124.72. NRS:153.10 to 185.70]) (both p<.001). Meanwhile, a lower peak braking force (95%CI [IRS:0.38 to0.46. NRS:0.42 to0.50]) and peak vertical force 1(95%CI [IRS:1.79 to 2.07BW. NRS:1.95 to 2.26BW]) were present in the IRS in comparison to the NRS during running (both $p<.001$) (**Figure 25**). There was no shoe effect on step frequency and step length ($p=.410$. $p=.400$). A lower peak MTPJ plantarflexion moment (95%CI [IRS:1.42 to 1.70Nm/kg. NRS:1.61 to 1.99Nm/kg]) was found for IRS compared to NRS ($p=0.047$). Meanwhile, no significant difference was found in the joints' work and peak moment (ankle and knee) in the sagittal plane (**Table 8**).

The SPM test exhibited a significant effect of shoes on the vertical and anteroposterior components of GRF (both P＜.001). The IRS decreased vertical GRF from 11% to 17% of the stance phase ($p = 0.009$) (**Figure 23-A**) and decreased braking anteroposterior GRF from 22% to 27% of the stance phase ($p = .043$) compared to the NRS (**Figure 23-B**). No significant moment differences between shoe conditions were reported at the ankle, knee, and hip levels (all p＞.05) (**Figure 24**).

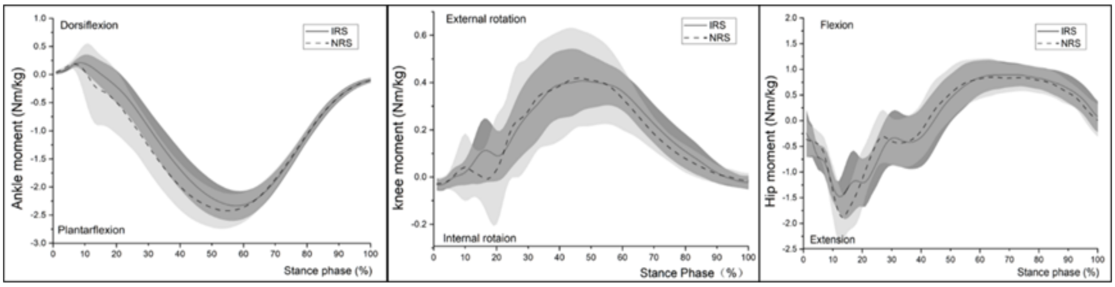


Figure 24. Lower limb joint moment time- and weight-normalized.

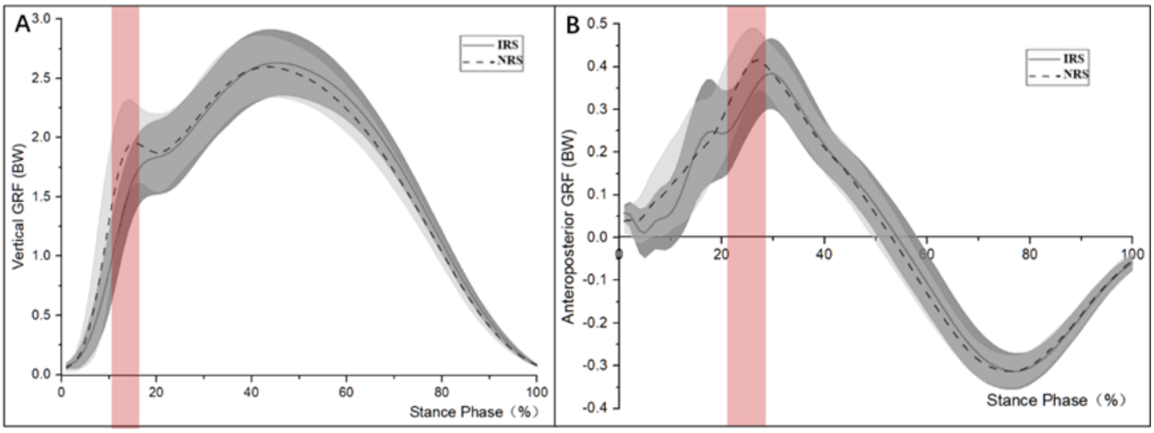


Figure 25. Mean vertical and anteroposterior ground reaction force-time- and weight-normalized. Positive and negative values were braking and propulsive forces. Standard deviations were presented by white and gray shaded areas. The red horizontal bars within the figure during corresponding time periods represent significant shoe effects (SPM T-paired) between IRS and NRS.

4.3.3 Subjective Perception

Runners didn't figure out the apparent difference between the two kinds of shoe conditions about shoe weight, arch support, fit, stability, and over preference. Still, most of them think that the cushioning of IRS was significantly better than that of NRS (**Figure 26**).

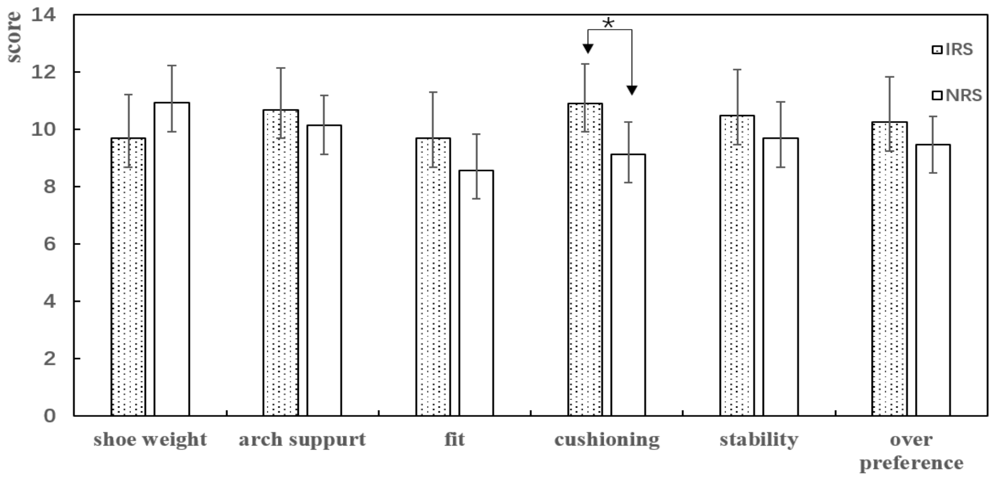


Figure 26. Mean and standard deviations for subjective perception were displayed (higher value, better performance). Note: * showed significant effect between IRS and NRS.

Discussion

Authors should discuss the results and how they could be interpreted from previous studies and the working hypotheses. The findings and their implications should be discussed in the broadest context possible. Future research directions may also be highlighted. This study was targeted to clarify the consequence of IRS towards female runners and compare the bio-mechanical difference between these kinds of running shoes during the stance phase. To get the LBS of forefoot and peak acceleration of rearfoot by mechanical testing, and then on the biomechanical characteristics while running at 3.6m/s.

According to the first hypothesis, the IRS would improve the shoe cushioning by increasing the HTD with three kinds of layers of shoe midsole during landing. In line with this behavior theory, the mechanical testing indicated a lower peak acceleration in the IRS (9.9g) compared to the NRS (13.7g), which induced significantly biomechanical changes particular during the braking phase, like a longer braking phase and the time to peak vertical force 1. Meanwhile, a certainly lower vertical force transient (average and peak) and peak vertical force 1 were present in the IRS in comparison to the NRS during running (all $p<.001$). The SPM test exhibited that the IRS decreased vertical GRF from 11% to 17% of the stance phase ($p = 0.009$) (Figure 4-A) and decreased braking anteroposterior GRF from 22% to 27% of the stance phase ($p = .043$) compared to the NRS (Figure 4-B). All the results were consistent with previous studies [160]. It was reported that the vertical loading rate and the associated transient peak increased when shoe drop decreased. In other words, increasing the shoe drop was the benefit of improving the cushioning of the running shoes.

As for the braking force, a reduced peak braking force was present in the IRS in comparison to the NRS during running. Some authors argue that peak braking force (higher value) was associated with the risk of injury hazards such as iliotibial band syndrome and should be considered a target for gait retraining interventions [174]. In other words, IRS in this study could effectively reduce the risk of lower extremity injuries than that of the NRS [149, 174, 175].

During the braking phase, the SPM analysis also showed a significantly higher ankle dorsiflexion angle for IRS compared to NRS between 0% and 4% of the stance time, which was consistent with previous reports [174, 176]. Horvais and Samozino [177] confirmed a positive correlation between shoe drop and shoe ground angle at touchdown in rearfoot runners. For example, the foot/ground angle at touchdown increased when the shoe drop increased.

Besides, the study exhibited no obvious difference in the shoe ground angle at touchdown, which was not consistent with previous reports [173, 176]. The reason was that the relationship between shoe ground angle and the vertical GRF parameters was complex [173]. For example, a rearfoot strike was not always linked to the appearing two kinds of impact peak.

It should be pointed out that the stability of the shoes was a particularly critical issue when increasing the HTD over 12mm. In this paper, there was no obvious difference in running posture at touchdown between the two kinds of shoes except the ankle of IRS had a more dorsiflexed. In addition, there was no significant difference between IRS and NRS in some stability parameters such as peak ankle eversion angle, peak ankle eversion velocity, and ankle ROM in the frontal plane in this study. This may be related to the ability of female runners to adapt to high heels, and also induced by the special hardness composition of the midsole in IRS. From subjective perception perspectives, runners didn't figure out the obvious difference between the two kinds of shoe conditions about stability. Still, most of them think that the cushioning of IRS was better than that of NRS which was consistent with our biomechanical results. What's more, there was a smaller knee internal rotation angle for IRS than that of NRS from 0% to 6% and 63% to 72% of stance time (both $p < .05$) (Figure 23). It was associated with the structure of the midsole with three layers (up and lower layers were for cushioning, the middle layer was for support), which played the role of motion control. Motion-control shoes were beneficial to reduce knee internal rotation in runners with over-pronation, no matter whether fresh or fatigued, which may assist pronated runners in maintaining their stability throughout fatiguing running [178, 179].

The mechanical testing proved larger LBS in the IRS (0.307±0.01Nm/deg) compared to the NRS (0.169±0.01Nm/deg), which significantly modified the running biomechanics during running. Several authors figured out that increasing the LBS of running shoes might induce a series of biomechanical changes such as a decreased MTPJ range of motion, a lower peak MTPJ dorsiflexion angle, peak MTPJ dorsiflexion velocity, and a lower peak MTPJ plantarflexion moment [54, 55, 66, 68, 69], those results were line with this study. In other words, IRS could improve the running performance by increasing the LBS compared to that of NRS. From the joint's moment view, as previously shown that a higher net flexion moment on the knee joint or ankle with a higher heel-to-toe offset might increase strain around the joint [180]. But a 16mm shoe drop didn't cause significant changes in joints (knee and ankle) torque in this study which would be associated with the three layers of the midsole to modify the stress around the joints (ankle and knee). This suggested that it was an important to combine the HTD and hardness of the midsole into account when designing a shoe to improve the cushioning and reduce the risk of injuries.

Special running shoes had two key points: the heel-to-toe drop of 16mm, and the second was the special hardness component of the midsole. It was the first time to explore the mechanism of shoe drop reaching 16mm on running biomechanical. Compared with normal running shoes, special running shoes in this study could effectively improve the cushioning and propulsion performance, but the stability had not changed significantly. Besides, the special shoe with a 16mm shoe drop didn't cause significant changes in joints (knee and ankle) torque in this study. This research adds new insight into the mechanism of shoe drop on the runner. Referring to the limitations, the experiment shoes differed in heel-to-toe offset height and shoe properties. Future investigations should only modify shoe drop or hardness components of the midsole. Besides, tendon stiffness might be associated with the regular use of high-heeled shoes in some females, leading to higher hypertrophy and shortening of the Achilles tendon, a higher pre-activation amplitude of the peroneal muscle [55], and greater gluteus maximus muscle activation [106]. Besides, motion-control shoes prevent exacerbated fatigue-related increases in pronated female runners [179]. It was also extremely important to take the muscle activation of the lower limb and fatigue during running into account when wearing a running shoe with the midsole's three-layer and 16mm shoe drop.

4.4 Conclusion

To research the effects of female running shoes (high heel-to-toe drop and special structure of midsole) on the biomechanics of the lower limbs and perceptual sensitivity in female runners. In this study, fifteen healthy female runners were recruited to run through a 145m runway with planted force plates at one peculiar speed (3.6m/s±5%) with two kinds of shoe conditions (Female running shoes: a 16mm heel-toe drop with three layers of the midsole, Normal running shoes: regular jogging shoe) while getting biomechanical data. Perception of shoe characteristics was assessed simultaneously through a 15 cm visual analogue scale. The Statistical Parametric Mapping technique calculated the time-series parameters. Regarding 0D parameters, the ankle dorsiflexion angle of female running shoes at touchdown was higher, the peak dorsiflexion angle, range of motion, peak dorsiflexion velocity, and plantarflexion moment on the metatarsophalangeal joint of female running shoes during running were significantly smaller than that of normal running shoes (all $p<.001$). In addition, the braking phase and the time of peak vertical force 1 of female running shoes were found longer than that of normal running shoes (both $p<.05$). Meanwhile, a lower average vertical loading rate 1, peak vertical loading rate 1, peak braking force, and peak vertical force 1 were present in the female running shoes compared to the normal running shoes during running (both $p<.01$). The Statistical Parametric Mapping analysis exhibited a higher ankle dorsiflexion angle for female running shoes compared to normal running shoes between 0% and 4% of the stance time ($p< .05$). Smaller knee internal rotation angle for female running shoes from 0% to 6% and 63% to 72% of stance time (both $p< .05$) were also found than that of normal running shoes. Besides, the female running shoes reduced vertical ground reaction force from 11% to 17% of the stance phase ($p = 0.009$) and braking anteroposterior ground reaction force from 22% to 27% of the stance phase ($p = .043$) compared to the normal running shoes. Runners were able to perceive that the cushioning of female running shoes was better than that of normal running shoes. Compared with normal running shoes, female running shoes in this study could effectively improve the cushioning and propulsion performance and played the role of motion control. It would benefit the industrial utilization of shoe producers in light of reducing the risk of running injuries for female runners.

5 Conclusion and further work

Conclusion

Increasing the LBS of running shoes through carbon plates requires consideration of the characteristics of marathon sports, such as the stress problem of the forefoot sole after long-distance running. It could be optimized by changing the shape of the carbon plates without affecting sports performance. From the view of stiff material chosen to increase the LBS of running shoes, especially when designing running shoes for female runners, it was necessary to consider their physiology, sports characteristics, and wearing habits. When designing running shoes for female runners who were used to wearing high heels, it was more important to focus on improving the cushioning of running shoes by increasing the thickness of the midsole, special in the rearfoot area. The center of gravity of the running shoe would become higher with the increase of the heel thickness, which would affect the dynamic stability of the running. So, combining a 16mm HTD and three layers of the midsole (up and lower layers for cushioning, the middle layer for support) improves the rearfoot's stability and enhances the running performance.

The ideal with a 16mm HTD and three layers of the midsole won the Contemporary Good Design Award in 2020. The Contemporary Good Design Award (CGD) was an international design award organized by the German Red Dot Award organization. In contrast, the red dot design museum Essen was the primary support of the red dot product design award and the red dot brand and communication design award, the red dot design museum Singapore was the primary support of the red dot design concept award, and the red dot design museum Xiamen was the primary support of the recent good design award. In 2015, the German Red Dot Award organization and Xiamen Media Group launched the Contemporary Good Design Award. Red Dot was responsible for the organization of the international jury and the selection of works, using more than 60 years of experience in operating top international design awards and design resources to ensure the professionalism, seriousness, and authority of the Contemporary Good Design Award. The products that stand out would be honored to recognize their outstanding design achievements and receive professional services from the winners.

Direction for further studies

As described, compared to normal running shoes with a 11.5mm HTD, the shoes combining a 16mm HTD and three layers of the midsole (up: 40C, middle: 50C, under: 40C) was benefit to improve the rearfoot's stability and enhances the running performance. For runner, it should be taking the muscle activation of the lower limb and fatigue during running into account when wearing it, thus, it is important to add EMG and endurance test to future test validation.

From the view of extensibility study, building four kinds of model (A: a 16mm HTD and three layers of the midsole (up: 40C, middle: 50C, under: 40C), B: a 16mm HTD and one layer of the midsole (50C), C: a 16mm HTD and two layers of the midsole (up: 40C, middle: 50C, under: 50C), D: a 16mm HTD and two layers of the midsole (up: 50C, middle: 50C, under: 40C)) is benefit to investigate the independent effect of the hardness of the midsole on running biomechanics and finite element simulation.

From the view of innovation study, combining the animal bionic research and this shoes with a16mm HTD and three layers of the midsole (up: 40C, middle: 50C, under: 40C) that could relatively counter the adverse effect of high heel and drop on the human musculoskeletal system, such as, meat pad of animal scratch (special morphological structure allows animal to absorb two to three times their body weight while resting on their small distal joint). In other words, Adjusting the structure of forefoot of this innovation shoes by design the bionic structure of the animal's foot grasp may provide valuable information.

New scientific thesis points

1st thesis point: I investigated the independent effect of forefoot carbon-fiber plate, inserted into the midsole, and I deduced the following scientific results:

- By inserting the carbon-fiber plate, I could increase the peak plantarflexion angular velocity on the metatarsophalangeal joint by 20% with fast speed. This increased angular velocity will be benefit to propulsion phase during running for runner.
- I deducted that if a carbon-fiber plate is used only in the midsole, not a full forefoot plate, then this configuration will result in increased positive work at the knee joint by 9%, and a reduction of maximum pressure on the midsole by 29%. It could reduce the risk of injuries such as metatarsal stress fractures and plantar fasciitis by decreasing stress on the metatarsal region, especially the second metatarsal region
- Based on my numerical and experimental results, I concluded that changing the shape of the carbon plate does not affect the running performance (the difference is less than 1%). Therefore, the simplest shape can be chosen for use since the biomechanical parameters will not be affected while the manufacturers can inexpensively produce these plates.

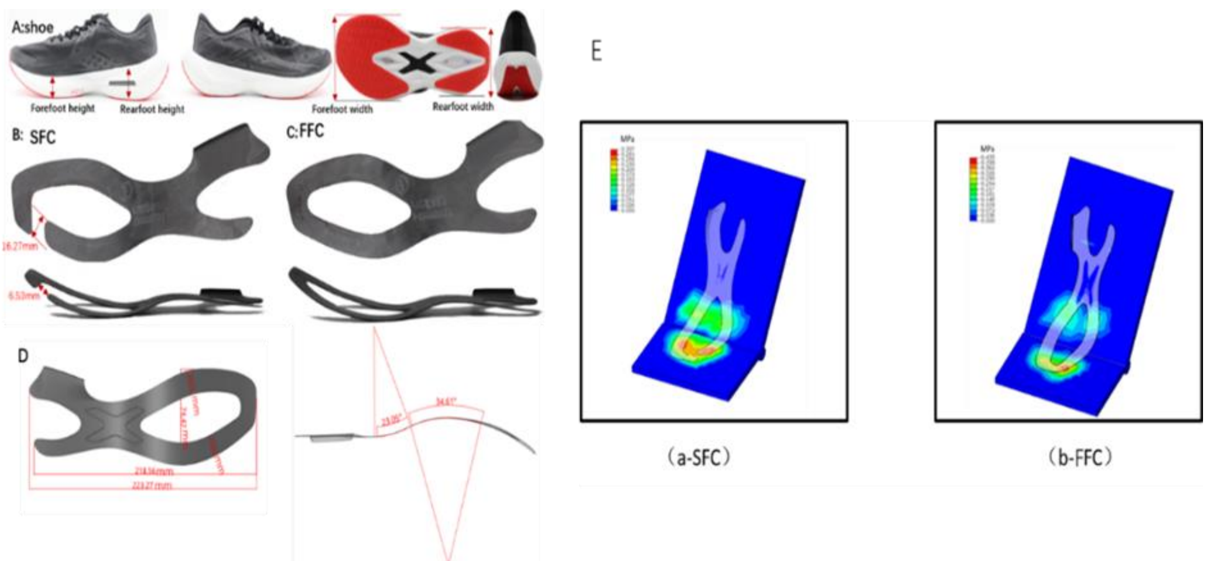


Figure 27. (A): Experiment shoe (Forefoot height: vertical thickness at 12% of external length, Rearfoot height: vertical thickness at 75% of external length, offset: offset = rearfoot height − forefoot height); (B): the forefoot area of carbon fiber plate (carbon fiber plate was made up of 63 % carbon fiber and 37% epoxy resin fiber) was designed to a segment construction inserted to midsole (SFC), (C): the forefoot area of carbon fiber plate was designed to a full construction inserted to midsole (FFC)); (D): the information about the geometry and dimensions of the carbon plates; (E): The pressure on the SFC and FFC during bending.

Related articles to the first thesis point:

[1]**Fu, F**., Levadnyi, I., Wang, J., Xie, Z., Fekete, G., Cai, Y., & Gu Y. (2021). Effect of the Construction of Carbon Fiber Plate Insert to Midsole on Running Performance. MDPI: Materials, 14, 51-56. **IF: 2.623, Q2**

2st thesis point: I investigated the biomechanical differences of people who are experienced and inexperienced (IEW) in wearing high-heel shoes. The main reason was to identify the most influencing common parameters and those which are different between the two groups.

- I experimentally deduced that both groups, experienced moderate high-heel wearers and inexperienced moderate high-heel wearers, regulate their stride speed through stride length. This is the common parameter.
- I also experimentally deduced that the experienced moderate high-heel wearers group had higher GRF results by 11%, which means that these people are subjected to 31% higher vertical average loading rate than the inexperienced moderate high-heel wearers group. Due to this fact, this group is more exposed to the risk of injury in the musculoskeletal system of the lower limbs. This is a different parameter.
- Furthermore, I also experimentally deduced that the inexperienced moderate high-heel wearers group use 12% higher stride frequency to regulate the stride speed. This is a different parameter, which helps to compensate for the reduced balance.
- As a conclusion, in the design of running shoes for EW should pay more attention to the cushioning of rearfoot while enhancing the stability of shoes during running for IEW was a curial point for manufacturers.

Related articles to the second thesis point:

[1]**Fu, F**., Zhang, Y., Shu, Y., Gu, Y. (2016). Lower-limb mechanics during moderate high-heel jogging and running in different experienced wearers. Human movement science, 48, 15-27. **IF:1.955, Q3**

[1]**Fu, F.Q**., Wang, S., Shu, Y., et al., A Comparative Biomechanical Analysis the Vertical Jump between Flatfoot and Normal Foot. Journal of Biomimetics, Biomaterials and Biomedical Engineering. Trans Tech Publications, 2016. 28, 26-35. **Q4**

3rd thesis point: I investigated the effects of innovative running shoes (high heel-to-toe drop and special structure of midsole) on the biomechanics, and I deduced the following scientific results:

- By choosing the stiff material to increase the longitudinal bending stiffness of innovative running shoes, I could decrease the metatarsophalangeal joint range of motion by 9%, and peak metatarsophalangeal plantarflexion moment by 13%. These biomechanical changes will be benefit to greatly reduce the work by metatarsophalangeal joints and improve the economy of running.
- I deducted that increasing the height of the rearfoot to 32mm and offset to 16mm, then this configuration will result in a reduced peak braking force by 9%, average vertical force transient by 22%, peak vertical force transient by 37%, and peak vertical force by 9% during running. It could be benefit to release ground impact force on musculoskeletal tissue which is related to reduce risks of running injuries.
- Adjusting the hardness composition of the midsole in innovative running shoes into three layers of the midsole (up and lower layers were for cushioning: 40Asker C, the middle layer was for support: 50Asker C), induced 23% smaller knee internal rotation angle while it did not cause significant changes (difference is less than 1%) at the joint (knee and ankle) torques.

Based on my numerical and experimental results, I concluded that innovative running shoes would benefit the industrial utilization of shoe producers in the light of reducing the ground impact force and strengthening the running economy by decreasing the metatarsophalangeal joint work for experienced moderate high-heel wearers.

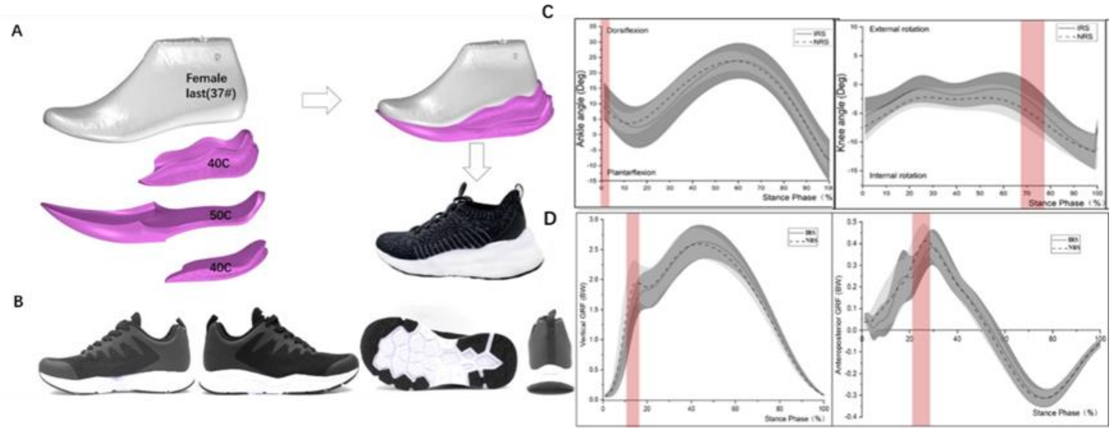


Figure 28. (A): Picture of the special running shoes (IRS) prototype used during running; (B): Normal running shoes (NRS); (C): Lower limb joint angles time-normalized; (D): Mean vertical and anteroposterior ground reaction force-time- and weight-normalized. Positive and negative values were braking and propulsive forces. White and gray shaded areas present standard deviations. Note: The red horizontal bars within the figure during corresponding time periods represent significant shoe effects (SPM T-paired) between IRS and NRS.

Related articles to the third thesis point*:*

[1]**Fu, F.Q.**, Guo, L.M., Tang, X.F., Wang, J.Y., Xie, Z.H., Fekete, G., Cai, Y.H., Hu, Q.L., Gu, Y.D., Effect of the innovative running shoes with the special midsole structure on the female runners ‘lower limb biomechanics. Frontiers in Bioengineering and Biotechnology, 2022, 11. **IF: 5.48, Q1**

Isherwood, J., Rimmer, E., [3]**Fu, F.Q**., et al. Biomechanical and Perceptual Cushioning Sensitivity based on Mechanical Running Shoe Properties. Footwear Science, 2021. (5), 221-231. **IF： 0.6, Q3**

List of all publications

1. **Fu, F.,** Levadnyi, I., Wang, J., Xie, Z., Fekete, G., Cai, Y., & Gu Y. , Effect of the Construction of Carbon Fiber Plate Insert to Midsole on Running Performance. MDPI: Materials, 2021. 14, 51-56. **IF: 2.623**

2. Isherwood, J., Rimmer, E., **Fu, F.Q.,** et al. Biomechanical and Perceptual Cushioning Sensitivity based on Mechanical Running Shoe Properties. Footwear Science, 2021. (5), 221-231. **IF：0.6**

3. **Fu, F.Q.**, Wang, S., Shu, Y., et al., A Comparative Biomechanical Analysis the Vertical Jump Between Flatfoot and Normal Foot. Journal of Biomimetics, Biomaterials and Biomedical Engineering. Trans Tech Publications, 2016. 28, 26-35.

4. **Fu, F.**, Zhang, Y., Shao, S., et al., Comparison of center of pressure trajectory characteristics in table tennis during topspin forehand loop between superior and intermediate players. International journal of Sports Science & Coaching, 2016. 11(4), 559-565.

5. **Fu, F.**, Zhang, Y., Shu, Y., et al., Lower limb mechanics during moderate high-heel jogging and running in different experienced wearers. Human movement science, 2016. 48, 15-27. **IF: 1.955**

6. Mei, Q., Gu Y., **Fu, F.**, et al., A biomechanical investigation of right-forward lunging step among badminton players. Journal of sports sciences, 2016. 1-6. **IF: 2.713**

7. Fu, F.Q., Guo, L.M., Tang, X.F., Wang, J.Y., Xie, Z.H., Fekete, G., Cai, Y.H., Hu, Q.L., Gu, Y.D., Effect of the innovative running shoes with the special midsole structure on the female runners' lower limb biomechanics. Frontiers in Bioengineering and Biotechnology, 2022, 10, 866321. **IF: 5.48**

8. Wang, M., Fu, L., Gu, Y., Mei, Q., **Fu, F**., Fernandez, J. (2018). Comparative study of kinematics and muscle activity between elite and amateur table tennis players during topspin loop against backspin movements. Journal of Human Kinetics, 64, 25-33.

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ABBREVATION

R&D -research and development

LBS-longitudinal bending stiffness

CFP-carbon fiber plate

MTPJ-metatarsophalangeal joint

LBS- forefoot bending stiffness

3D-three dimensional

ACL- Anterior Cruciate Ligament

CMJ-Counter Movement Jump

VF-Nike Vaporfly 4%

FFC-full forefoot plate construction

SFC-segmented forefoot plate construction

FE-finite element

CAD-Computer Aided Design

ASIS-anterior superior iliac spine

PSIS-posterior superior iliac spine

SPSS-Statistical Product and Service Solutions

RM ANOVA-repeated-measures analysis of variance

SPM-Statistical Parametric Mapping

BW-body weight

EW-experienced moderate high-heel wearers

IEW-inexperienced moderate high-heel wearers

SL-Stride length

SF-Stride frequency

ST/GC-Stance phase percentage

ROM-range of motion

VALR-vertical average loading rate

RFS-rearfoot strike

HTD-heel-toe drop

TRS-special running shoes

NRS-normal running shoes

VAS-visual analogue scale

1D-one-dimensional

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gray shaded areas present standard deviations. The red horizontal bars within the figure during corresponding time periods represent significant shoe effects (SPM T-paired) between IRS and NRS.

Figure 26. Mean and standard deviations for subjective perception were displayed (higher value, better performance). Note: * showed significant effect between IRS and NRS.

Figure 27. (A): Experiment shoe (Forefoot height: vertical thickness at 12% of external length, Rearfoot height: vertical thickness at 75% of external length, offset: offset = rearfoot height − forefoot height); (B): the forefoot area of carbon fiber plate (carbon fiber plate was made up of 63 % carbon fiber and 37% epoxy resin fiber) was designed to a segment construction inserted to midsole (SFC), (C): the forefoot area of carbon fiber plate was designed to a full construction inserted to midsole (FFC)); (D): the information about the geometry and dimensions of the carbon plates; (E): The pressure on the SFC and FFC during bending.

Figure 28. (A): Picture of the special running shoes (IRS) prototype used during running; (B): Normal running shoes (NRS); (C): Lower limb joint angles time-normalized; (D): Mean vertical and anteroposterior ground reaction force-time- and weight-normalized. Positive and negative values were braking and propulsive forces. White and gray shaded areas present standard deviations. Note: The red horizontal bars within the figure during corresponding time periods represent significant shoe effects (SPM T-paired) between IRS and NRS.

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