ÓBUDAI EGYETEM
ÓBUDA UNIVERSITY

DOCTORAL (PhD) THESIS FOR HOME DEFENSE

XINYAN JIANG

Biomechanical Alteration in Response to Long-Distance Running, Running Experiences, Speed and the Gender of Runners

Supervisor: Prof. Dr. Bíró István

DOCTORAL SCHOOL ON SAFETY AND SECURITY SCIENCES

2024/11/19

Content

Motivation of the work

Running is recognized as one of the most prevalent forms of physical activity worldwide, with a significant increase in participation over the past few decades. Due to its accessibility, minimal cost, and ease of implementation, running is frequently adopted by individuals seeking to improve health outcomes, such as weight management and enhanced physical fitness [1,2].

Studies have demonstrated that running not only enhances physical function but also effectively promotes mental well-being of runners [3]. Despite the positive health impacts of running, the sharp increase in participation has led to a corresponding rise in running-related injuries (RRIs) [4,5]. During long-distance running, runners are subjected to vertical ground reaction force (GRF) equivalent to two to three times their body weight (BW) [6]. As a result, they repeatedly experience the impact of vertical GRF. Reports on RRIs indicate that the incidence rate of such injuries ranges from 30% to 79% [5,7]. A majority of these RRIs (50%-75%) are attributed to overuse of the knee joint and areas below it, with the knee and ankle being the most commonly affected regions [8].

The factors influencing RRIs are multifaceted, encompassing both intrinsic and extrinsic factors. Intrinsic factors include biomechanical and morphological differences among runners, as well as age, gender, medical history, and body mass index (BMI). Extrinsic factors involve training experience, physical fitness, type of running shoes, and other athletic equipment [9-13]. Despite significant efforts by clinicians and researchers to reduce the incidence of RRIs, alongside continuous advancements and innovations in running gear such as shoes, the injury rate has not declined over the past 40 years [4,8,14]. Research indicates that novice runners with no prior running experience are at higher risk of sustaining RRIs [9,15]. Therefore, it is particularly important for novice runners to focus on injury prevention during running, as this can enhance their long-term participation and contribute to the promotion of public health.

Sex-specific anatomical variations are widely recognized to affect lower extremity kinematics during running, particularly in parameters such as hip adduction, hip internal rotation, and knee abduction [16,17]. Female runners typically exhibit a greater range of motion in both the frontal and transverse planes when compared to male runners. These differences are largely attributed to the distinctive morphology of female runners, including a higher hip-width to femoral length ratio, which may play a role in the differential risk of RRIs. Abnormal movement mechanics, often cited as a contributing factor to injury, also differs between sexes. Female runners, in contrast to males, show increased hip internal rotation and adduction, along with greater peak knee abduction, all of which may contribute to a heightened susceptibility to injury. Biomechanical differences in the lower limbs between male and female runners can impact running economy, affecting energy efficiency and performance [18]. A comprehensive understanding of the kinetic and kinematic differences between male and female runners may provide insights into sex-specific injury rates and patterns. Considering these biomechanical variations can enhance the effectiveness of injury prevention strategies.

As running speed increases, the magnitude of forces acting on the body also rises. Studies have documented changes in GRFs, joint moments, muscle activity, leg stiffness, and body segment motions at varying running speeds [19]. Understanding the biomechanical behavior of the lower limbs across different speeds is critical for advancing knowledge of human performance and identifying factors contributing to injury. Higher running speeds amplify the forces transmitted through the lower extremities. At slower speeds, stride length and contact time decrease while step frequency increases, potentially allowing more time for force dissipation upon ground impact [20]. In controlled overground conditions, higher speeds result in shorter contact times and greater peak forces [21]. Consequently, reducing running speed may serve as an effective strategy to lower biomechanical load. The positive correlation between ground reaction force and running speed is well established in the literature [22,23]. Based on the above findings, this dissertation further aims to explore strategies to reduce running injury rates and prevent lower limb injuries during long-distance running, thereby providing meaningful guidance for the practice of running and the prevention of RRIs.

Research objectives

Based on the work motivation, the main research purposes of this dissertation are as follows.

The first research objective: This study aims to develop musculoskeletal modeling and simulation techniques to compare muscle forces and knee reaction force between novice and experienced runners. Novice runners are defined as individuals who run between 2–10 km per week and do not participate in any formal running competitions or training programs. In contrast, experienced runners consistently run at least 30 km per week and have a minimum of three years of running experience. Although increased running experience is associated with a reduced risk of RRIs, the underlying biomechanical mechanisms remain unclear. Since recent advancements in musculoskeletal modeling and power computing, researchers have been allowed to develop motion simulations to value muscle forces, and then joint forces. Muscles reduce the bending stress on bones and dampen the peak dynamic loads from unprotected impulsive loads that can cause harm to musculoskeletal tissues. The knee muscle groups were the important contributors during running, due to the large amount of work those muscles generate. Information on this is especially pertinent to the fields of injury prevention and running performance. The objective of this research is to investigate the biomechanical differences between runners with varying levels of experience to enhance understanding of the factors that may contribute to reduced injury risk among runners.

The second research objective: To quantitatively explore the underlying mechanisms contributing to the development of RRIs during long-distance running, this study focuses on examining biomechanical changes in the lower extremities. Specifically, we aim to investigate alterations in joint angles and moments over the course of a 5-kilometer run in two distinct groups of runners: experienced and novice. Through a detailed analysis of these biomechanical variations, we seek to identify key risk factors that may increase the likelihood of injury. By focusing on the differences in joint loading patterns and kinematic behaviors between the two groups, this study will offer critical insights into the biomechanical triggers that lead to RRIs. Moreover, the findings will contribute to the development of evidence-based guidelines for safer, more effective long-distance practices. These guidelines will not only assist in reducing injury risks but also support long-term performance enhancement and the overall health of runners. Ultimately, the practical recommendations derived from this research will be applicable to both novice and seasoned runners, promoting injury prevention and facilitating healthier, sustained participation in running activities.

The third research objective: During running training sessions, speed is frequently adjusted as a key indicator of the task's physical intensity. While many studies have investigated the biomechanical effects of varying running speeds, the majority have focused on only one gender, leaving the gender-specific biomechanical responses to speed largely unexplored. The current study seeks to fill this gap by examining the differences in gait patterns between male and female runners across seven discrete running speeds: 10, 11, 12, 13, 14, 15, and 16 km/h. Specifically, the study will explore the relationship between GRFs and running speed in both genders. Understanding GRF variations across different speeds is crucial for identifying biomechanical factors that may contribute to RRIs. Despite the importance of this relationship, current research provides insufficient evidence regarding the confounding effects of running speed and gender on GRFs. Key questions remain unanswered, including whether males and females adapt differently to changes in running speed, how speed influences GRFs during overground running, and whether GRF parameters can reliably predict changes in running speed. Addressing these questions could lead to more effective injury prevention strategies and a deeper understanding of the biomechanical adaptations to varying speeds in runners.

1. LITERATURE REVIEW

1.1 Overview of running biomechanics

1.1.1 Gait cycle characteristics of running

Running is one of the most popular physical activities due to its convenience and the lack of need for specific equipment or venues. It offers a range of benefits to runners, including promoting physical and mental health, enhancing cardiovascular function, reducing psychological stress, and providing recreational enjoyment [2,3,24]. Evidence suggests that running provides significant health benefits in preventing chronic diseases and reducing premature mortality, regardless of a runner's gender, age, or health status. There are well-established physiological mechanisms underlying the health improvements and increased lifespan associated with running [25,26]. From a public health perspective, running may be the most cost-effective form of exercise. Additionally, from an evolutionary standpoint, medium- to long-distance running has been crucial for human survival and adaptation [27].

The gait cycle serves as the fundamental unit in gait analysis, encompassing the interval between the initial contact of one foot with the ground and its subsequent contact in the next stride [28,29]. In walking, the gait cycle is composed of two main phases: stance and swing. The stance phase, during which one foot is in contact with the ground, constitutes 60% of the cycle, while the swing phase, where the foot is off the ground, makes up the remaining 40%. During the stance phase, there are two periods of double support—each accounting for 10% of the cycle—where both feet are in contact with the ground. Single limb support coincides with the swing phase of the opposite leg.

The running gait cycle consists of two primary phases: stance and swing (as shown in Figure 1 and 2). However, unlike walking, the running gait cycle also includes a distinct float phase within the swing phase, which occurs twice during the gait cycle: once at the beginning of the initial swing and once at the end of the terminal swing, when both feet are airborne and not in contact with the ground. The stance phase can be subdivided into two parts: the first half is dedicated to force absorption (pronation), while the second half is focused on propulsion (supination). This phase is further divided into initial contact to midstance and midstance to toe-off. From a biomechanical perspective, the stance phase during running is often described in three distinct stages: (1) initial contact to foot flat, (2) foot flat to heel-off, and (3) heel-off to toe-off. The swing phase in running is divided into initial swing and terminal swing [30].

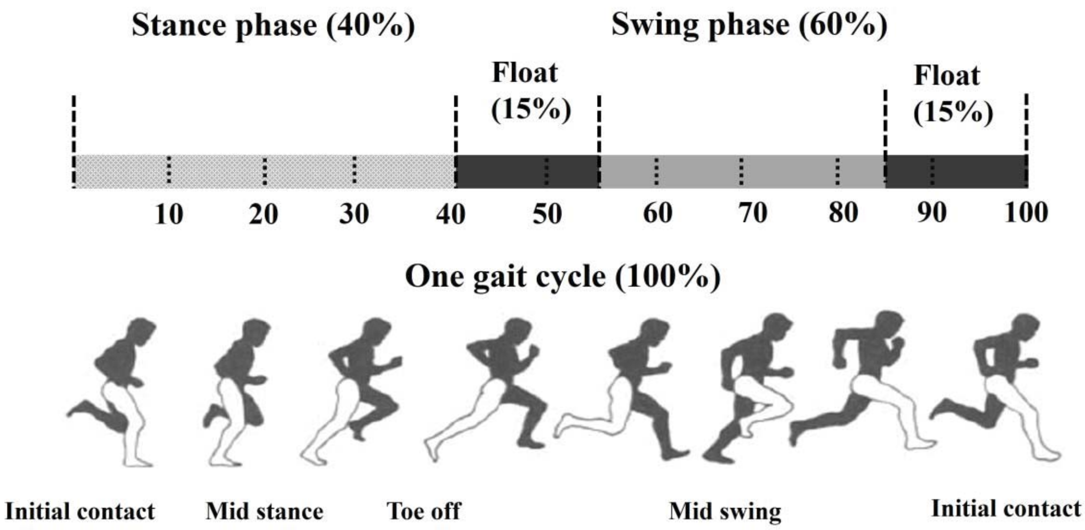

Figure 1 Overview of the running gait cycle

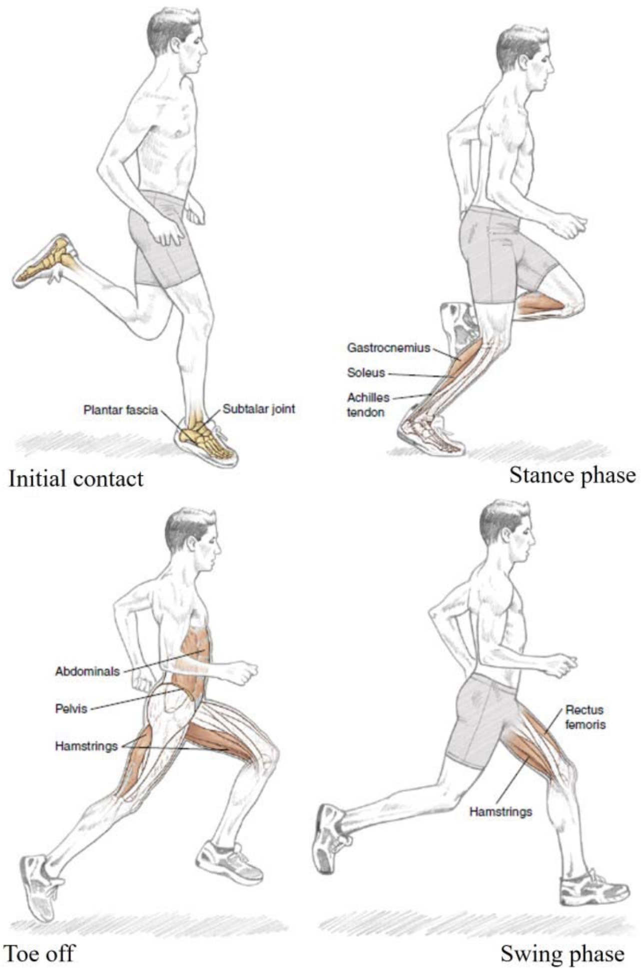

Figure 2 The gait cycle [31]

As running velocity increases, distinct alterations in gait characteristics can be observed. While the body's center of gravity follows a sinusoidal path in both walking and running, a key difference in running is that the body maintains a forward-leaning posture throughout the gait cycle. Several parameters are used to describe running speed and stride mechanics, including cadence, stride length, and step length. Cadence refers to the number of steps taken per minute, while stride length is the distance between successive initial ground contacts of the same foot. Step length, in contrast, measures the distance between the initial contact of one foot and the subsequent initial contact of the opposite foot. Temporal and spatial aspects of running gait are closely related. As speed increases, runners primarily extend their step length, followed by an increase in cadence to achieve higher speeds [32]. With greater running velocity, more time is spent in the float phase, during which both feet are airborne. Stride and step lengths are influenced by factors such as leg length and overall body height, and the ability to increase these lengths plays a significant role in enhancing running speed [28].

The stance phase has generally received more attention in the study of running performance and RRIs, as it is during this phase that the foot bears the forces generated by BW and GRFs [33]. Understanding the mechanics of the stance phase is critical for identifying factors that influence performance and contribute to injury risk, given the significant load the body experiences during this period of the gait cycle.

1.1.2 Kinematic characteristics of running

Running kinematics refers to the detailed characterization of motion parameters, including the position, velocity, and acceleration of the lower extremities throughout the running cycle. This analysis encompasses the assessment of gait patterns, the dynamic changes in the body's center of mass (COM) during locomotion, and the range of motion exhibited by various body segments during movement [34].

Kinematic analysis is conducted using advanced three-dimensional (3D) motion capture systems, which digitally model the human body as a multisegment structure [35]. By placing infrared markers on designated anatomical landmarks, cameras can triangulate the markers' positions, allowing for the calibration of the individual's body within the system. This enables the precise calculation of joint angles, joint angular velocities, and accelerations by determining the coordinates and orientation of the rigid body segments. Data are collected for each joint across all three cardinal planes of motion. Despite the potential influence of skin movement artifacts, current 3D motion capture technology remains the most reliable and noninvasive approach for obtaining accurate kinematic measurements [28,36,37].

As the body transitions from walking to running to sprinting, the COM is lowered and the body tilts in space shifted forward. The combined effect is to maximize the propulsion phase. In running, the hip peak range of extension is similar to that of walking; however, peak extension occurs at toe-off. Increased peak hip flexion is seen during running to advance the limb in swing. Overall hip flexion/extension and abduction/adduction mobility are increased in running. In running the hip must extend in the later part of swing so as to place the foot in the correct orientation under the body. If this did not occur, foot contact would be too far ahead of the COM and shift the GRFs posterior, thus causing deceleration [35]. Mediolateral movement during running is less pronounced than motion in the sagittal plane. When the runner is in midstance on the right foot, the right hip adducts, the opposite side of the pelvis lowers, and the lumbar spine slightly bends toward the right. This interconnected motion of the hip, pelvis, and lumbar spine serves to stabilize the trunk and head, helping to maintain balance and minimize unnecessary upper body movement during running [29].

The knee follows similar movement patterns in both walking and running, flexing during stance to absorb impact and again during swing to clear the limb. However, the degree of knee flexion during swing increases significantly from around 60° in walking to over 90° in running. In elite sprinters, knee flexion during swing can reach between 105° and 130°. During stance phase in running, the knee is flexed approximately 25° at initial contact and continues to flex, reaching a peak of about 45° at midstance. In sprinters, less peak flexion is observed during stance due to the reduced ground contact time, a key factor in maximizing sprinting efficiency.

Foot and ankle mechanics are essential components in gait analysis, particularly in running [38,39]. The foot and ankle function as a dynamic lever system, transitioning into an open-packed position (full pronation) to effectively absorb and dissipate shock upon ground contact, thereby preventing excessive shock transmission through the kinetic chain. During push-off, proper resupination of the foot is crucial to optimize force transfer. In running, the tibia is positioned more vertically than in walking, demanding greater ankle dorsiflexion to achieve initial contact. Following this, the ankle in walking plantarflexes to allow the foot to flatten, while in running, it dorsiflexes as the limb bears load. Due to shorter ground contact times, sprinters typically land on their forefoot, resulting in reduced peak dorsiflexion during midstance. During propulsion, sprinters display increased plantar flexion at toe-off and require less dorsiflexion for limb clearance during the swing phase, aided by increased knee mobility. As running speed increases, the timing of peak ankle values shifts earlier in the stance phase [35]. The anatomy of lower limb is illustrated in Figure 3 [31].

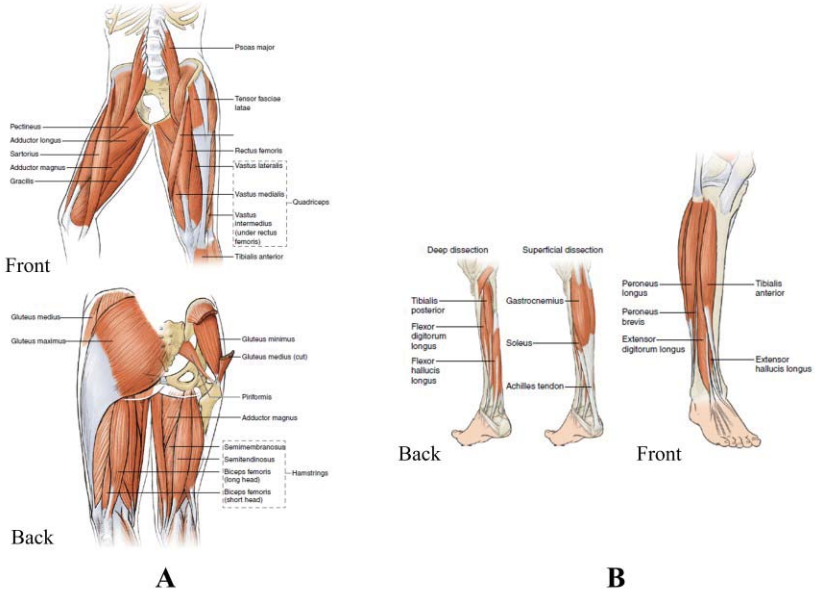

Figure 3 Anatomy of the lower limb: (A) Upper leg and (B) Lower leg and foot [31]

1.1.3 Kinetic characteristics of running

Kinetics in running refers to the study of forces that contribute to movement. It examines the external and internal forces, as well as the energy and power, that influence locomotion [40]. During the stance phase of the running cycle, the body experiences its highest mechanical load, known as the GRF (as shown in Figure 4) [41]. The center of pressure (COP) represents the origin of the force applied to the foot during this phase. By analyzing COP, GRFs, and joint kinematics, joint kinetics (e.g., joint moments) can be calculated, revealing the interactions between external forces such as GRFs, inertia, and gravity, and internal forces generated by muscles, tendons, ligaments, and bone structures that stabilize the joints. Joint power, derived from the rate of work performed by muscles, reflects the velocity at which joint moments are produced. While direct monitoring of kinetics typically requires a laboratory setting, an understanding of these biomechanical principles is critical for clinicians in assessing the mechanics of running and how they vary throughout the gait cycle [35].

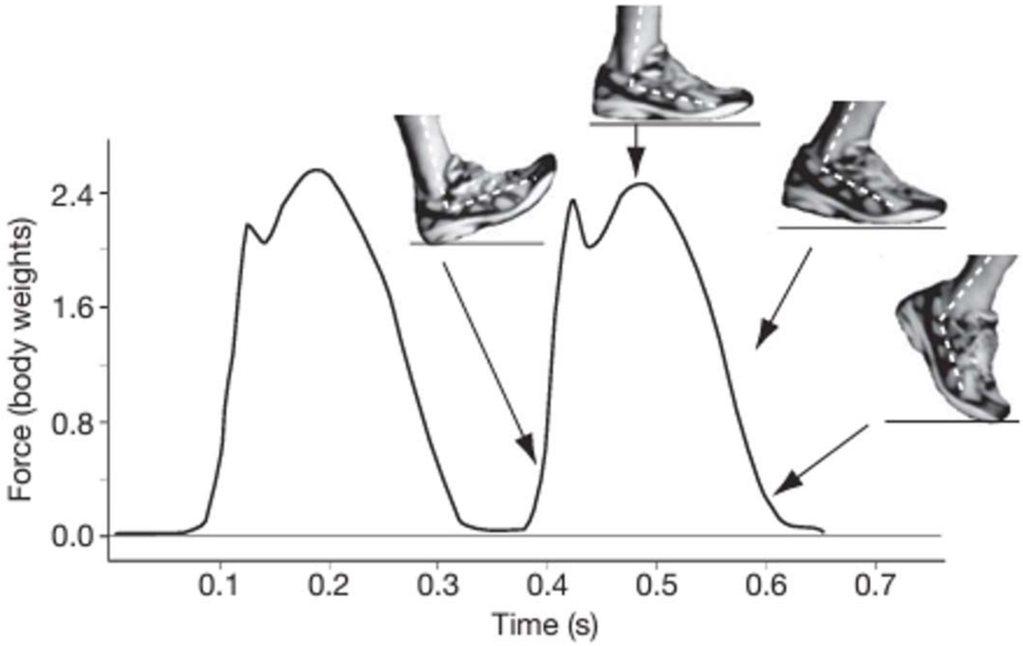

Figure 4 Vertical GRFs and foot kinematics during rear foot strike running [6]

In current research on running biomechanics, plantar pressure testing systems and 3D force platforms are primarily used to measure plantar pressure distribution and GRF data during the stance phase of running. The pressure exerted on the foot by external forces during the stance phase can be assessed through COP detection, allowing software analysis to evaluate the specific distribution of pressure across different regions of the foot. Nagel et al. [42] investigated the plantar pressure of 200 marathon runners before and after their races, revealing that peak pressure and impulse in the forefoot region significantly increased after running, while the pressure in the toe region decreased. Using a 3D force platform, it is possible to measure forces exerted on the foot in three directions: vertical GRF, anterior-posterior GRF, and medial-lateral GRF. During middle- to long-distance running, runners typically experience vertical GRF that is two to three times their BW [6,33]. For heel-strike runners, vertical GRF generates two peaks during the running cycle: the first peak occurs during heel contact with the ground, and the second during the push-off phase. Parameters such as peak GRF and vertical loading rate are closely associated with exercise-related injuries [43].

Joint moments and joint power are parameters derived from inverse dynamics, calculated by combining kinematic data with GRF. Tom [29] found that the ankle joint moment pattern during running is similar to that during walking, with plantarflexion moments occurring between 5% and 10% of the gait cycle. However, initial plantarflexion moments do not appear during sprinting. Ankle joint power is generated to provide propulsion during the gait cycle, and in sprinting, ankle power is typically greater than in long-distance running, with power magnitude directly related to running speed. Unlike the ankle, the knee joint moment patterns are similar in both sprinting and long-distance running. During the initial foot-ground contact, knee flexion moments increase, primarily due to the activation of the hamstrings, followed by a decrease in extension moments dominated by the quadriceps. The peak knee joint moments in long-distance running are greater than in sprinting, which is related to the knee flexion angle of the runner. After foot contact, hip extensors dominate, increasing the hip extension moment, followed by an increase in hip flexion moments due to the activation of hip flexors. Throughout the running cycle, both the hip extensors and flexors contribute to energy production. The kinetic parameters of the ankle, knee, and hip joints in the sagittal and coronal planes remain relatively stable, owing to the stabilizing function of the associated muscles and ligaments.

1.1.4 Muscle activity characteristics of running

Electromyography (EMG) is a technique that records the electrical signals generated during skeletal muscle excitation. In the field of biomechanics, surface EMG is the most commonly used tool. This sensor can collect real-time data during movement without interfering with the athlete's performance, offering the advantage of being non-invasive. Changes in EMG parameters can be used to assess muscle activation timing, amplitude, and fatigue during physical activities. EMG has been widely applied in the study of running biomechanics (as shown in Figure 5) [44,45].

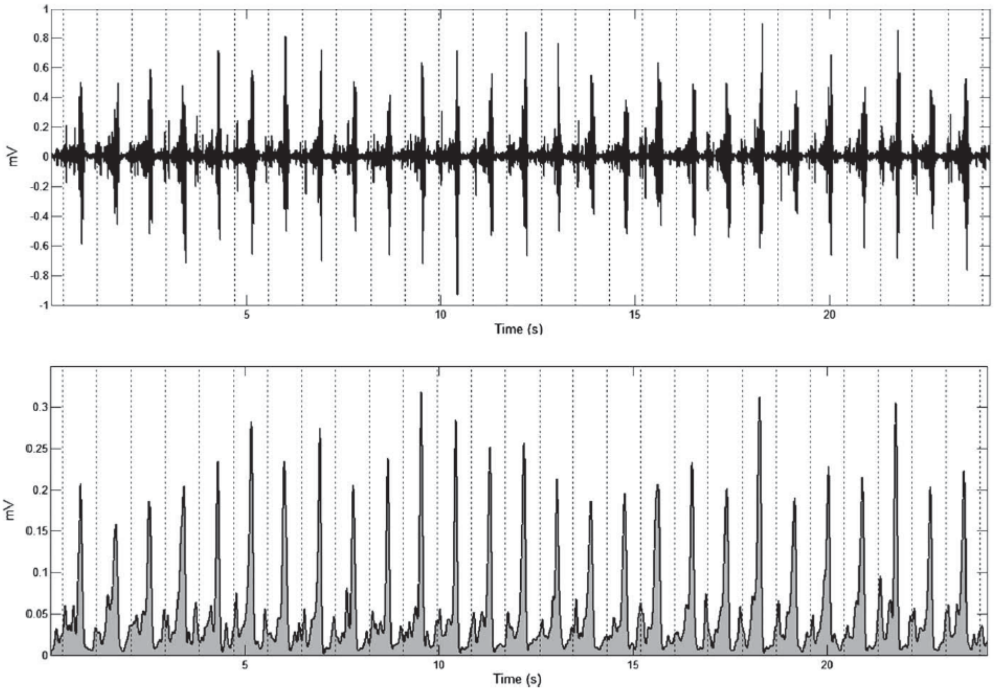

Figure 5 The sample EMG data from the vastus lateralis muscle includes two figures: the top shows band-pass filtered EMG signals, while the bottom presents integrated EMG data. Dotted vertical lines mark the start of each stride, aligning muscle activity with the running gait cycle [44]

Muscle activity tends to peak during the period immediately preceding and following initial ground contact, indicating that muscle contraction plays a more critical role in this phase compared to the preparation for and execution of toe-off. This observation aligns with DeVita's argument that the biomechanical events occurring around initial contact are of greater significance than those near toe-off [29,46]. Based on this premise, DeVita recommends presenting the swing phase before the stance phase when visually depicting the running gait cycle.

The quadriceps and rectus femoris are activated from late swing through midstance to prepare the limb for ground contact and to absorb impact forces during the stance phase. Quadriceps activation begins at 87% of the gait cycle, 78 milliseconds prior to initial contact, supporting the generation of muscle force necessary just before ground contact. Notably, the rectus femoris is active during mid-swing, where it plays a critical role in restraining the posterior movement of the tibia as the knee flexes. The hamstrings and hip extensors extend the hip during the second half of the swing phase and the first half of stance, with the hamstrings decelerating the forward momentum of the tibia as the knee extends before ground contact. Like the rectus femoris, the biarticular hamstrings contribute to energy transfer between segments. Both the hamstrings and gastrocnemius-soleus complex perform significant eccentric and concentric functions, while the hip extensors likely operate only concentrically. The tibialis anterior dorsiflexes the ankle to ensure foot clearance during swing (concentric contraction), facilitates initial ground contact with the hindfoot, and eccentrically controls the forefoot's descent during the early stance phase [29,47] (as shown in Figure 6).

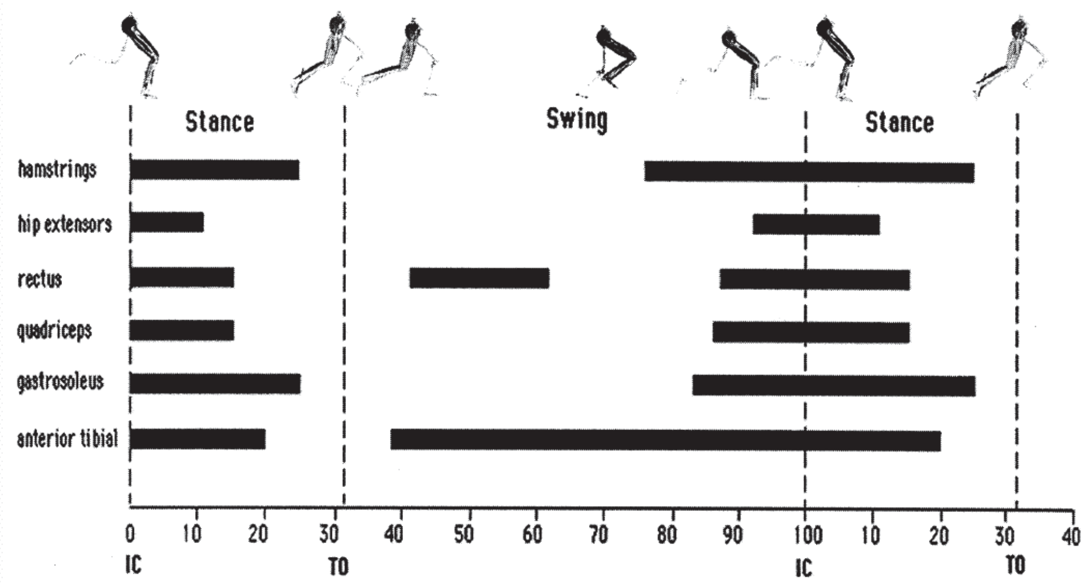

Figure 6 The EMG activity during the running gait cycle is depicted with solid bars representing muscle activation throughout. IC: Initial contact, TO: Toe off [29]

The comprehensive analysis of biomechanical characteristics during locomotion, including spatial-temporal, kinematic, and kinetic parameters, holds significant potential for advancing our understanding of running mechanics. By examining these factors, researchers can gain valuable insights into the intricacies of running movement patterns, which can, in turn, inform clinical interventions aimed at reducing the risk of injury. Moreover, this biomechanical data can be instrumental in developing targeted strategies for prevention of injury and rehabilitation, while also offering practical guidance for improving overall running performance and efficiency. Consequently, such investigations contribute not only to enhancing athletic outcomes but also to promoting long-term musculoskeletal health among runners.

1.2 Running-related injuries

1.2.1 Overuse injuries

Running is widely recognized as one of the most effective forms of physical activity for achieving cardiovascular fitness and promoting sustained engagement in exercise. Moreover, numerous studies have established a clear link between regular physical activity and increased longevity. Participation in running spans across all age groups, with longitudinal data indicating that 56% of runners continue the activity after a decade, and 81% engage in regular exercise. Despite these benefits, the high incidence of RRI remains a significant concern. It is generally reported that approximately 50% of runners sustain injuries annually, and at any given time, around 25% are affected [48]. The annual injury incidence among long-distance runners varies considerably, with estimates ranging from 19.4% to 79.3% [5]. Given this elevated injury risk, the prevention of RRIs is a critical focus in sports medicine and public health.

Overuse injuries represent the most prevalent type of RRIs, with epidemiological studies estimating that up to 70% of both recreational and competitive runners experience such injuries within a 1-year period [49,50]. When the elasticity and resilience of a runner's musculoskeletal system and associated soft tissues gradually become misaligned with the intensity of their running activity, minor damage to the musculoskeletal system may occur. Repeated microtrauma and strain can progressively lead to more significant injuries, which are classified as overuse injuries [51,52]. These injuries arise when a series of repetitive forces are applied to a structure, such as a muscle or tendon, with each force being below the acute injury threshold of the tissue [53]. Overuse injuries in running are typically defined as musculoskeletal conditions directly attributed to running that result in the restriction of running speed, distance, duration, or frequency for at least one week [49,54]. Common examples of overuse injuries in runners include stress fractures, medial tibial stress syndrome, chondromalacia patellae, plantar fasciitis, and Achilles tendinitis [51]. These injuries can negate the health benefits of running by reducing or eliminating participation in the activity, and they can pose significant financial, medical, and emotional burdens. Moreover, commonly employed treatments, including rest, physical therapy, bracing, medications, and surgery, may alleviate symptoms but often fail to address the underlying causes, potentially leading to diminished long-term activity levels [4,55,56].

The knee is the most frequent site of overuse injuries in runners, representing nearly 50% of all reported cases [57-59]. A systematic review and meta-analysis identified the knee as the predominant location of musculoskeletal injuries among runners [5]. Clinical data from a cohort of over 2,000 injured runners revealed that patellofemoral pain syndrome is the most prevalent knee injury, followed by iliotibial band syndrome, meniscal injuries, and patellar tendinitis. Injuries to the foot, ankle, and lower leg, including plantar fasciitis, Achilles tendinitis, and medial tibial stress syndrome (commonly referred to as shin splints), account for approximately 40% of remaining injuries. Less than 20% of running injuries occur above the knee. Although the specific etiology of most overuse running injuries remains unclear, the fact that over 80% of these injuries occur at or below the knee suggests the involvement of shared biomechanical factors [52,58].

Although the precise causes of overuse running injuries remain undetermined, it is well-established that their etiology is multifactorial and diverse [4,60]. The majority of factors contributing to these injuries can be broadly categorized into three groups: training, anatomical, and biomechanical factors [51]. Training-related variables most frequently associated with overuse injuries include running frequency, duration, distance, and speed, with clinical observations estimating that over 60% of running injuries can be attributed to training errors [49,51]. Anatomical factors, such as high longitudinal arches (pes cavus), ankle range of motion, leg length discrepancies, and lower extremity alignment abnormalities, have also been implicated [27,52,61,62].

However, there remains no consensus on the impact of these variables due to conflicting findings in the literature.

Biomechanical factors are considered to play a significant role in understanding overuse injuries, as they can be modified through specific interventions [63,64]. It has been proposed that certain biomechanical patterns may result in abnormal loading on neuromusculoskeletal structures, thereby increasing the risk of RRIs [65]. Biomechanical factors linked to overuse injuries are primarily divided into kinetic, kinematic and spatiotemporal variables [56-58] (as shown in Figure 7 and Figure 8). Kinetic factors hypothesized to contribute to overuse injuries include the magnitude of impact forces, the rate of impact loading, the magnitude of active (propulsive) forces, and the forces and moments at the knee joint [33,51,66-68]. Vertical loading rate has been linked to specific RRIs, including tibial stress fractures and plantar fasciopath[69,70]. Additionally, elevated frontal plane knee joint angular impulses may contribute to increased patellofemoral joint stress during repetitive running cycles [71]. Kinematic variables, particularly the magnitude of joint ankle, have also been frequently associated with overuse injuries [72,73]. Increased hip adduction has been associated with elevated strain on the iliotibial band and heightened patellofemoral joint stress. Additionally, reduced peak knee flexion may theoretically indicate decreased efficiency in load absorption at the knee, potentially increasing tension in the calf and Achilles tendon. However, conflicting findings have been reported regarding peak ankle eversion velocity, and evidence remains inconsistent for variables such as peak ankle eversion, peak rearfoot eversion, and reduced ankle eversion range of motion. Consequently, current prospective evidence does not support the commonly held belief that ankle and rearfoot eversion are significant risk factors for RRIs [65]. For spatiotemporal characteristics, particularly step rate, and RRIs have shown inconsistent findings. Although research on the impact of modifying step rate on both injury risk and performance remains limited, existing evidence suggests that increasing the running step rate may effectively reduce loading on specific tissues. This intervention could be particularly beneficial for certain injury presentations, such as patellofemoral pain [74].

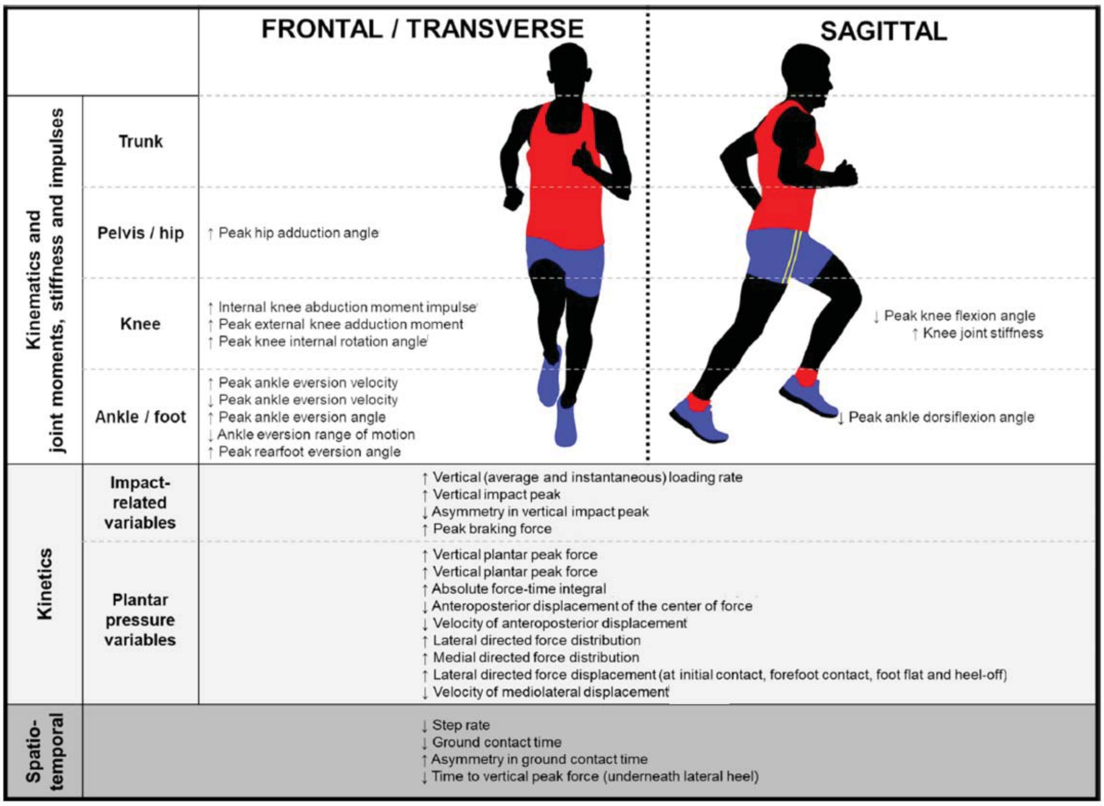

Figure 7 Biomechanical risk factors for running injuries [65]

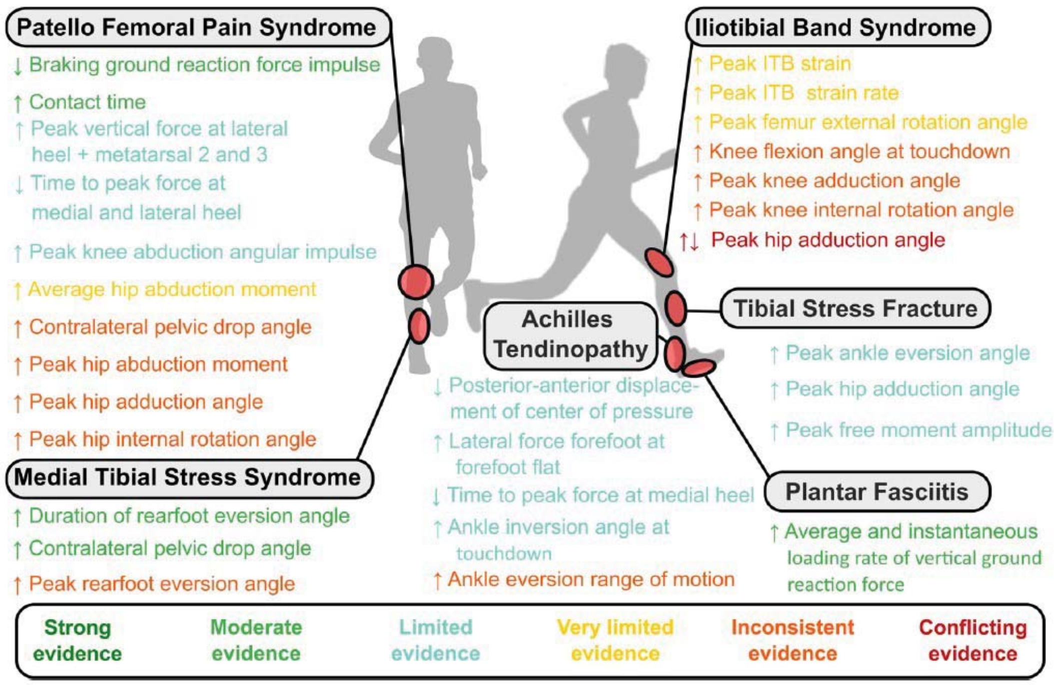

Figure 8 Overview of the running-related risk biomechanical factors [64]

1.2.2 Acute injuries

The incidence of RRIs was reported at 0.08 injuries per 1000 km of running exposure, with overuse injuries being more common, occurring at a rate of 0.07 per 1000 km, compared to acute injuries at 0.01 per 1000 km [75]. Acute running injuries are rare, common acute injuries include ankle sprains and hamstring strains [76].

Ankle sprains typically result from inversion mechanisms that damage the lateral ankle ligaments. A recent survey suggests that muscle fatigue leaves the foot in a vulnerable position, increasing the likelihood of acute injuries. This vulnerability arises from the collapse of joint stability due to an impaired ability to dynamically resist inversion or eversion forces, primarily caused by peroneal muscle weakness. Consequently, this condition predisposes individuals to ankle sprains and excessive strain injuries [77,78]. A meta-analysis of 46 systematic reviews on ankle sprain treatments indicates that functional bracing is the most effective intervention. Bracing, including taping, external supports, and orthoses, is recommended during physical activity for six to twelve months post-injury to enhance stability and reduce the risk of recurrence [79]. Additionally, strong evidence supports the use of exercise therapy to prevent reinjury. Significant attention has been directed toward understanding the role of footwear as a protective mechanism for the foot, as it aids in preventing both acute injuries and chronic conditions during physical activity. Proper shoe design and function are critical in mitigating the risk of injury by providing support, cushioning, and stability, thereby reducing the impact on the musculoskeletal system during exercise [33,39,80].

A hamstring strain injury is defined as the sudden onset of pain in the posterior thigh, leading to the immediate cessation of physical activity. This is another common acute injury characterized by proximal pain, often occurring suddenly during sprinting [81]. These strains may be accompanied by bruising or a palpable focal muscle defect. While plain radiography is generally unhelpful unless an avulsion of the ischial tuberosity is suspected, magnetic resonance images (MRI) can be used to assess the severity of the injury and estimate the time required for return to activity [82]. Management of acute hamstring strains includes limiting activity until normal walking is restored, followed by physical therapy to gradually improve range of motion [76,83].

Although numerous studies have investigated the prevalence and incidence of RRIs, it is crucial to acknowledge that differences in injury definitions, research methodologies, and running modalities across these studies may contribute to significant discrepancies in the reported incidence rates. For instance, some studies may define an injury based on medical diagnosis, while others might rely on self-reported pain or discomfort, leading to inconsistencies in data. Additionally, variations in the types of running surfaces, distances covered, and the level of running experience among participants can further complicate comparisons between studies. These methodological differences mean that the reported incidence rates of RRIs should be interpreted with caution. Despite these discrepancies, one consistent finding is that runners, particularly in the lower extremities, are at a significantly higher risk of injury. The lower limbs bear much of the repetitive load during running, making them particularly susceptible to injuries.

1.3 Factors affecting running biomechanics

1.3.1 Running experience

Over the past years, running experience has been currently one of the most widely debated topics in running research, primarily because it has been proposed that it is related to RRI risk and running performance. Notably, it appears that lacking running experience is associated with a higher risk of RRI. Runners with years of running experience may have more adapted musculoskeletal systems, while novice runners may not have the same tolerance for running loads [51,84,85]. Novice runners are particularly vulnerable to injury in all runner groups. Studies [84,86,87] have shown that their injury rate was higher than that of experienced runners; the rate of injury risks was 17.8 per 1000 hours of running against 7.7, respectively. Nielsen et al. [85] found that with an increase in running volume, the risk of injury per 1,000 hours of running significantly decreases. Novice runners experience approximately 30 to 38 RRIs per

1,000 hours, whereas marathon runners who maintain a weekly running volume of over 200 minutes have fewer than 10 injuries per 1,000 hours. The high injury rate among novice runners is largely attributed to their lack of structured running training and limited marathon experience. Placing an emphasis on RRI prevention for novice runners is essential, as early injuries can be a barrier to continuing the running program [88].

Running biomechanics are increasingly considered important factors in the study of injury development. Boyer et al. [89] found differences in pelvic rotation, hip internal rotation, and hip and knee abduction and adduction angles during running between lower and higher mileage runners. Quan et al. [90] found that, compared with experienced runners, runners with less experience had a greater plantar flexion angle, dorsiflexion angle, range of motion (ROM), plantar flexion moment, and angular velocity in the ankle joint, and a greater flexion angle and range of motion in the hip joint, which indicate higher injury risks. A greater peak hip internal rotation angle was found among novice runners, which may be linked to knee injuries [88]. Experienced runners also showed less variability in stride interval than novice runners, which indicated that larger running volumes could develop stable and consistent movement patterns [91]. Fatigue-induced alterations in running kinematics are likely to be more pronounced in novice runners due to their insufficient training adaptations and technical proficiency, which limit their ability to sustain pre-fatigue movement patterns during prolonged exertion [92-94]. However, Agresta et al. [95] suggested that running experience does not change joint kinematics and kinetics or GRF variables during running. They suggested that the importance of expertise in preventing injury may not lie in enhanced running mechanics, but rather in enhanced motor patterns and functional adaptation to the environment or biological stresses.

The running level of individuals can significantly influence the occurrence of RRIs, making it essential to define runners' skill levels. Some studies suggest that any runner who has participated in a marathon but whose race performance is not comparable to professional runners can be classified as an amateur runner [96]. Other research defines novice runners as those who, within the past year, have run fewer than four times and covered less than 2 km per session [97]. Maas et al. [92] defined novice runners as individuals with a weekly running volume of less than 10 km, who have not participated in competitive running events or followed a structured training plan. Sinclair et al. [98] considered recreational runners to be those who train at least three times per week and have more than five years of long-distance running experience. Although there is ongoing debate regarding the exact definition of running levels, a generally accepted view is that novice runners typically lack systematic training and a regular running habit, while experienced runners can be considered those with a basic level of training and consistent running practice.

1.3.2 Gender differences

Running biomechanics are significantly influenced by the anatomical and physiological characteristics of the individual, which vary between males and females. Differences in pelvic and thigh morphology between sexes contribute to distinct biomechanical patterns during running [99]. Specifically, gender-related anatomical variations impact lower extremity kinematics, with female runners generally exhibiting greater ranges of hip adduction, hip internal rotation, and knee abduction compared to male runners, particularly in the frontal and transverse planes [16,17]. These sex-specific biomechanical differences have been the focus of considerable attention, especially in relation to their potential role in RRIs. Evidence indicates that female runners are nearly twice as likely to experience RRIs such as patellofemoral pain syndrome, stress fractures, iliotibial band syndrome, or gluteus medius injury. The incidence of RRIs is reported to range from 62–76% in female runners, compared to 24–32% in their male counterparts [58].

Gender-related differences in kinematics and kinetics during running have previously been reported. Besson et al. [100] found that female runners showed larger hip and knee joint motion in the non-sagittal plane than male runners. Almonroeder and Benson [101] also noticed that hip adduction and internal rotation are greater in females than in males. A study conducted by Sinclair and Selfe [98] showed that among recreational runners, females demonstrated significantly larger extension and abduction moments in the knee joint, as well as greater patellofemoral contact forces and pressures than males, which may relate to the greater risk of patellofemoral pain in female runners. Most studies of gender differences in running biomechanics have focused on lower limb joint biomechanics. Studies examining differences in GRFs between runners of different genders are limited and inconsistent. Bazuelo-Ruiz et al. [102] conducted a prospective study and found that females have a significantly greater loading rate and peak propulsive force, and a smaller active peak force than males. Isherwood et al. [103] also observed that females exhibited a greater loading rate than males. However, the findings of a study conducted by Greenhalgh. [104] indicate that no significant differences in GRF were observed between males and females. During running, runners experience vertical GRF between 1.5 and 3 times their BW, which is believed to be a significant risk factor for lower limb injuries.

Differences in lower limb biomechanics between male and female runners also play a crucial role in influencing running economy [18]. Mechanical work is a key determinant of energy expenditure during movement, and factors such as joint kinematics, joint kinetics, and muscle activation patterns significantly impact running efficiency [105]. Folland et al. [106] highlighted that variations in vertical pelvic oscillation, knee joint flexion, and horizontal pelvic velocity, associated with running performance, can markedly affect running economy. Previous studies [107] have consistently shown that male runners are generally more efficient, utilizing less oxygen at a given running speed compared to females. This disparity in running economy may partly account for the observed differences in performance outcomes between male and female runners.

A comprehensive understanding of the biomechanical differences between male and female runners is essential for gaining insights into the sex-specific patterns of RRIs. These variations in biomechanics may explain why certain injuries are more prevalent in one sex compared to the other. By identifying and addressing these differences, targeted injury prevention strategies can be developed, tailored to the unique needs of male and female runners. Such strategies would not only reduce the overall incidence of RRIs but also improve training efficiency and long-term performance for both sexes. Additionally, this understanding could lead to more personalized approaches in coaching, footwear design, and rehabilitation protocols, ultimately fostering safer and healthier running practices

1.3.3 Running speed

To improve cardiovascular fitness and increase running speed over a fixed distance, structured training is essential. Running speed can be increased through two primary mechanisms: exerting greater force on the ground ("strategy 1") or increasing the frequency of ground contacts ("strategy 2"), or by combining both strategies [108]. At the initial stages of speed increase, strategy 1—applying more force during ground contact—appears to be the dominant approach. This increased force leads to a longer stride length, as the body spends more time in the air. This observation aligns with the biomechanical response seen when transitioning from jogging to slow-pace running, where stride length increased by 63%, while stride frequency increased by only 4%. Thus, early speed increases predominantly on greater force application, rather than more frequent steps.

Enhancing the running pace in recreational runners to maximize health-related benefits may be optimized through a better understanding of joint loading strategies [109]. Modulating the biomechanical contributions of the lower limb joints to sustain a given running speed may involve systematic adaptations, with one joint contributing more significantly than others as speed increases. This phenomenon has been observed in walking, where the hip exhibits a proportionally greater contribution to maintaining faster speeds. Understanding similar biomechanical adaptations in running could inform more effective coaching strategies for improving performance while mitigating injury risk [110].

During running training sessions, speed is commonly adjusted and serves as an indicator of the task's physical intensity [111]. As running speed increases, there is often a rise in stride length, frequency, joint range of motion, joint moment, joint load, and vertical impact force [109,112,113]. Runners are believed to experience greater forces on their bodies as they run faster. Specifically, within a speed range of 2–7 m/s, runners achieve a longer stride length by producing increased GRFs [19,114]. However, not all biomechanical parameters change with increased running speed. Floría et al. [111] observed no impact of speed on coordination variability when compared to three different running speeds. Girard et al. [115] accomplished a study on the impact of varying running speeds, from 10 to 25 km/h, on the extent and variation of asymmetry in essential biomechanical aspects. They concluded that the speed of running does not affect the mechanical asymmetry of the lower limb. A prospective study conducted by Muñoz-Jimenez et al. [116] revealed no significant differences in foot strike patterns, frequencies or percentages between low-speed and high-speed running. Kyröläinen et al. [45] examined the EMG activity of the leg muscles and GRFs in 17 male runners during both isometric maximal voluntary contractions (MVC) and running at various speeds. The results showed that the average EMG activity of all the muscles studied increased as running speed increased, particularly during the pre-contact and braking phases. At higher speeds, the EMG activity of the gastrocnemius, vastus lateralis, biceps femoris, and gluteus maximus exceeded 100% MVC in these phases. While numerous studies have examined running biomechanics at varying speeds, most have focused on a single gender. It remains to be studied how different speeds affect the running mechanics of male and female runners and the gender differences between them.

A theoretical framework has categorized six common RRIs into two primary groups: those associated with excessive pace ("Pacing injuries") and those linked to excessive training volume ("Volume injuries") [117]. "Pacing injuries" were predominantly observed in the posterior lower leg and plantar regions, including Achilles tendinopathy, gastrocnemius injuries, and plantar fasciitis. In contrast, "Volume injuries" were typically localized in the anterior knee, such as patellofemoral pain syndrome, iliotibial band syndrome, and patellar tendinopathy. Notably, this classification was derived from clinical, non-experimental studies. To further explore the concept of "Pacing injuries" from a biomechanical perspective, it is essential to assess how variations in running speed influence the loading on the ankle and knee joints [118]. Given that the activity of the plantar flexors (i.e., the triceps surae and deep plantar flexor muscles) is thought to be associated with injuries in the posterior lower limb and plantar regions, and that quadriceps femoris activity is linked to injuries in the anterior knee, the classification proposed by the theoretical framework suggests that increased running speed places a greater mechanical demand on the plantar flexors compared to the knee extensors [117]. This implies that as running speed increases, the load on the muscles responsible for plantar flexion becomes disproportionately higher, which could explain the prevalence of "Pacing injuries" in these areas.

Reducing running speed is a potential strategy for mitigating load during running. A well-established positive relationship exists between GRFs and running speed [23,119], and it is intuitive to assume that a similar relationship applies to bone strain. However, reducing running speed also necessitates an increased number of loading cycles to cover a given distance. Therefore, it remains unclear whether the reduction in strain from slower running compensates for the increase in loading cycles required to maintain the same mileage. Runners aiming to lower their risk of tibial stress fractures may benefit from reducing running speed. In the context of the study by Edwards et al. [120], stress fracture development was found to be more strongly influenced by loading magnitude than by loading exposure, suggesting that reducing speed might effectively decrease the risk of tibial stress fractures despite the increase in loading cycles.

1.3.4 Long distance running

Major city marathons and international running events have significantly reshaped public perception of distance running, contributing to a surge in demand for running-related leisure activities. For participants, distance running and engagement in running events are seen as accessible pursuits for ordinary but determined individuals. Long distance running, typically defined as distances greater than 3000 meters, is commonly advised to promote a healthy lifestyle, and distance runners are increasingly recognized as a significant and growing group within society [121,122]. Despite the popularity of these events, between 37% and 56% of recreational runners who regularly train and participate in long-distance races experience a RRI annually [123]. A systematic review of injuries among long-distance runners highlights that the majority of injuries are localized to the knee (7.2%-50%), lower leg (9%-32.2%), foot (5.7%-39.3%), or thigh (3.4%-38.1%) [5]. Therefore, it is essential to investigate biomechanical variations in runners before and after long-distance runs. Such analysis can guide the development of effective training strategies, enhance performance, and help mitigate the risk of RRIs.

A recent review demonstrated that long-distance running generally leads to increased stride frequency and step frequency, along with reductions in stride length and aerial time [124]. These adaptations are likely a response to limiting overall mechanical loading, as excessive strain on the muscles during prolonged running may impair muscular function. To compensate, the body may adopt a softer foot strike with reduced impact forces and shorter, quicker steps [125]. Weist et al. [126] observed significant increases in loading under the second and third metatarsal heads, coupled with a notable decrease in electromyographic activity in the medial and lateral gastrocnemius and soleus muscles following high-intensity running. They attributed this load redistribution to localized muscle fatigue, particularly in the toe flexor muscles. As these muscles become fatigued, the reduced activation of the toes during the push-off phase results in a transfer of loading from the toes to the forefoot, leading to increased dorsiflexion at the metatarsophalangeal joints and heightened forefoot loading [42].

Long distance running appears to influence the biomechanics of the lower limb joints [27,80,94,127]. Quan et al. [128] reported a decrease in ankle joint work and an increase in negative knee power during post-fatigue running. At the hip joint, a reduction in the extension angle was observed, along with an increase in hip range of motion, hip positive work, and hip positive power. In a previous study, we found reductions in ankle dorsiflexion, hip flexion, hip adduction, and hip internal rotation angles after a 5 km run, along with increases in ankle eversion, ankle external rotation, knee adduction, and knee internal rotation angles [80]. Mei et al. [27] reported that after a prolonged running session, hip joint moments and contact forces increased during the initial foot contact. Additionally, there was an observed increase in knee abduction moment and superior-inferior knee contact force, while the knee extension moment decreased. Ankle plantarflexion moment and ankle contact forces were also found to increase during the stance phase. After long-distance running, lower vertical GRFs and higher impact acceleration are also common. Gao et al. [129] observed that following a fatigue protocol induced by running, the symmetry angles for knee extension, knee abduction moment, and hip flexion moment increased by 17%, 10%, and 11%, respectively. In contrast, the knee joint's flexion angle decreased by 5%. Additionally, the symmetry of internal rotation in the ankle, knee, and hip joints increased post-fatigue, while the symmetry angles of external rotation in these joints significantly decreased. The results offer preliminary evidence that fatigue alters lower limb symmetry during running gait. These biomechanical changes suggest altered loading patterns in the lower limb following extended running, which may have implications for injury risk and performance [33,124].

All sustained physical activities induce varying degrees of fatigue within the body, and fatigue is widely recognized as a major limiting factor in distance running performance, manifesting as a reduction in power output and a decline in overall performance. Fatigue can arise from three primary sources: within the central nervous system (CNS), at the level of neural transmission between the CNS and muscles, and within the muscle fibers themselves [130]. Running, as a natural form of locomotion, involves the CNS regulating groups of muscles in response to mechanical demands. Recently, a model has been proposed that views the CNS as a key controller of motor performance. This model reduces the degrees of freedom in how the CNS manages muscle groups and movements, allowing the muscles to be activated collectively as functional units. This streamlined approach enhances motor efficiency during running and other activities [131]. Local muscular fatigue, caused by intense activity in specific muscles, is a common form of fatigue experienced by distance runners [93,132]. While long-distance runners are certainly affected by fatigue, middle-distance runners may be more susceptible in certain respects. This increased vulnerability may be attributed to their higher proportion of fast-twitch muscle fibers, which are more prone to fatigue, and to their competitive pacing strategies, which often include multiple bursts of high acceleration during races [133,134]. Despite the shorter race distances, these factors can make middle-distance runners more vulnerable to fatigue than their long-distance counterparts.

In summary, biomechanical differences in running are influenced by a complex interplay of factors such as experience level, gender, running speed, distance, and even individual physical characteristics. These variables not only shape the way runners move but also contribute to distinct patterns of running mechanics that can lead to varying degrees and types of RRIs. For instance, novice runners may exhibit different joint angles and ground reaction forces compared to experienced runners, while gender differences can influence stride patterns and loading behaviors. Running speed and distance further compound these variations, making a comprehensive investigation into these factors essential for understanding the biomechanical mechanisms behind injury development. A detailed analysis of these variables is crucial for identifying the specific factors that contribute to RRIs. By exploring how these elements interact and affect the body during running, researchers can provide valuable insights that inform the design of more effective injury prevention strategies. Ultimately, these findings can be used to develop individualized training programs that optimize performance while minimizing injury risk, promoting healthier and more sustainable running practices across different populations

2. MATERIAL AND METHODS

2.1 Experiments

2.1.1 Participants

Section 1: The required sample size was determined on the basis of a previous study [92]. A sample size of 12 runners per group was calculated, considering an effect size of 0.93 and a power of 0.9. In order to accommodate any potential data loss, 30 healthy male runners were enrolled in this study, including 15 experienced runners and 15 novice runners (Table 1). Novice runners were defined as those who ran 2–10 km per week and did not take part in a running competition or training program. However, it is important to note that they did have regular exercise habits and reported a minimum score of 5 out of 10 on the Tegner activity scale. Experienced runners consistently run at least 30 km per week and have more than 3 years of running experience [92,135]. Participants were heel-strike runners, with their dominant leg being the right (which was defined as the leg that was preferred for kicking a ball). Both novice and experienced runners had treadmill running experience and had no lower limb injuries or musculoskeletal system disorders in the 6 months before the test. This study was a cross-sectional controlled study.

Table 1 Mean (SD) of participant characteristics of novice and experienced runners

	Novice	Experienced	p Value
Age(years)	23.80(1.97)	23.65(1.67)	0.398
Height(m)	1.76 (0.49)	1.75 (0.56)	0.702
Body weight(kg)	71.93(7.70)	72.73(6.44)	0.794
BMI (kg/m^2)	23.13(1.18)	23.65(1.67)	0.456
Running experience (years)	1.53(0.74)	6.07(1.62)	<0.001
Running volume (km/week)	7.13(2.67)	35.67(9.23)	<0.001

Note: Statistical significance was set to $p < 0.05$. The bold represented significant differences.

Section 2: Thirty male (age: 25.80 ± 3.44 years, height: 1.76 ± 0.05 m, body mass: 75.70 ± 6.14 kg) and eighteen female recreational runners (age: 24.89 ± 2.77 years, height: 1.63 ± 0.04 m, body mass: 54.83 ± 5.15 kg) participated in this study. Participants were recruited from sports clubs and via social media. All runners self-identified as rearfoot strike pattern runners and ran at least 20 km per week. The exclusion criteria for the study were: 1) any lower limb injury within the past 6 months; 2) any low back or lower limb pain during running; 3) less than 3 years of running experience.

The studies involving humans were approved by the Ethical committee of Ningbo University (protocol code RAGH20210627). The studies were conducted in accordance with the local legislation and institutional requirements. The participants provided their written informed consent to participate in this study (as shown in Figure 9).

Research Academy of Grand Health, Ningbo University **Human Informed Consent Form**

Instructions to the Student Researcher(s): An informed consent/assent/permission form should be developed in consultation with the Adult Sponsor, Designated Supervisor or Qualified Scientist.
This form is used to provide information to the research participant (or parent/guardian) and to document written informed consent, minor assent, and/or parental permission.

- When written documentation is required, the researcher keeps the original, signed form.
- Students may use this sample form or may copy ALL elements of it into a new document.

If the form is serving to document parental permission, a copy of any survey or questionnaire must be attached.

Student Researcher(s): ______

Title of Project: ______

I am asking for your voluntary participation in my science fair project. Please read the following information about the project. If you would like to participate, please sign in the appropriate area below.
Purpose of the project:

If you participate, you will be asked to:

Time required for participation:

Potential Risks of Study:

Benefits:

How confidentiality will be maintained:

If you have any questions about this study, feel free to contact:

Adult Sponsor/QS/DS: ______ Phone/email: ______

Voluntary Participation:
Participation in this study is completely voluntary. If you decide not to participate there will not be any negative consequences. Please be aware that if you decide to participate, you may stop participating at any time and you may decide not to answer any specific question.
By signing this form I am attesting that I have read and understand the information above and I freely give my consent/assent to participate or permission for my child to participate.

Adult Informed Consent or Minor Assent Date Reviewed & Signed: ______

Research Participant Printed Name: Signature:

Parental/Guardian Permission (if applicable) Date Reviewed & Signed: ______

Parent/Guardian Printed Name: Signature:

Figure 9 Human informed consent form

2.1.2 Instruments and Materials

The Biomechanics Laboratory at the Ningbo University Research Academy of Grand Health is fully equipped with advanced instrumentation necessary for conducting comprehensive research in the field of sports biomechanics. All data collection and experimental procedures were carried out within this facility, ensuring precise and controlled conditions. The specific details of the experimental setup and testing protocols are outlined as follows:

1) Vicon motion system

Kinematic data were collected using a Vicon 3D motion capture system (Oxford Metrics Ltd., Oxford, UK), which consisted of eight infrared high-speed cameras and the accompanying Nexus software. The sampling frequency for this experiment was set at 200 Hz.

2) AMTI force plate

Kinetic data were collected using an AMTI 3D force platform system (AMTI, Watertown, MA, USA), consisting of a force plate measuring 60 cm in length and 80 cm in width, equipped with four built-in piezoelectric three-dimensional force sensors and a control system. The sampling frequency for this experiment was set at 1000 Hz. During data collection, the system was synchronized with the Vicon Nexus software to simultaneously capture both kinematic and kinetic data during the running trials.

3) Delsys Trigno surface electromyogram (EMG)

Surface EMG signals were collected using a 16-channel Delsys Trigno Wireless EMG system (Delsys, USA) along with the accompanying EMGworks Analysis software. The sampling frequency for this experiment was set at 2000 Hz.

4) Timing gates

The speed monitoring equipment (SmartSpeed, Fusion Sport Inc., Burbank, California, USA) was positioned on both sides of the force platform to control the runners' pace during the trials. The timing gates were placed 3 m apart alongside the runway just before and after the force plate.

5) Heart rate monitor

The runners' heart rate was continuously monitored in real-time during the long-distance running session using a Polar heart rate monitor (RS 400; Polar Electro Oy, Kempele, Finland).

6) h/p/cosmos treadmill

The 5 km run was conducted on an indoor treadmill (h/p/cosmos, Nussdorf-Traunstein, Germany).

7) Experimental shoes

All participants in this experiment wore standardized footwear, specifically ANTA neutral running shoes (ART NO. 11725599-7, ANTA, China) (Figure 10).

Figure 10 Shoes for the experiment

2.1.3 Experimental procedures

Runners were first required to complete a personal information form. The experimenters then explained the experimental procedures and precautions. After understanding the details of the experiment, subjects signed an informed consent form. Participants were then provided with standardized compression pants and running shoes. Baseline anthropometric measurements, such as height, weight, and leg length, were taken. To prevent injuries during the long-distance running trials, runners performed a 10-minute warm-up by jogging on a treadmill (speed set to 8 km/h) while wearing the experimental shoes to become accustomed to them. Following the warm-up, runners adjusted their gait and familiarized themselves with the laboratory track to ensure that the right foot would land on the force plate during the formal trials.

In accordance with the experimental modeling requirements, the OpenSim 2392 musculoskeletal model (featuring 23 degrees of freedom and 92 muscle-tendon units) was used. Reflective markers were attached to specific anatomical landmarks on the subjects' bodies. The detailed marker placement protocol is provided in Figure 11. All marker placements were conducted by the same operator for consistency. After the reflective markers were affixed, a static trial was performed. Subjects were instructed to adopt a standard anatomical position: standing on the force plate with feet shoulder-width apart, arms abducted at approximately 45 degrees downward, and gazing straight ahead. They remained stationary until the experimenters completed the collection of static data.

Figure 11 Illustration of markers placement

For section 1 (Figure 12), the experiment consisted of two phases of biomechanical data collection: one before and one after the 5 km run. The data collection process was identical for both phases. Subjects were instructed to run naturally along the test track without deliberately aiming to step on the force plate. The criteria for a successful trial were as follows: the subject's right foot must completely land on the force plate; there must be no noticeable adjustment of stride during data collection; the running speed must be maintained within the acceptable range of 2.77–3.33 m/s; and no reflective markers should become detached during the trial. Kinematic, kinetic, and surface EMG data were collected synchronously during the tests. The EMG sensors collected muscle activity data from the rectus femoris, vastus lateralis, vastus medialis, medial gastrocnemius, lateral gastrocnemius, and tibialis anterior muscles. The EMG electrodes were affixed according to the SENIAM project (http://www.seniam.org/). Each runner was required to complete three successful data trials before and after the 5 km run for subsequent data processing and analysis. Following the biomechanical data collection prior to the 5 km run, runners completed the 5 km run on a treadmill at a self-selected pace (9–12 km/h), wearing a heart rate monitor to track heart rate throughout the run. Reflective markers remained in place during the entire run. Within two minutes of completing the 5 km run, post-run biomechanical data were collected following the same protocol as before the run.

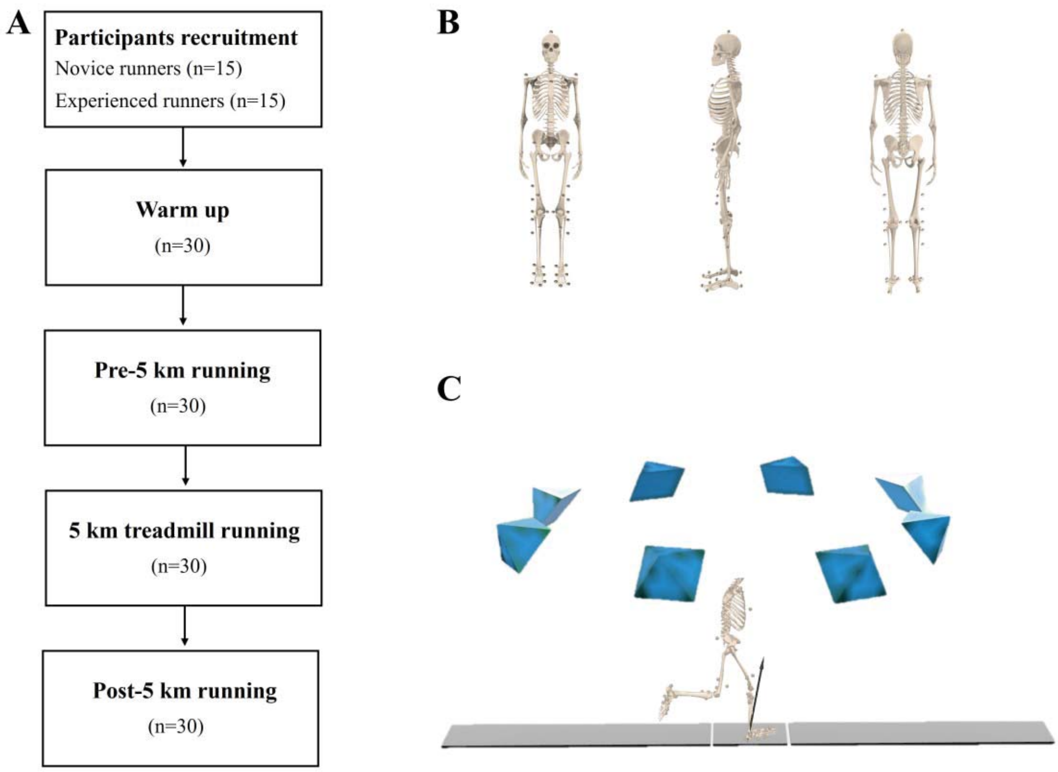

Figure 12 Illustration of session 1 testing protocol: (A) flowchart of participation in this study; (B) marker placement; (C) biomechanical data collection

For section 2 (Figure 13), after 10 min of laboratory familiarization and a warm-up, all runners performed running tests on a 20 m runway at seven speeds: 10, 11, 12, 13, 14, 15, and 16 km/h. Each participant completed three successful trials at each speed (± 2%) on the runway. The trial was considered successful only if runners struck the force plate with their right foot fully on without targeting, and the speed was within 2%

of the prescribed running speed. Runners were required to maintain a steady-state speed until they exited the runway. Running speed was measured by two infrared timing gates placed 3 m apart alongside the runway just before and after the force plate. The order of running speeds was non-random for practical reasons [109,113,136]. Adequate rest was provided between speed increases to prevent fatigue. Additionally, our experimenters had monitored the participants' fatigue levels throughout the process. If a participant had reported experiencing fatigue during the experiment, they would have been allowed to rest before continuing.

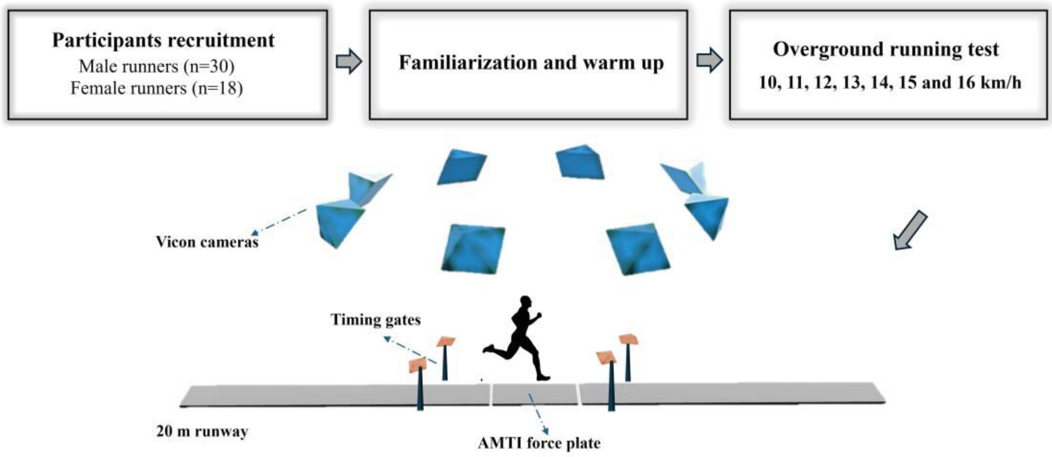

Figure 13 Illustration of session 2 testing protocol

2.2 Data analysis

2.2.1 Preprocessing raw data

The collected data was preprocessed using Vicon Nexus 1.8.5 software. Reflective markers were labeled, and the stance phase of each running cycle, defined as the period from initial contact of the right foot with the ground until its departure, was extracted. Missing markers during this phase were reconstructed, and any extraneous or erroneous markers were removed. The processed running data were then exported as C3D files and imported into OpenSim 4.2 (Stanford University, Stanford, CA, USA) for further processing and analysis. The C3D files were also imported into Visual 3D 6.0 (c-motion Inc., Germantown, MD, USA) for kinematic and kinetic processing [137].

Surface EMG data were used to validate the reliability and accuracy of the OpenSim simulation. The EMG data were filtered using a fourth-order Butterworth band-pass filter with a cutoff frequency of 10-500 Hz in EMGworks Analysis 4.0 software. A 10 Hz low-pass filter was applied to perform full-wave rectification. The EMG amplitudes for each muscle were normalized by dividing by the corresponding muscle's maximum root mean square (RMS) amplitude. The normalized muscle activation level during the stance phase of running was calculated by dividing the muscle activity level by the maximum muscle contraction level. These experimentally measured muscle activation levels were then compared to the muscle activation values calculated by the Static Optimization algorithm in OpenSim to verify the reliability and accuracy of the simulation results [73,135].

2.2.2 Simulation modeling

In this study, musculoskeletal modeling was conducted using OpenSim 4.3 (Stanford University, Stanford, CA, USA), for the extraction of muscle activities and knee joint forces, an upgraded version of the OpenSim 2392 musculoskeletal model was employed, which includes 10 rigid body segments, 23 degrees of freedom, and 92 muscle-tendon units [138,139]. The muscle-tendon units in the model are based on the Hill-type muscle model, adhering to muscle force-length-velocity relationships as well as tendon force-length and tendon force-strain relationships [140]. Each runner's height and weight were factored into the 3D model's dimensions. The segment lengths were computed using the reflective marker positions in the static posture, and anthropometric data and the runner's body mass were used to calculate the segment masses. An algorithm for computed muscle control (CMC) computes muscle excitations and forces to generate a muscle-driven simulation of the runner's running gaits. Next, use the analyze tool from OpenSim to determine the joint reaction force at the knee. Knee joint forces were divided into anterior-posterior (shear), medial-lateral (shear), and components superior-inferior (compressive) [141].

Visual 3D software (c-motion Inc., Germantown, MD, USA) was used to process calculations of the right lower-limb joint kinematics and kinetics of the running stance phase. For the denoising process of marker trajectories, A fourth-order low-pass Butterworth filter was used to filter kinematics and GRFs at frequencies of 10 Hz and 20 Hz, respectively. The stance phase was determined when the vertical GRF crossed a threshold of 20 N. A Cardan X–Y–Z rotation sequence was used to calculate ankle, knee, and hip joint angles. Inverse dynamics based on the Newton–Euler approach was used to compute the lower limb joint moments [80]. Matlab version 2019b (The Math Works, Natick, MA, USA) time-normalized the biomechanical data of the running stance phase to 101 points.

2.2.3 Variable selection and calculation

The kinematic parameters include the joint angles of the ankle, knee, and hip joints in the sagittal, frontal, and horizontal planes during the stance phase of running, as well as the ROM of these joints. The kinetic parameters include the 3D joint moments of the ankle, knee, and hip joints, as well as GRF parameters, knee reaction forces were also included in this study. Joint moments were normalized to the body mass, while GRF and knee reaction force were normalized to BW of each participant. The stance phase is defined as the period from initial contact (vertical GRF > 20 N) to toe-off (vertical GRF < 20 N). The interest GRFs were peak vertical impact force and peak vertical active force, vertical average loading rate (VALR), vertical impulse, contact time, peak propulsive force and propulsive impulse, peak braking force and braking impulse (Figure 14). These variables were the most relevant parameters selected according to previous research on GRFs during running [33,66,142]. Muscle force parameters include the peak muscle forces and muscle force curves of the biceps femoris long head (BFLH), rectus femoris (RF), vastus lateralis (VL), vastus medialis (VM), soleus (SO), gastrocnemius lateralis (GL), gastrocnemius medialis (GM), and tibialis anterior (TA) during the stance phase of running (Figure 15).

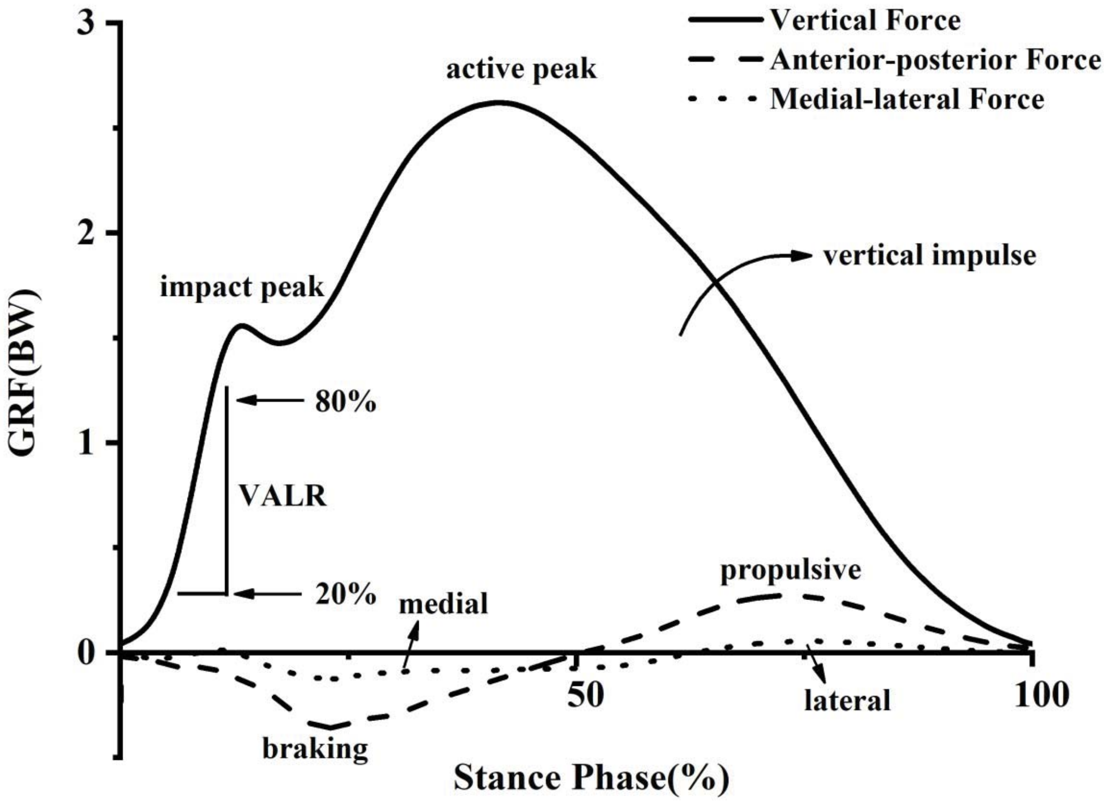

Figure 14 Example of ground reaction force trajectories for the stance phase of heel strike runners. Key variables of these trajectories are identified.

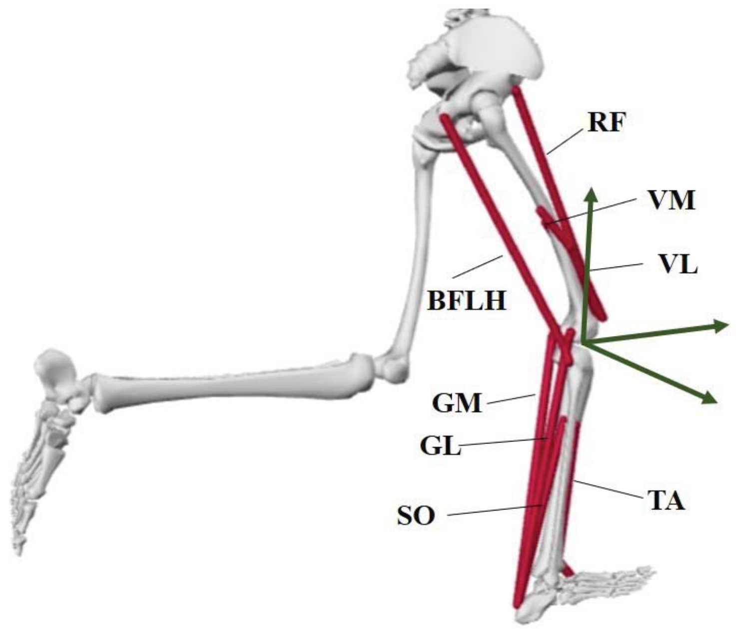

Figure 15 The musculoskeletal model of the lower limbs for the running simulation.

Note: Biceps femoris long head (BFLH), rectus femoris (RF), vastus lateralis (VL), vastus medialis (VM), soleus (SO), gastrocnemius lateralis (GL), gastrocnemius medialis (GM), and tibialis anterior (TA)

The muscle activations from the CMC algorithm were then compared to the activations derived from the EMG signals for validation [73]. The comparison of the EMG signals and musculoskeletal models of muscle activations exhibited in Figure 16. The knee forces were compared with previous literature.

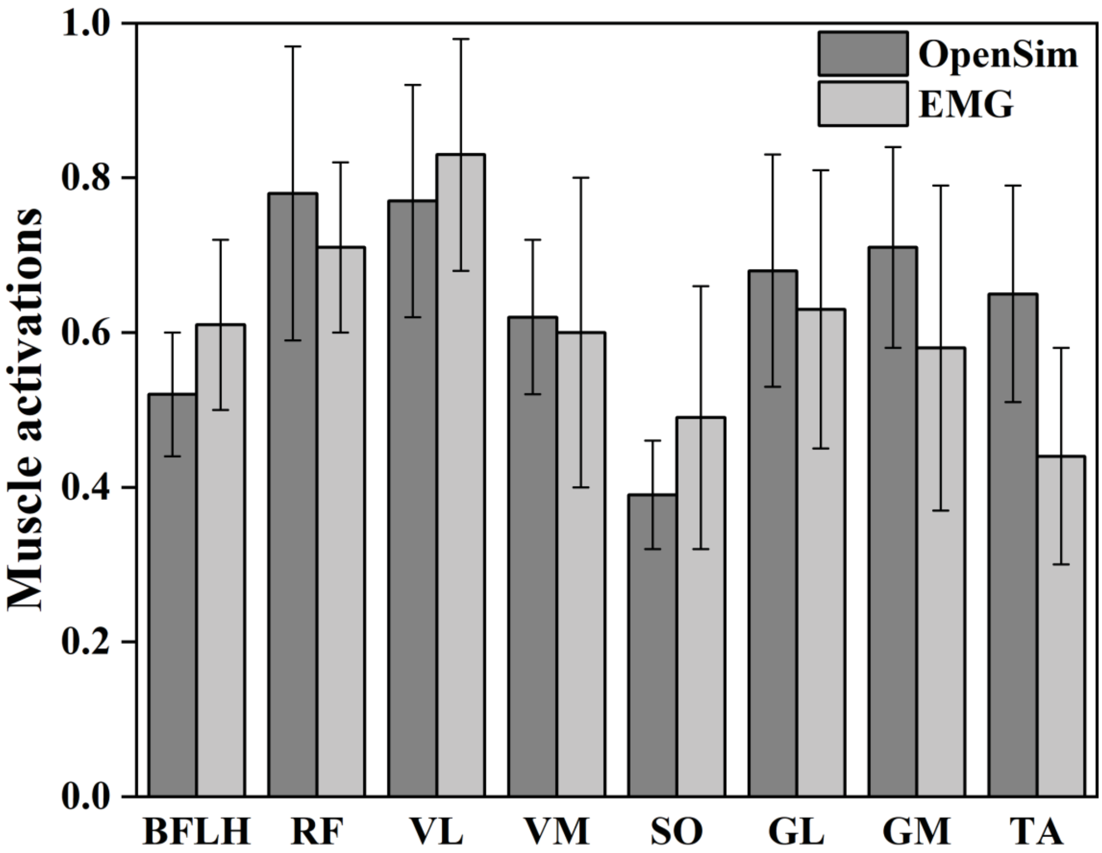

Figure 16 Comparison between the EMG muscle activation and OpenSim simulation

The peak vertical impact force was defined as the first peak in the vertical GRF, while the peak vertical active force was defined as the second peak. The VALR was computed from 20% to 80% of the stance period between the initial foot contact and the peak vertical impact. VALR was the average slope in the period [143]. Vertical impulse was determined as the time integral of the vertical GRF over stance. Impulses were calculated as the zone surrounded by the zero lines and GRF curves for each direction[144]. Contact time was considered as the time during which a vertical force greater than 20 N was applied to the force plate.

$$\boldsymbol{Impulse} = \sum_{i=1}^{n-1} \frac{1}{2} (\boldsymbol{F}_{i+1} + \boldsymbol{F}_i) \times (\boldsymbol{t}_{i+1} - \boldsymbol{t}_i) \tag{1}$$

In this equation, n is the number of frames, i is the i-th frame, F is the GRF and t is the time value.

Based on the principles of human anatomy, the definitions for the angles, angular velocities, moments, and power data of the lower limb joints are as follows: for the ankle joint, dorsiflexion is defined as positive, plantarflexion as negative, inversion as positive, eversion as negative, internal rotation as positive, and external rotation as negative. For the knee joint, extension is defined as positive, flexion as negative, adduction as positive, abduction as negative, internal rotation as positive, and external rotation as negative. For the hip joint, flexion is defined as positive, extension as negative, adduction as positive, abduction as negative, internal rotation as positive, and external rotation as negative. GRFs are defined that, in the sagittal plane, in the frontal plane, the anterior direction is positive and the posterior direction is negative; and in the horizontal plane, the vertical direction is positive. Due to high variability within and between subjects, medial-lateral GRFs were excluded from this study [145].

Principal component analysis (PCA) is a method based on a mathematical algorithm that reduces data dimensionality while retaining the majority of the common modes of variation in the dataset and providing information that may significantly increase classification accuracy. It is used to explain a set of correlated variables [146,147]. PCA has been widely utilized to investigate human movement tasks such as running, lifting, and walking [89,148, 149]. The advantages of the PCA method are that it permits a more comprehensive way of evaluating motion modes and has the potential to be a meaningful discriminator of sports-related injury risk. PCA can reduce locomotion and time series data without losing temporal information, and it produces independent principal components and scores [150]. By analyzing the modes of variation via PCA during running, it is possible to explain specific patterns in a set of variables. Therefore, through PCA analysis, differences between groups in joint motion of all lower limb joints and GRF can be investigated systematically.

The PCA method applied in the study was based on an approach described previously [151]. For every dimension of the joint angle and moment waveforms, the ensemble curves were separately combined into a matrix for PCA. Thus, PCA was performed on 18 separate $X_{n\times p}$ matrices, where n is the number of running trials, and p represents the 101 data points of the stance phase. For the present analysis, the waveforms of the three trials of 15 novice runners and 15 experienced runners for each of the two running conditions were inputted as row vectors, yielding $X_{180\times 101}$ matrices for each interest variable, resulting in the display of principal component models as follows:

$$X_{180\times 101} = \begin{bmatrix} x_{1,1} & x_{1,2} & \cdots & x_{1,101} \\ x_{2,1} & x_{2,2} & \cdots & x_{2,101} \\ \vdots & \vdots & \ddots & \vdots \\ x_{180,1} & x_{180,2} & \cdots & x_{180,101} \end{bmatrix} \tag{2}$$

The waveform data were transformed into uncorrelated principal components. The covariance matrix $S_{101\times 101}$ was subjected to eigenvalue analysis to perform PCA; $\bar{x}_{1\times 101}$ was the mean waveform of $X_{180\times 101}$ at each timepoint.

$$S_{101\times 101} = \frac{(X_{180\times 101} - (1_{180\times 1} \times \bar{x}_{1\times 101}))' \times (X_{180\times 101} - (1_{180\times 1} \times \bar{x}_{1\times 101}))}{(180-1)} \tag{3}$$

The eigenvector matrix $U_{101\times 101}$ was determined by orthonormalizing $S_{101\times 101}$. The columns of $U = u_1, u_2, \ldots u_{100}$ are named PC loading vectors. The spread along the direction of the eigenvectors was explained by the corresponding eigenvalues $L_{1\times 101}$.

$$L_{1\times 101} = diag(\, U_{101\times 101}{}' \times S_{101\times 101} \times \, U_{101\times 101}) \tag{4}$$

After $U_{101\times 101}$ and $L_{1\times 101}$ determined, the PC scores $Z_{180\times 101}$ for each waveform could be computed by the deviation of each waveform trial from the overall mean with the transpose of the eigenvector matrix. Thus, each runner's raw waveform was transformed into a set of PC scores, which indicate the similarity of their waveform shape to each specific feature.

$$Z_{180\times101} = \left(X_{180\times101} - (1_{180\times1} \times \bar{x}_{1\times101})\right) \times U'_{101\times101} \tag{5}$$

To assess the adequacy of the retained principal components in representing the original data, a residual analysis was completed using the Q-statistic. The Q-statistic is computed as the sum of squared residuals between the original waveform and the reconstructed curve generated from the retained PCs [152].

In this study, the first k PCs required to be retained were determined by 90% trace criteria [153]. This criterion ensures that the chosen PCs capture the main patterns of variation and account for a significant portion of the overall variation in the running data. k is the number of PCs retained in the model ($k \leq n$).

The interpretation of PCA involved visually analyzing the PC loading vectors and examining the waveforms that obtained low and high scores on each PC. This approach allowed for a comprehensive understanding of the relationships between the PCs and the corresponding waveform patterns. The high PC waveforms were defined by one standard deviation above (plus SD) each PC, and the low PC waveforms were defined by one standard deviation below (minus SD) each PC [148,154]. All the PCA processing calculations were completed in Matlab.

2.3 Statistical analysis

For session 1, the Shapiro–Wilk test confirmed that the PC scores and other discrete values of lower limb biomechanics retained for analysis were normally distributed. Raw data of the time sequential lower limb biomechanical variables were interpolated to 101 in data length to describe the running stance. Independent t-tests were employed to compare the demographics and running experience between novice and experienced runners; the significance level (alpha) was set at 0.05. A two-way repeated-measures analysis of variance (ANOVA) was conducted to quantify the main effects of running experience levels and 5 km run factors, as well as their interaction; statistical significance was accepted at $\alpha = 0.05$. A Bonferroni correction adjusted post hoc pairwise comparisons to $\alpha = 0.008$ when the significant interaction effect was observed [40,41]. The waveforms of muscle forces and knee joint forces were analyzed using one-dimensional statistical parametric mapping (SPM1d) in Matlab. Pearson's correlation coefficients were performed to evaluate the relationship between peak values of knee joint forces and muscle forces in both novice runners and experienced runners. All data were presented as means ± SD (standard deviation). Statistical analyses were completed using SPSS 25.0 (IBM, Armonk, NY, USA).

For section 2, average data for each participant was included in the analysis. The normality of the GRF variables was checked via Shapiro-Wilk tests. Pearson's correlation coefficients were computed to evaluate the relationship between GRF variables and running speed. Correlations were defined as: no relationship or little ($r \leqslant 0.25$), low to fair ($0.25 < r < 0.50$), moderate to good ($0.50 < r < 0.75$), and strong ($r \geqslant 0.75$) [155]. The significance level for determining whether a correlation is statistically significant was set at 0.05. To further determine the level of variance in running speed that was explained by the specific GRF variables, two stepwise linear regressions were performed (one for males and one for females). The discrete GRF variables that were significantly correlated with the running speed were input into one model as the independent variables, while the running speed was considered the dependent variable. The criteria for entering or removing variables from the model were set at alpha levels of 0.05 and 0.10, respectively. The data were analyzed via SPSS software (version 25.0, IBM Corporation, Armonk, NY, USA). Meanwhile, both vertical and anterior-posterior GRFs were normalized into 101 data points by using the cubic spline interpolation approach to represent stance phase (from 0 to 100%). Given the one-dimensional time-varying characters of GRF curves, a two-tailed independent t-test with SPM analyses was used to determine gender differences in each running speed, and a one-way repeated measures ANOVA with SPM1d was used to determine the main effect of running speed in both males and females.

3 RESULTS

3.1 Muscle force and knee reaction force between novice and experienced runners

3.1.1 Muscle force

Figure 17 showed significant differences between the novice and experienced runners in muscle forces during the running stance phase. The novice group exhibited greater muscle forces in BFLH at 68.25~100% ($p<0.001$), in RF at 0.44~10.14% ($p<0.001$) and 26.68~39.83% ($p<0.001$), in GL at 10.91~30.16% ($p<0.001$) and 42.59~100% ($p<0.001$), in SO at 72.53~96.96% ($p<0.001$) during the running stance phase. The characteristics of VL and VM muscle force patterns were similar, novice runners showed greater muscle forces than experienced runners during initial contact (0~20.03%; $p<0.001$) (4.83~23.79%; $p<0.001$), then showed smaller muscle forces during mid-stance (26.36~78.29%; $p<0.001$) (33.00~79.39%; $p<0.001$). There were significant differences found in GM (38.17~57.24%; $p < 0.001$) (80.34~95.29%; $p<0.001$) and in TA (0~15.58%; $p<0.001$) (23.67~83.37%; $p<0.001$) between novice runners and experienced runners during the stance phase.

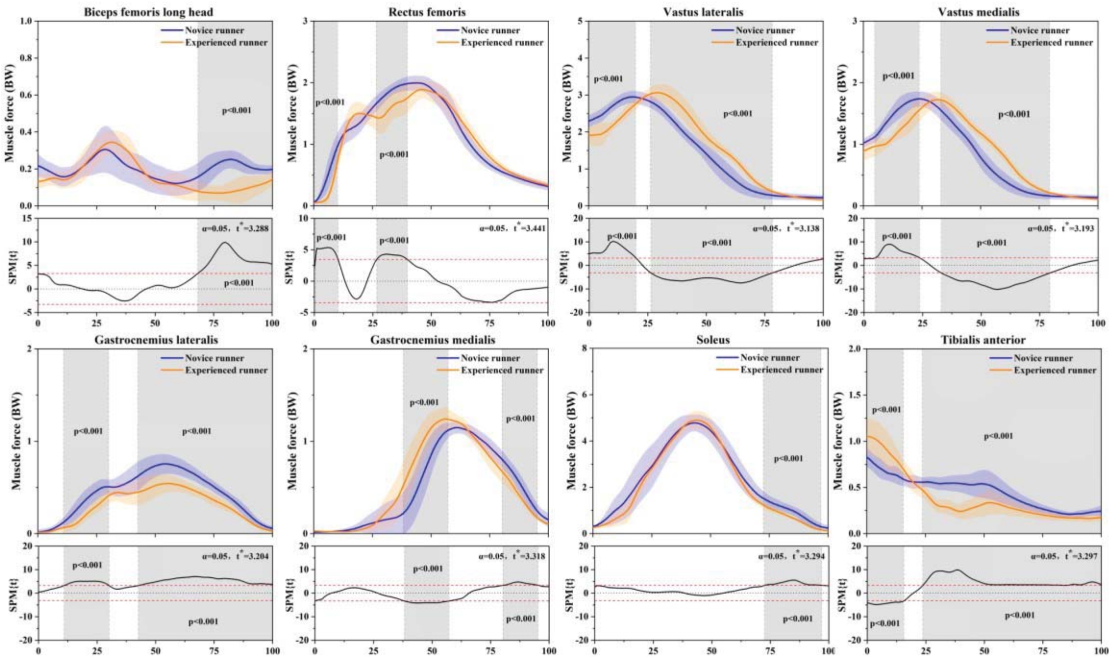

Figure 17 Normalized mean (SD) muscle forces for novice and experienced runners during the stance phase with SPM1d analyses.

Note: grey shades indicate differences between novice and experienced runners ($p<0.05$).

3.1.2 Knee reaction force

Figure 18 illustrated the knee joint forces for novice and experienced runners during the stance phase. Knee forces displayed statistically significant differences in medial-lateral direction (39.80~80.35%; $p<0.001$), anterior–posterior direction (4.37~22.22%; $p<0.001$) (47.27~84.69%; $p<0.001$) and superior-inferior direction (8.11~31.71%; $p=0.002$) (55.49~83.66%; $p<0.001$).

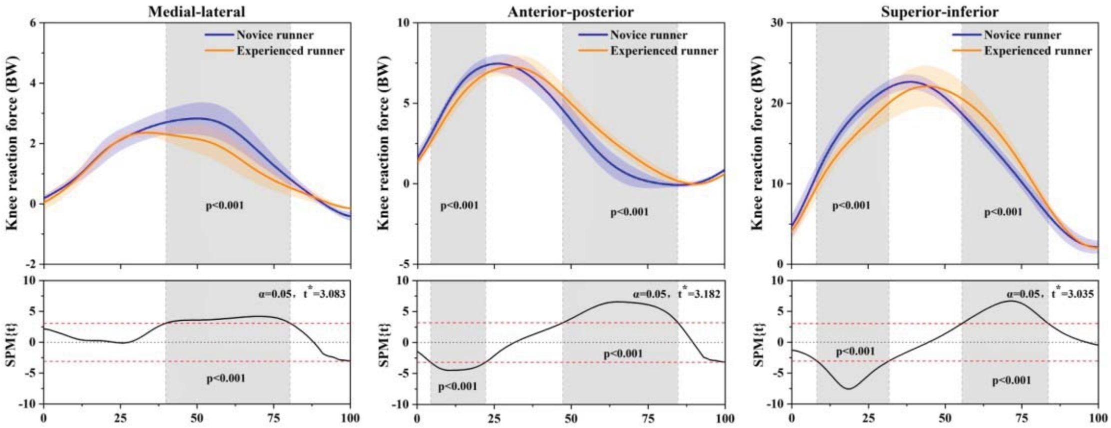

Figure 18 Normalized mean (SD) knee joint reaction forces for novice and experienced runners during the stance phase with SPM1d analyses.

Note: grey shades indicate differences between novice and experienced runners ($p<0.05$).

3.1.3 Correlations between muscle force and knee reaction force

Correlations between peak values of knee joint forces and muscle forces were presented in Figure 19-20. Among the novice group, there was a moderate correlation between knee medial-lateral force and RF ($R^2=0.424$), VL ($R^2=0.300$) and SO ($R^2=0.300$). The correlation of knee superior-inferior force also showed similar trends, there was also a moderate correlation in RF ($R^2=0.429$), VL ($R^2=0.385$) and SO ($R^2=0.322$). In the experienced group, the correlation between knee medial-lateral force and GL ($R^2=0.467$), knee anterior-posterior force and VL ($R^2=0.394$), knee superior-inferior force and GM ($R^2=0.433$) were moderate.

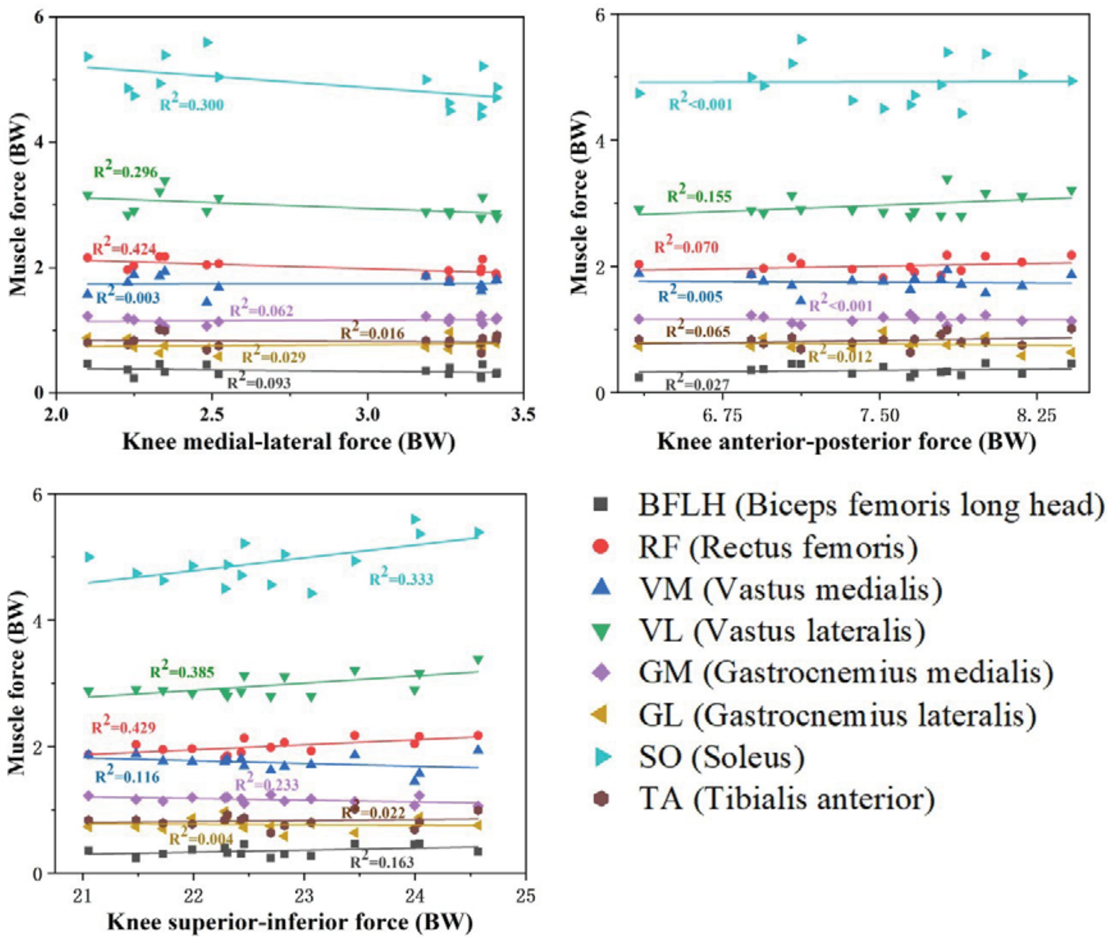

Figure 19 Correlations of peak knee reaction forces and peak muscle forces in novice runners

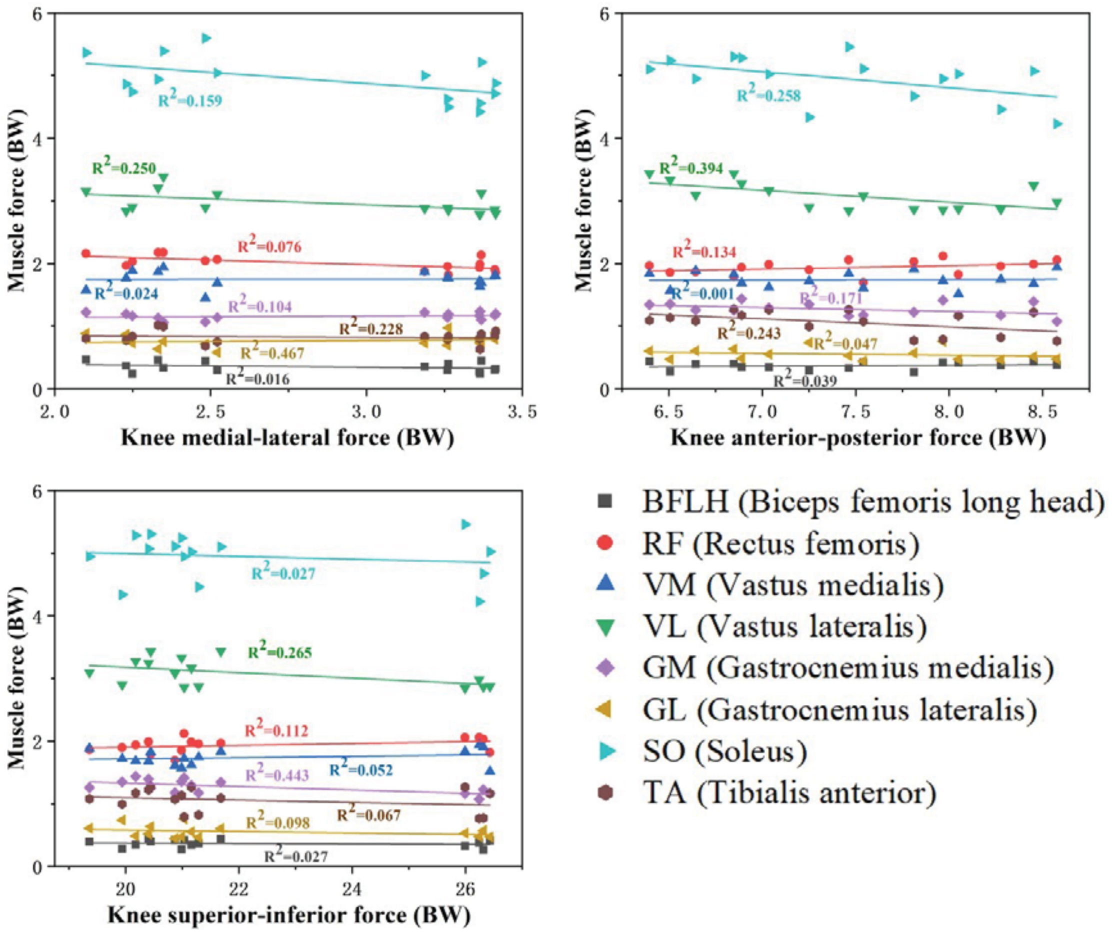

Figure 20 Correlations of peak knee reaction forces and peak muscle forces in experienced runners

3.2 Joint biomechanics of a 5 km run and running experience

3.2.1 Joint range of motion

The angle ROM of the ankle Dorsi/Plant, knee Ext/Flex, and hip Adduct/Abduct were significantly different between runners with and without running experience, where novice runners showed greater ankle Invert/Evert ROM ($p = 0.035$) and hip Adduct/Abduct ROM ($p < 0.001$), but smaller knee Ext/Flex ROM ($p < 0.001$) than experienced runners (Table 2). In both novice and experienced runners, the post-5 km running resulted in significant in ROM of the knee Adduct/Abduct ($p = 0.001$) and hip Adduct/Abduct ($p = 0.001$) compared to the pre-5 km running (Table 2). The interaction between the running experience and the 5 km run had a significant effect only on the angle ROM of the knee Adduct/Abduct ($p < 0.001$) (Table 2).

Table 2 Mean (SD) of joint range of motion (ROM) for novice and experienced runners of pre-5 km run and post-5 km run

Joint	ROM (°)	Novice/Pre	Novice/Post	Experienced/Pre	Experienced/Post	Runner Main Effect	5 km Main Effect	Interaction Effect
Ankle	Dorsi/Plant	44.80 (10.77)	41.71 (6.41)	44.55 (6.41)	45.41 (7.96)	F = 0.878; *p* = 0.355	F = 2.270; *p* = 0.139	F = 3.515; *p* = 0.078
	Invert/Evert	17.16 (4.82)	17.21 (4.75)	15.32 (2.91)	16.04 (1.91)	F = 4.720; *p* = 0.035	F = 1.442; *p* = 0.236	F = 1.104; *p* = 0.299
	Int Rot/Ext Rot	14.94 (2.97)	14.20 (1.50)	13.77 (2.41)	14.07 (2.84)	F = 1.978; *p* = 0.167	F = 0.720; *p* = 0.401	F = 4.024; *p* = 0.051
Knee	Ext/Flex	26.18 (4.05)	27.14 (3.26)	32.23 (3.55)	29.90 (2.94)	F = 57.932; *p* < 0.001	F = 1.941; *p* = 0.171	F = 5.917; *p* = 0.035
	Adduct/Abduct	2.85 (0.63)	3.90 (1.55)	3.38 (0.79)	3.43 (1.20)	F = 0.025; *p* = 0.876	F = 12.818; *p* = 0.001	F = 21.117; *p* < 0.001
	Int Rot/Ext Rot	6.62 (2.28)	6.70 (1.98)	7.73 (2.67)	7.73 (2.40)	F = 0.033; *p* = 0.857	F = 4.675; *p* = 0.057	F = 2.572; *p* = 0.090
Hip	Flex/Ext	43.17 (3.12)	42.81 (3.05)	41.98 (3.91)	43.12 (5.41)	F = 0.503; *p* = 0.482	F = 0.676; *p* = 0.415	F = 5.406; *p* = 0.025
	Adduct/Abduct	14.10 (3.66)	14.76 (4.68)	10.37 (1.90)	12.00 (1.22)	F = 23.459; *p* < 0.001	F = 13.369; *p* = 0.001	F = 2.967; *p* = 0.092
	Int Rot/Ext Rot	10.96 (4.44)	12.66 (6.26)	10.48 (3.31)	10.69 (2.61)	F = 1.378; *p* = 0.247	F = 6.664; *p* = 0.013	F = 5.682; *p* = 0.022

Note: Dorsi/Plant = dorsiflexion/plantarflexion, Invert/Evert = inversion/eversion, Int Rot/Ext Rot = internal rotation/external rotation, Ext/Flex = extension/flexion, Adduct/Abduct = adduction/abduction, Flex/Ext = flexion/extension. Significant difference ($p < 0.05$). The significant differences in interaction effect were determined using Bonferroni corrections ($\alpha = 0.008$).

3.2.2 Joint angles

The p-values of PC scores of joint angles are provided in Table 3. PC score statistical analysis of joint angles showed that differences were found between novice runners and experienced runners with respect to PC2 in the ankle Dorsi/Plant, and PC1 and PC2 in the ankle Invert/Evert. Statistical differences in ankle angle PC scores of the 5 km run were found in PC2 in the ankle Invert/Evert, and PC1 and PC3 in the ankle Int Rot/Ext Rot. The waveforms, PC loading vectors, and reconstructed waveforms of ankle angles are presented in Figure 21. For each variable, the waveforms were reconstructed by utilizing the scores and coefficients of the retained PCs; the high PC and low PC can be used to visually understand differences in amplitude. Experienced runners demonstrated significantly less ankle inversion angle than novice runners, which was also consistent with lower PC1 and PC2 scores than experienced runners in ankle Invert/Evert, and this magnitude difference was obvious throughout the running stance phase.

Table 3 Mean (SD) of joint angles for all principal components (PCs) retained according to the 90% trace criterion.

Joint	Angle	PC	Variance Explained (%)	Mean (SD) PC scores				Runner p Value	5 km p Value	Interaction p Value
				Novice/Pre	Novice/Post	Experienced/ Pre	Experienced/ Post	Main Effect	Main Effect	Effect
Ankle	Dorsi/Plant	1	47.67	0.77 (9.57)	-0.68 (7.34)	-0.88 (3.05)	0.79 (6.21)	0.950	0.868	0.012
		2	32.94	-1.23 (5.46)	3.67 (6.02)	0.29 (4.63)	-2.72 (4.99)	0.030	0.192	0.025
		3	13.91	0.69 (1.76)	-0.42 (1.90)	-0.16 (4.61)	-0.11 (5.34)	0.674	0.176	0.177
	Invert/Evert	1	73.62	-2.86 (7.75)	-5.02 (7.66)	3.90 (7.92)	3.98 (7.15)	<0.001	0.117	0.100
		2	20.65	2.12 (3.47)	0.97 (3.18)	-1.43 (4.04)	-1.67 (5.99)	<0.001	0.014	0.100
	Int Rot/Ext Rot	1	64.56	-2.25 (6.78)	3.92 (7.02)	-2.85 (8.60)	1.18 (8.09)	0.284	<0.001	0.206
		2	14.51	1.34 (4.08)	-1.48 (3.92)	-0.79 (3.43)	0.93 (3.20)	0.845	0.126	0.011
		3	11.04	-0.18 (3.33)	0.46 (4.61)	-0.87 (2.85)	0.60 (1.87)	0.705	0.007	0.217
Knee	Ext/Flex	1	66.67	-0.53 (8.81)	-1.41 (9.54)	-0.75 (8.67)	2.69 (4.44)	0.213	0.209	0.023
		2	19.17	-1.98 (4.42)	-1.42 (3.96)	3.07 (4.44)	0.33 (3.26)	<0.001	0.020	0.010
		3	9.81	-2.12 (3.04)	-1.63 (3.07)	2.73 (1.32)	1.01 (1.97)	<0.001	0.093	<0.001
	Adduct/Abduct	1	85.74	-2.72 (8.00)	2.63 (12.66)	1.98 (4.49)	-1.89 (9.25)	0.964	0.352	0.017
		2	6.67	0.07 (1.64)	-2.09 (2.87)	0.25 (2.50)	1.76 (1.59)	<0.001	0.148	0.021
	Int Rot/Ext Rot	1	90.20	4.95 (5.79)	2.98 (7.02)	-2.30 (5.31)	-5.64 (13.69)	<0.001	0.018	0.416
Hip	Flex/Ext	1	83.65	0.70 (6.10)	-5.84 (11.02)	4.89 (5.88)	0.24 (9.29)	0.007	<0.001	0.327
		2	10.23	-0.98 (2.27)	-0.46 (2.63)	1.26 (3.04)	0.18 (4.22)	0.003	0.452	0.040
	Adduct/Abduct	1	86.23	14.10 (3.66)	14.76 (4.68)	10.37 (1.90)	12.00 (1.22)	<0.001	0.001	0.092
		2	9.29	-0.56 (3.25)	0.29 (4.56)	0.85 (1.70)	-0.58 (1.55)	0.655	0.372	0.028
	Int Rot/Ext Rot	1	75.65	2.85 (8.50)	0.32 (9.08)	-4.22 (6.00)	1.05 (9.60)	0.035	0.095	0.001
		2	16.76	-0.13 (4.20)	-0.96 (5.16)	0.08 (3.57)	1.01 (3.14)	0.256	0.829	0.023

Note: Dorsi/Plant = dorsiflexion/plantarflexion, Invert/Evert = inversion/eversion, Int Rot/Ext Rot = internal rotation/external rotation, Ext/Flex = extension/flexion, Adduct/Abduct = adduction/abduction, Flex/Ext = flexion/extension. Significant difference ($p < 0.05$). The significant differences in interaction effect were determined using Bonferroni corrections ($\alpha = 0.008$).

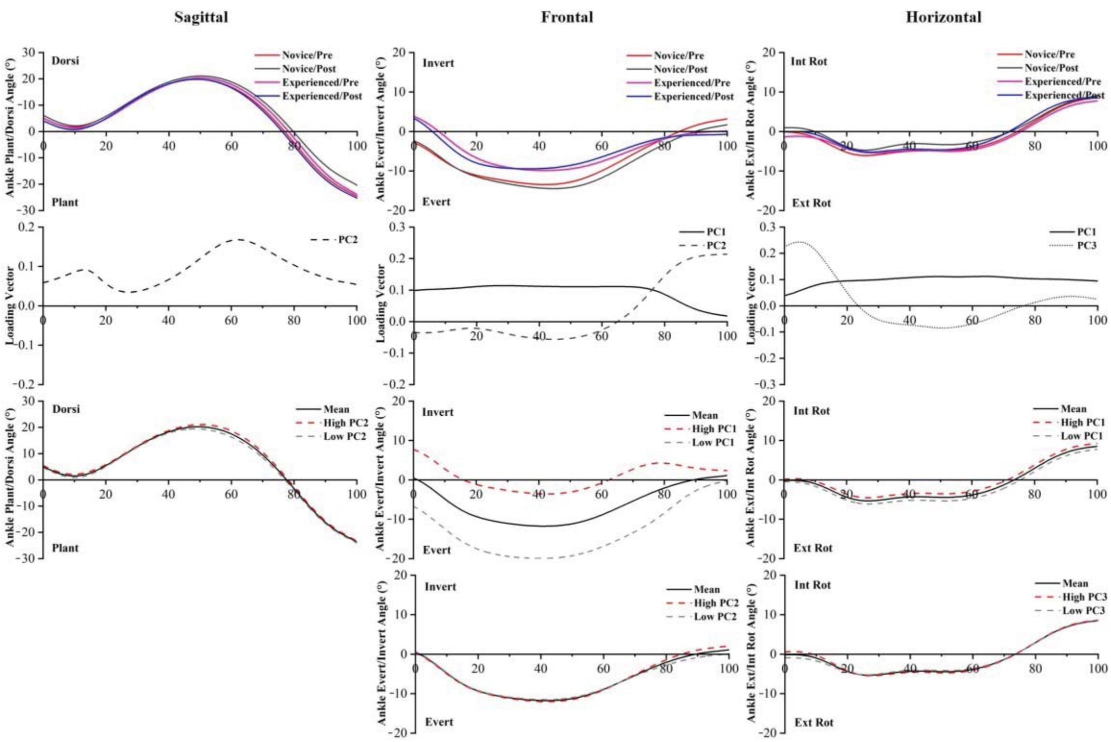

Figure 21 The mean of ankle angles for novice and experienced runners with a 5 km run during stance phase. The loading vectors for PC scores with significant differences. Single-component reconstruction for PC scores of 3D ankle angles.

PC score differences in knee angles between runners were found in PC2 and PC3 in the knee Ext/Flex, PC2 in the knee Adduct/Abduct, and PC1 in the knee Int Rot/Ext Rot. Statistical differences in knee angle PC scores between pre-5 km running and post-5 km running were found in PC2 in the knee Adduct/Abduct and PC1 in the knee Int Rot/Ext Rot. Compared to experienced runners, novice runners showed significantly more knee flexion angle in the early stance and more internal rotation angle throughout the running stance phase through visual inspection and PC scores. Post-5 km running showed less knee internal rotation angle than pre-5 km running (Figure 22, Table 3).

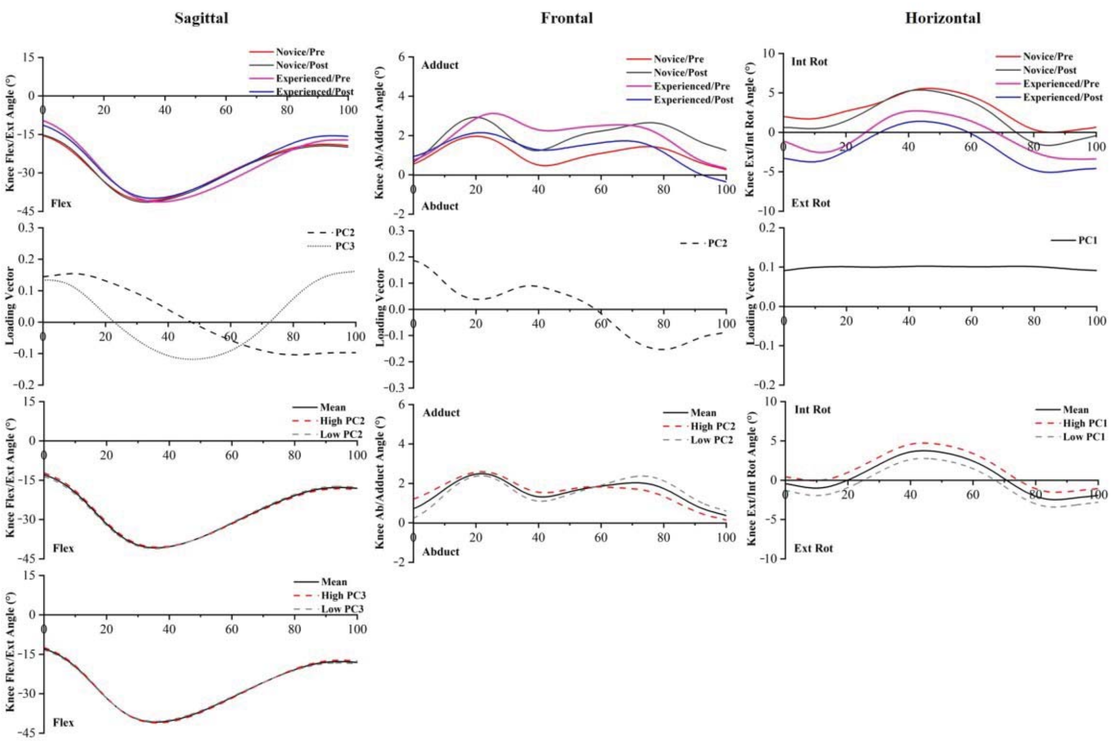

Figure 22 The mean of knee angles for novice and experienced runners with a 5 km run during stance phase. The loading vectors for PC scores with significant differences. Single-component reconstruction for PC scores of 3D knee angles.

PC score differences in hip angles between runners were found in PC1 and PC2 in the hip Flex/Ext, PC1 in the hip Adduct/Abduct, and PC1 in the hip Int Rot/Ext Rot. The effects of the 5 km run existed in PC1 in the hip Flex/Ext and PC1 in the hip Adduct/Abduct (Table 3). During the running stance phase, novice runners had significantly greater hip adduction and internal rotation angle than experienced runners. Meanwhile, post-5 km running showed a larger hip adduction angle (Figure 23). The interaction effects existed in PC3 in the knee Ext/Flex and PC1 in the hip Int Rot/Ext Rot.

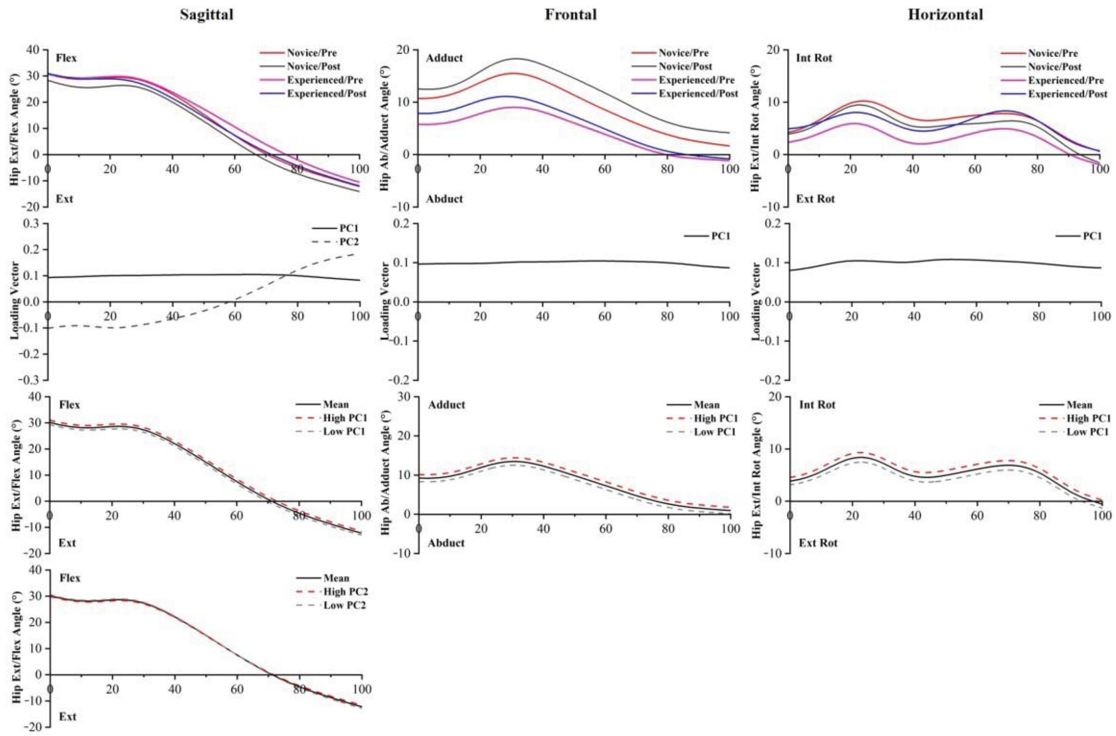

Figure 23 The mean of hip angles for novice and experienced runners with a 5 km run during stance phase. The loading vectors for PC scores with significant differences. Single-component reconstruction for PC scores of 3D knee angles.

3.2.3 Joint moments

The p-values of PC scores of joint moments are provided in Table 4. The analysis showed significant runner main effects in PC3 in the ankle Dorsi/Plant, PC1 and PC2 in the ankle Invert/Evert, and PC1 and PC3 in the ankle Int Rot/Ext Rot. The significant 5 km running main effects of ankle moments were found in PC1, PC2, and PC3 in the ankle Dorsi/Plant, and PC3 in the ankle Int Rot/Ext Rot. Compared to experienced runners, novice runners showed significantly larger ankle inversion moment and internal rotation moment throughout the stance phase. After 5 km of running, the ankle plantarflexion moment was smaller during the middle and later stances (Figure 24).

Table 4 Mean (SD) of joint moments for all principal components (PCs) retained according to the 90% trace criterion.

Joint	Moment	PC	Variance Explained (%)	Mean (SD) PC scores				Runner p Value	5 km p Value	Interaction p Value
				Novice/Pre	Novice/Post	Experienced/ Pre	Experienced/ Post	Main Effect	Main Effect	Effect
Ankle	Dorsi/Plant	1	50.03	-1.25 (9.46)	2.46 (7.90)	-0.75 (5.82)	-0.46 (3.34)	0.369	0.001	0.013
		2	27.92	-1.24 (4.74)	1.19 (6.02)	-2.81 (3.74)	2.85 (4.76)	0.966	<0.001	0.020
		3	10.50	1.04 (2.40)	0.09 (2.65)	-0.29 (3.70)	-0.84 (3.85)	0.017	<0.001	0.576
		4	4.96	1.02 (2.23)	-0.63 (2.13)	0.54 (2.23)	-0.93 (1.81)	0.334	0.051	0.632
	Invert/Evert	1	67.39	3.85 (7.53)	4.40 (7.42)	-5.04 (7.05)	-3.21 (6.63)	<0.001	0.197	0.579
		2	19.20	-1.09 (3.90)	-0.83 (4.87)	-0.87 (2.96)	2.79 (4.53)	<0.001	0.072	0.025
		3	6.21	0.20 (3.79)	0.08 (2.27)	0.03 (1.38)	-0.32 (2.00)	0.481	0.483	0.717
	Int Rot/Ext Rot	1	62.80	2.14 (7.29)	1.10 (8.74)	-3.48 (7.58)	0.24 (6.66)	0.007	0.062	0.001
		2	15.69	0.56 (3.53)	0.53 (2.87)	-1.41 (3.96)	0.31 (4.13)	0.143	0.067	0.011
		3	9.59	0.57 (3.62)	0.71 (2.41)	-1.73 (2.82)	0.44 (2.76)	0.007	0.001	0.016
		4	5.54	-0.36 (1.86)	0.01 (2.33)	0.10 (1.89)	0.25 (3.18)	0.254	0.313	0.671
Knee	Ext/Flex	1	37.17	-1.72 (4.67)	-0.98 (5.05)	2.59 (7.33)	0.11 (6.38)	0.053	0.091	0.062
		2	26.09	1.42 (5.23)	2.04 (3.86)	-3.39 (5.19)	-0.07 (4.46)	0.001	<0.001	0.013
		3	19.17	1.30 (3.86)	2.20 (2.01)	-2.53 (5.13)	-0.97 (4.39)	<0.001	0.004	0.345
		4	7.62	0.60 (2.16)	-1.63 (2.79)	0.44 (2.99)	0.59 (2.50)	0.058	0.072	0.031
	Adduct/Abduct	1	49.62	-0.67 (7.60)	-0.36 (6.36)	-4.08 (5.36)	5.11 (5.72)	0.459	<0.001	<0.001
		2	18.60	1.92 (3.32)	2.49 (4.00)	-3.16 (3.81)	-1.25 (3.60)	<0.001	<0.001	0.112
		3	10.16	0.34 (1.99)	-0.18 (2.52)	-0.81 (3.88)	0.65 (3.89)	0.768	0.174	0.011
		4	8.14	-1.11 (2.61)	-0.29 (1.69)	0.16 (1.95)	1.24 (2.97)	0.001	<0.001	0.652
		5	5.52	-0.33 (2.28)	0.15 (1.68)	-0.19 (2.01)	0.37 (2.32)	0.646	0.064	0.861
	Int Rot/Ext Rot	1	48.08	-0.98 (2.93)	-3.78 (4.97)	5.29 (7.05)	-0.53 (5.83)	<0.001	<0.001	0.023
		2	18.42	-1.88 (4.30)	-1.67 (4.51)	3.08 (3.69)	0.47 (2.63)	<0.001	0.053	0.015
		3	13.18	1.01 (2.10)	1.69 (2.96)	-0.78 (4.03)	-1.92 (2.85)	<0.001	0.387	0.009
		4	11.54	0.84 (2.47)	0.55 (3.44)	0.27 (2.60)	-1.66 (3.25)	0.067	0.050	0.013

Hip	Flex/Ext	1	51.64	4.60 (4.86)	3.17 (6.92)	-5.11 (6.07)	-2.66 (5.07)	<0.001	0.457	<0.001
		2	19.41	-1.44 (5.40)	-0.07 (3.22)	0.89 (3.34)	0.62 (3.67)	0.058	0.279	0.050
		3	11.79	0.48 (2.28)	-0.63 (2.62)	0.40 (2.80)	-0.25 (3.35)	0.764	0.050	0.546
		4	6.33	-0.51 (1.83)	-0.17 (2.50)	0.27 (2.10)	0.41 (3.39)	0.082	0.439	0.639
	Adduct/Abduct	1	45.07	-0.66 (5.94)	-0.27 (8.47)	-1.35 (3.40)	2.27 (4.56)	0.393	0.051	0.024
		2	22.64	1.24 (4.06)	1.24 (4.16)	-3.79 (4.87)	1.31 (1.89)	0.003	<0.001	0.031
		3	15.04	0.19 (1.93)	0.97 (3.29)	-0.28 (4.29)	-0.89 (5.19)	0.055	0.775	0.010
		4	8.14	1.94 (2.62)	0.93 (2.61)	-1.39 (2.07)	-1.48 (2.56)	<0.001	0.059	0.150
	Int Rot/Ext Rot	1	53.41	0.95 (8.29)	1.52 (7.21)	-3.48 (6.36)	1.00 (5.41)	0.056	0.085	0.009
		2	23.35	-1.72 (3.28)	-1.50 (3.25)	0.77 (4.21)	2.45 (5.33)	<0.001	0.057	0.028
		3	7.92	-0.79 (2.17)	-0.42 (2.72)	0.74 (1.78)	0.46 (3.96)	0.051	0.898	0.229
		4	6.50	-0.81 (1.61)	-0.47 (3.64)	0.24 (2.02)	1.04 (2.18)	0.012	0.081	0.379

Note: Dorsi/Plant = dorsiflexion/plantarflexion, Invert/Evert = inversion/eversion, Int Rot/Ext Rot = internal rotation/external rotation, Ext/Flex = extension/flexion, Adduct/Abduct = adduction/abduction, Flex/Ext = flexion/extension. Significant difference ($p < 0.05$). The significant differences in interaction effect were determined using Bonferroni corrections ($\alpha = 0.008$).

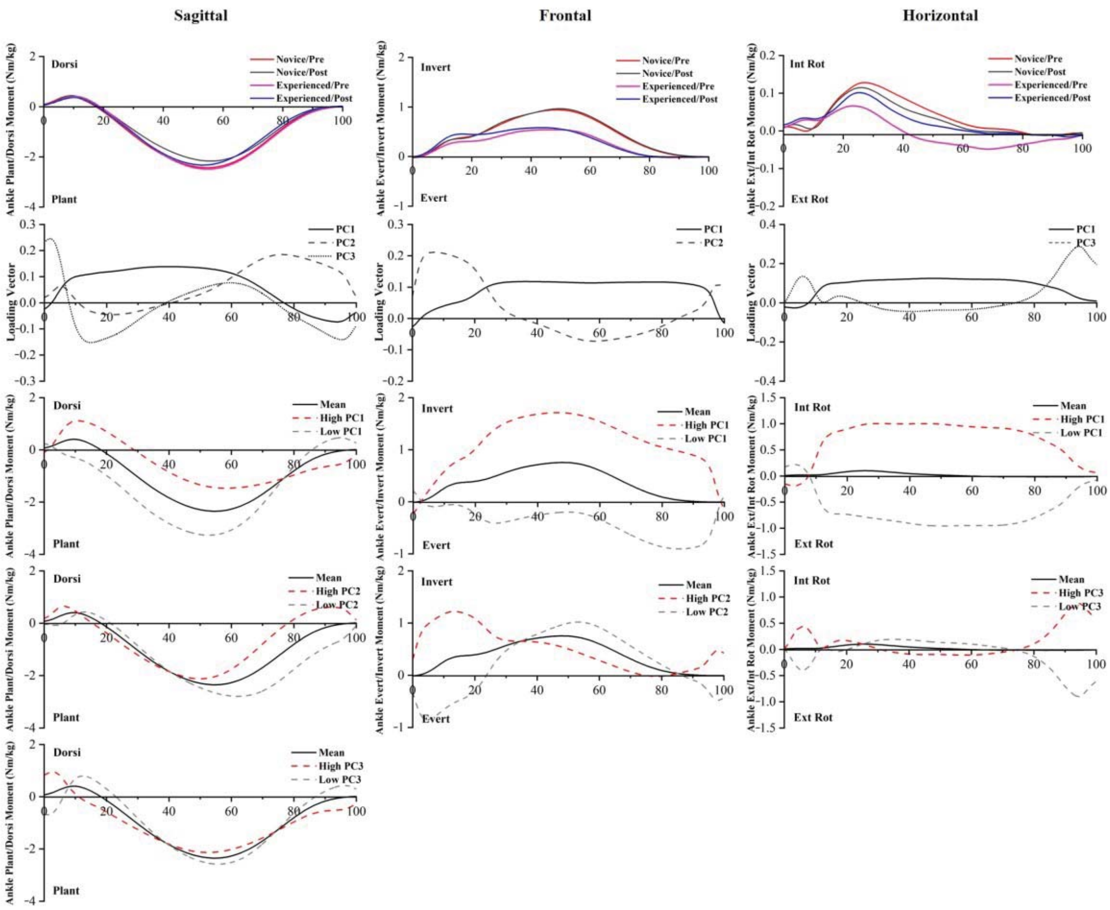

Figure 24 The mean of ankle moments for novice and experienced runners with a 5 km run during stance phase. The loading vectors for PC scores with significant differences. Single-component reconstruction for PC scores of 3D ankle moments.

PC score differences in knee moments were found between runners in PC2 and PC3 in the Ext/Flex, PC2 and PC4 in the Adduct/Abduct, and PC1, PC2, and PC3 in the Int Rot/Ext Rot. The significant 5 km running main effects of knee moments were found in PC2 and PC3 in the knee Ext/Flex, PC1, PC2, and PC4 in the knee Adduct/Abduct, and PC1 in the knee Int Rot/Ext Rot. Knee moment-related waveforms are presented in Figure 25.

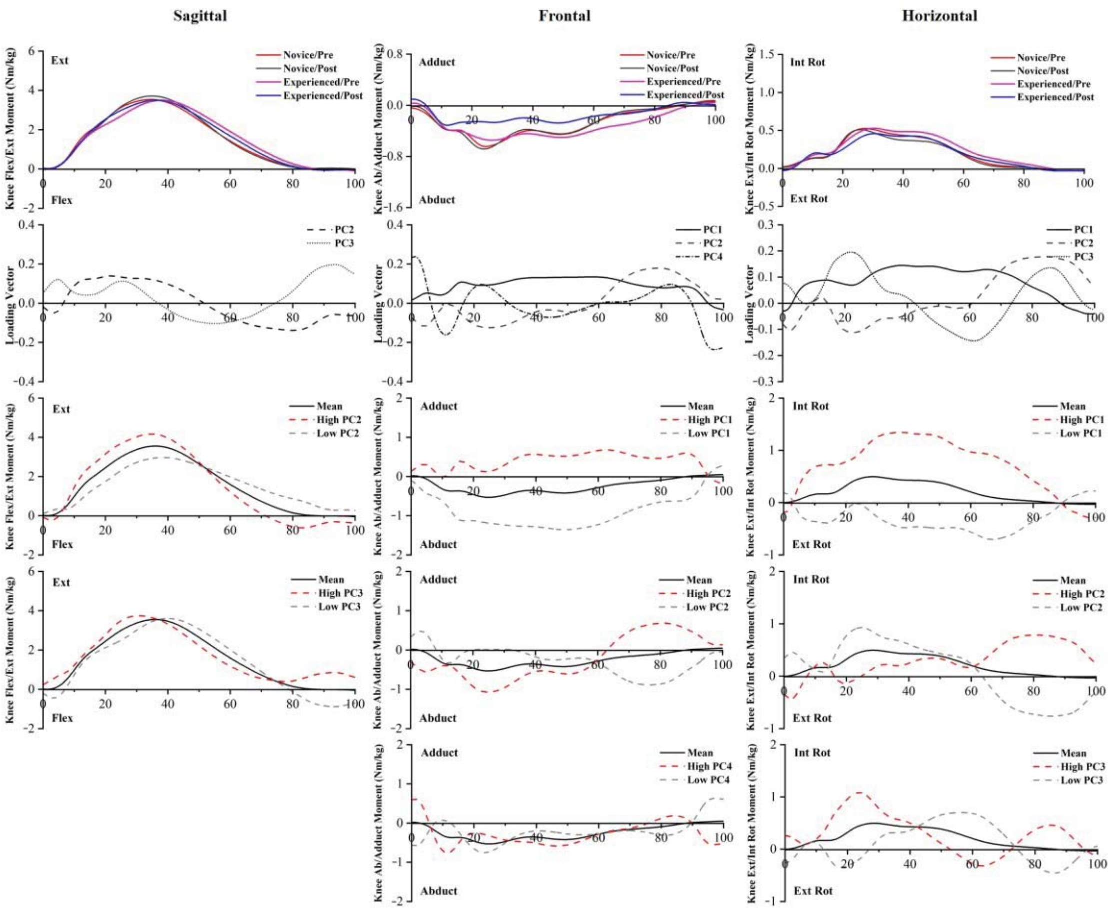

Figure 25 The mean of knee moments for novice and experienced runners with a 5 km run during stance phase. The loading vectors for PC scores with significant differences. Single-component reconstruction for PC scores of 3D knee moments.

Hip moment PC score differences between runners were found in PC1 in the hip Flex/Ext, PC2 and PC4 in the hip Adduct/Abduct, and PC2 in the hip Int Rot/Ext Rot. The significant 5 km running main effects of hip moments were found in PC2 in the hip Adduct/Abduct. Compared to experienced runners, novice runners showed significantly greater hip flexion moment throughout the running stance phase and greater external rotation moment in the early phase (Figure 26, Table 4). The interaction effects existed in PC1 in the ankle Int Rot/Ext Rot, PC1 in the knee Adduct/Abduct, and PC1 in the hip Flex/Ext.

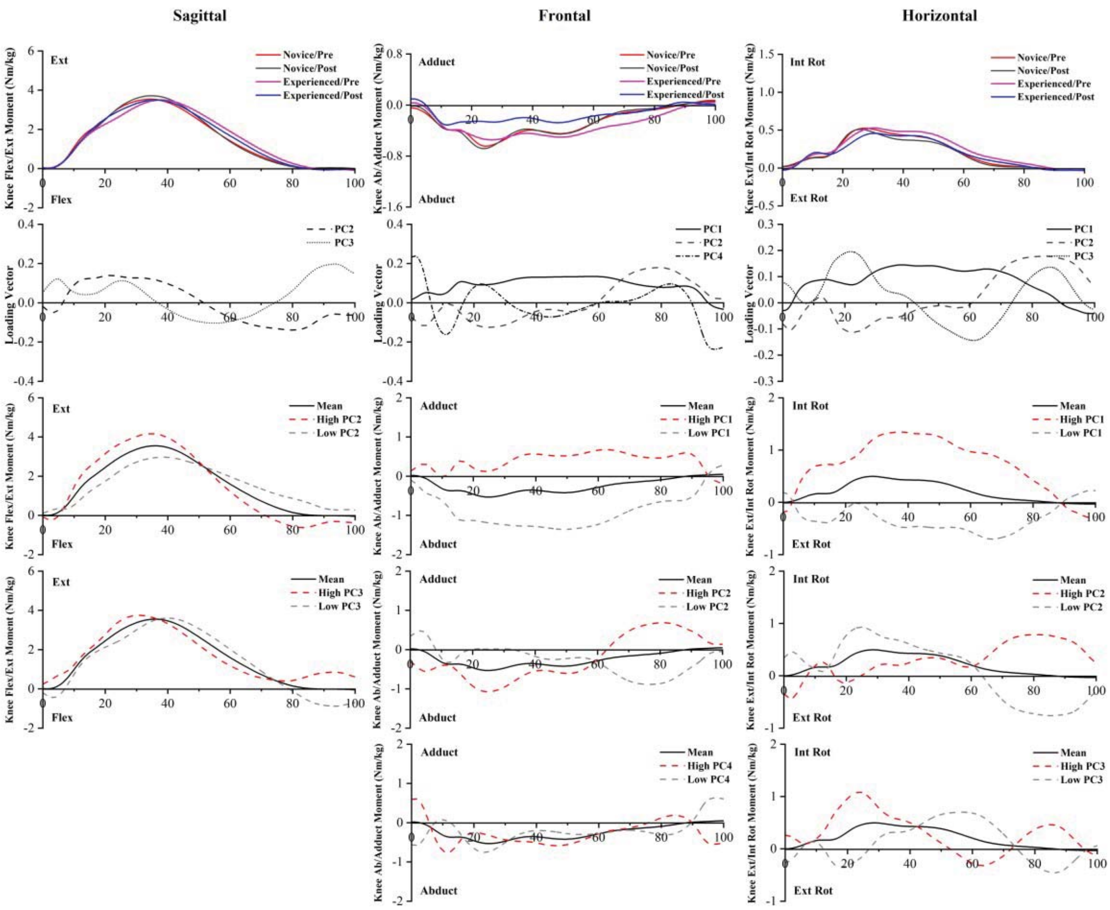

Figure 26 The mean of hip moments for novice and experienced runners with a 5 km run during stance phase. The loading vectors for PC scores with significant differences. Single-component reconstruction for PC scores of 3D hip moments.

3.3 Ground reaction force of running speed and runners' gender

3.3.1 Discrete GRF variables

The relationship between discrete GRF variables and running speed is detailed in Figure 27. In female runners, seven variables showed significant correlations with running speed. Specifically, braking impulse ($r = 0.617$, $p < 0.001$), propulsive impulse ($r = 0.568$, $p < 0.001$), peak propulsive force ($r = 0.822$, $p < 0.001$), VALR ($r = 0.687$, $p < 0.001$) and peak vertical impact force ($r = 0.702$, $p < 0.001$) increased linearly with running speed, whereas vertical impulse ($r = -0.814$, $p < 0.001$) and peak vertical active force ($r = -0.205$, $p = 0.021$) decreased linearly. In male runners, six variables were significantly correlated with running speed. Peak propulsive force ($r = 0.627$, $p < 0.001$), VALR ($r = 0.639$, $p < 0.001$) and peak vertical impact force ($r = 0.691$, $p < 0.001$)

increased linearly with speed, whereas the braking impulse (r = -0.177, p = 0.010), peak braking force (r = -0.597, p < 0.001) and vertical impulse (r = -0.707, p < 0.001) decreased linearly.

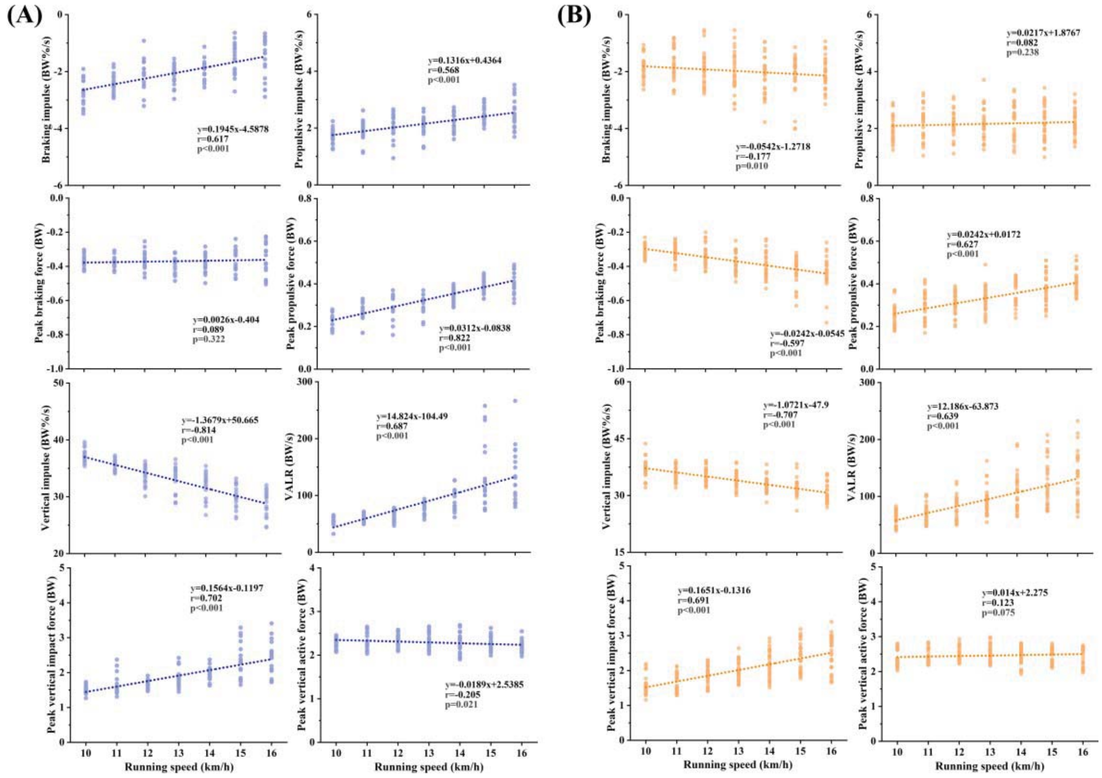

Figure 27 Correlation between running speed and ground reaction force (GRF) variables: braking impulse, propulsive impulse, peak braking force, peak propulsive force, vertical impulse, vertical average loading rate (VALR), peak vertical impact force and active force for (A) female runners, and (B) male runners.

The results of stepwise linear regression analysis for females and males are showed in Table 5 and Table 6, respectively. For females, the analysis identified peak propulsive force, peak vertical impact force, propulsive impulse, VALR and vertical impulse as the best predictors of the running speed ($R^2 = 0.901$, $p < 0.001$), explaining 90% of the variation. For males, the best predictors were vertical impulse, peak vertical impact force, peak propulsive force, braking impulse, VALR and peak braking force ($R^2 = 0.855$, $p < 0.001$), accounting for 85.5% of the variance in running speed.

Table 5 Results of stepwise linear regression for running speed in female runners

Variables	R	R2	Adjusted R2	F	p
Peak propulsive force	0.820	0.673	0.670	255.245	<0.001
Peak propulsive force + peak vertical impact force	0.917	0.842	0.839	326.931	<0.001
Peak propulsive force + peak vertical impact force + propulsive impulse	0.934	0.872	0.869	276.457	<0.001
Peak propulsive force + peak vertical impact force + propulsive impulse + VALR	0.942	0.888	0.884	240.288	<0.001
Peak propulsive force + peak vertical impact force + propulsive impulse + VALR + vertical impulse	0.949	0.901	0.897	217.968	<0.001

Table 6 Results of stepwise linear regression for running speed in male runners

Variables	R	R^2	Adjusted R^2	F	p
Vertical impulse	0.707	0.499	0.497	207.322	<0.001
Vertical impulse + peak vertical impact force	0.830	0.689	0.686	229.037	<0.001
Vertical impulse + peak vertical impact force + peak propulsive force	0.868	0.754	0.750	210.155	<0.001
Vertical impulse + peak vertical impact force + peak propulsive force + braking impulse	0.914	0.835	0.832	258.860	<0.001
Vertical impulse + peak vertical impact force + peak propulsive force + braking impulse + VALR	0.920	0.847	0.843	225.883	<0.001
Vertical impulse + peak vertical impact force + peak propulsive force + braking impulse + VALR + peak braking force	0.925	0.855	0.851	200.043	<0.001

3.3.2 Time varying GRF variables

SPM1d analyses revealed significant main effects of speed on the anterior-posterior and vertical GRF waveforms for female runners, as shown in Figure 28. Both propulsive force (45% – 98%, $p < 0.001$) and vertical force (1% – 18%, $p < 0.001$; 60% – 88%, $p < 0.001$) increased with running speed. Notably, peak propulsive force, propulsive impulse, braking impulse, vertical impulse, VALR, and peak vertical impact force all demonstrated significant speed main effects. Similarly, for male runners, Figure 29 indicates significant main effects of speed on their anterior-posterior and vertical GRF waveforms. Increased running speed resulted in greater braking force (12% – 47%, $p < 0.001$), propulsive force (67% – 98%, $p < 0.001$), and vertical force (7% – 23%, $p < 0.001$; 47% – 95%, $p < 0.001$). Significant main effects of speed were also found in male runners' peak propulsive force, peak braking force, vertical impulse, VALR, and peak vertical impact force.

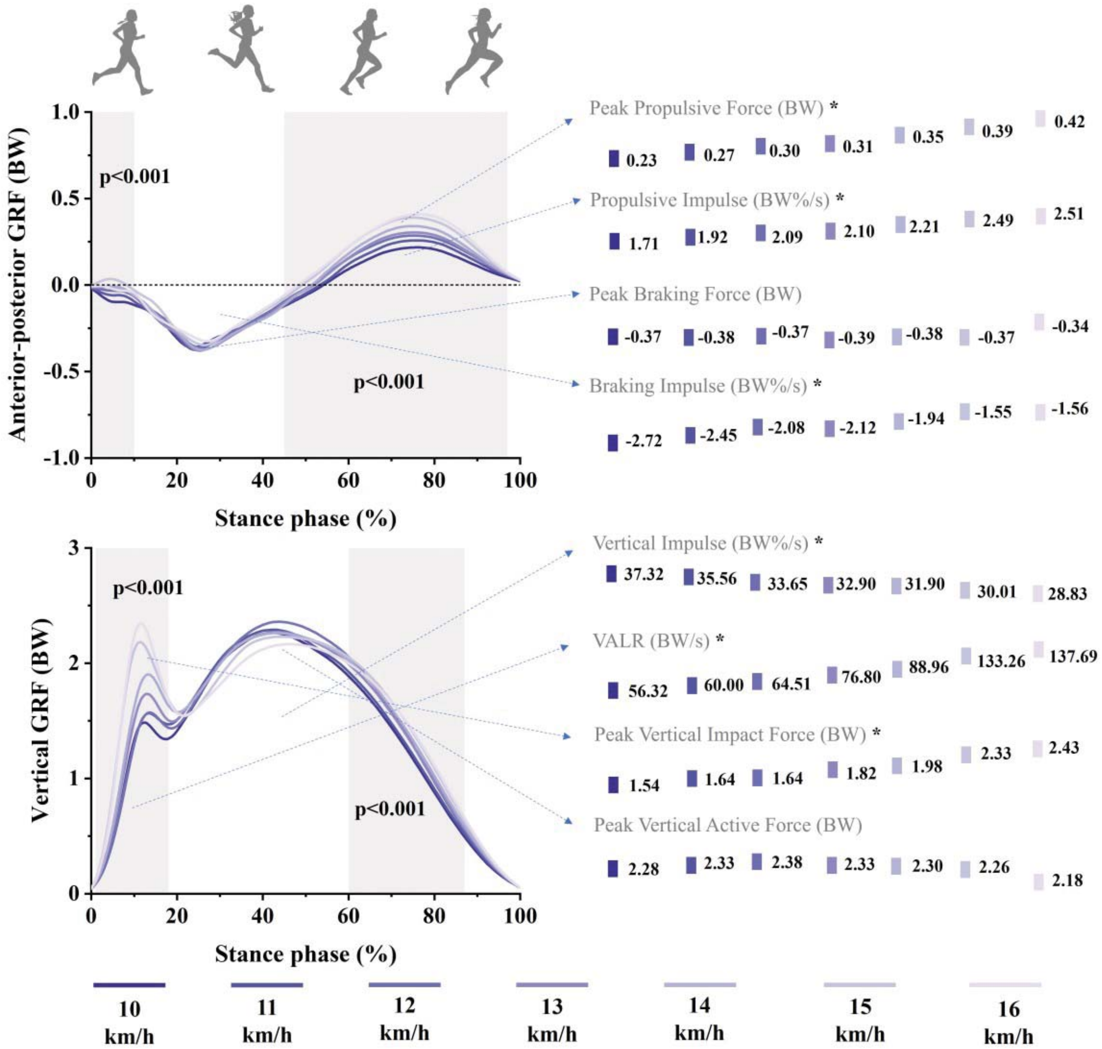

Figure 28 Mean anterior-posterior and vertical GRF waveforms across seven running speeds: 10, 11, 12, 13, 14, 15, 16 km/h for female runners during stance phase.

Note: standard deviations are not presented for further clarity. The grey shaded areas represent significant main effects of running speed from SPM analyses ($p < 0.05$). Point graphs in the figure illustrate mean values of specific GRF parameters at each of the seven speeds. Asterisks indicate significant differences across running speeds ($p < 0.05$).

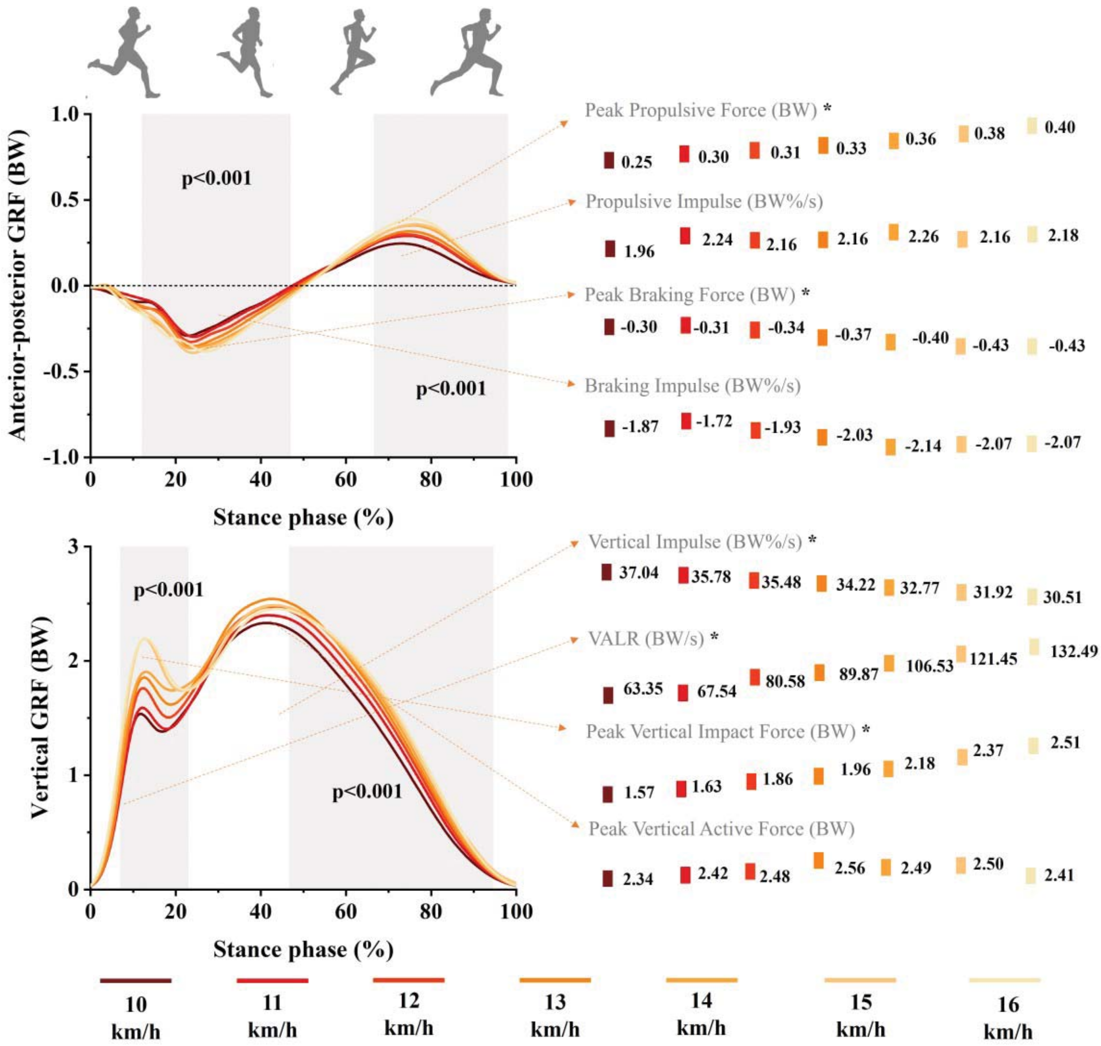

Figure 29 Mean anterior-posterior and vertical GRF waveforms across seven running speeds: 10, 11, 12, 13, 14, 15, 16 km/h for male runners during stance phase.

Note: standard deviations are not presented for further clarity. The grey shaded areas represent significant main effects of running speed from SPM analyses ($p < 0.05$). Point graphs in the figure illustrate mean values of specific GRF parameters at each of the seven speeds. Asterisks indicate significant differences across running speeds ($p < 0.05$).

Figure 30 presents the results of gender differences in anterior-posterior GRFs at each running speed, as determined by independent t-tests and SPM analyses. Females exhibited a larger braking force at speeds of 10 km/h (13% – 57%, $p < 0.001$) and 11 km/h (12% – 58%, $p < 0.001$). During the later stance phase, females demonstrated more propulsive force than males at all tested speeds. Specifically, this increase was observed during 93% – 100% of the stance phase at 10 km/h ($p = 0.003$), 94% – 100% at 11 km/h ($p=0.012$), 92% – 100% at 12 km/h ($p = 0.004$), 94% – 100% at 13 km/h ($p = 0.015$), 93% – 100% at 14 km/h ($p = 0.008$), 89% – 100% at 15 km/h ($p < 0.001$), and 84% – 100% at 16 km/h ($p < 0.001$). Figure 31 illustrates gender differences in vertical GRFs at each running speed. During the later stance phase, females exhibited higher forces than males at all selected speeds. Specifically, this increase was observed during 70% – 100% of the stance phase at 10 km/h ($p < 0.001$), 84% – 100% at 11 km/h ($p < 0.001$), 83% – 100% at 12 km/h ($p < 0.001$), 86% – 100% at 13 km/h ($p < 0.001$), 86% – 100% at 14 km/h ($p < 0.001$), 79% – 100% at 15 km/h ($p < 0.001$), and 86% – 100% at 16 km/h ($p < 0.001$). However, males showed larger force during 34% – 56% of the stance phase at 13 km/h ($p < 0.001$), 31% – 51% at 15 km/h ($p < 0.001$), and 31% – 53% at 16 km/h ($p < 0.001$), which include peak vertical active force.

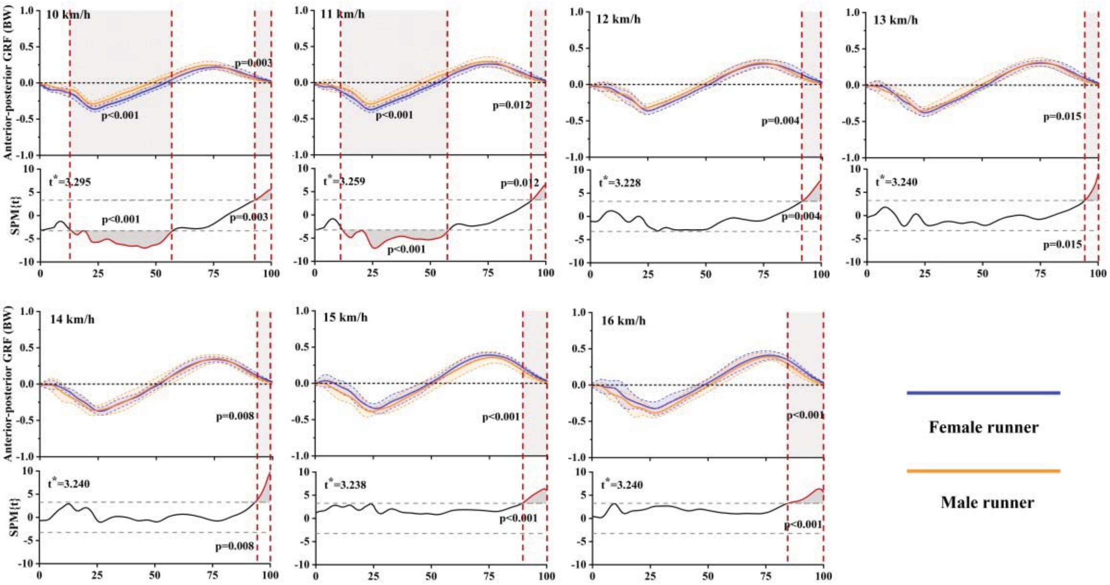

Figure 30 Mean (SD) anterior-posterior GRF waveforms for both female and male runners at each running speed, accompanied by the SPM results.

Note: grey shaded areas represent significant differences between female and male runners during the running stance phase ($p < 0.05$).

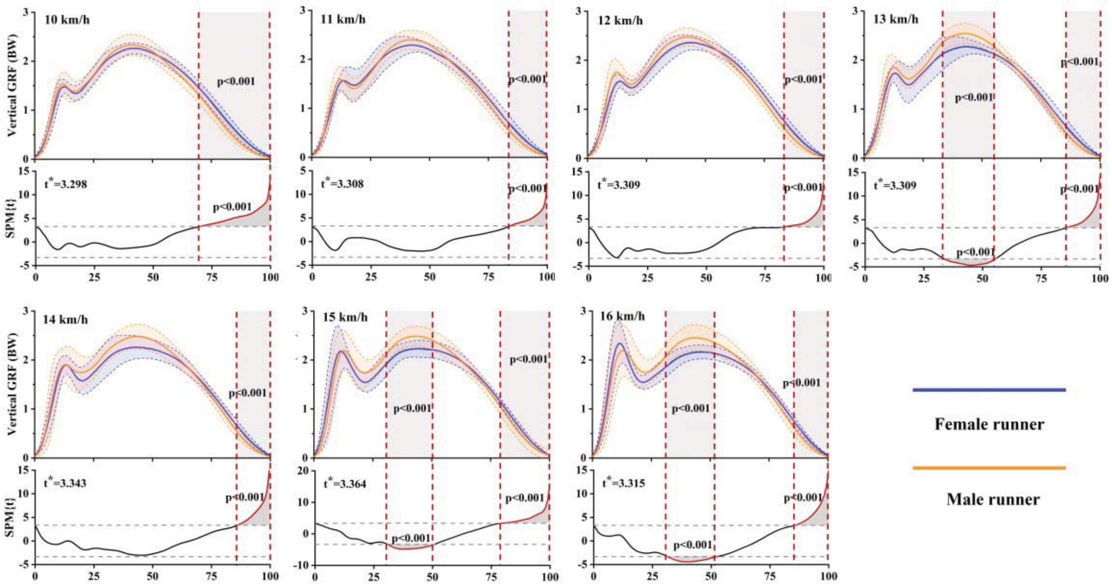

Figure 31 Mean (SD) vertical GRF waveforms for both female and male runners at each running speed, accompanied by the SPM results.

Note: grey shaded areas represent significant differences between female and male runners during the running stance phase ($p < 0.05$).

4 DISCUSSION

4.1 Muscle force and knee reaction force between novice and experienced runners

The purposes of this study were to compare lower limb muscle forces and knee forces between runners with different experiences using OpenSim software, as well as investigate the association between muscle forces and knee forces. Although muscle activities are the basic feature of running biomechanics, they remain largely unknown to runners with different running experiences. In our present study, the significant differences between different running groups in the muscles and knee joints will be discussed around the hypothesis and other results.

The similarity between the predicted knee muscle group activation patterns and EMG data provides confidence in our model's ability to estimate muscle forces and loading patterns at the knee which is also consistent with previous studies [73,135]. Significant differences in muscle forces were noticed between runners, a pattern was observed that the BFLH force of novice runners was greater than that of experienced runners during the push-off phase. BFLH is one of the biarticular muscles that crosses the knee and hip joints, which are important contributors to propulsion [156]. In contrast, the uniarticular knee extensors performed negative work as brakes. Studies [141,157] have shown that healthy subjects exhibit co-contraction of the quadriceps and hamstrings. In a comparative analysis between novice and experienced runners, it was observed that novice runners demonstrated increased quadriceps muscle activity in RF, VL, and VM during the initial contact phase. This elevated activity in RF, VL, and VM provides partial evidence for the quadriceps avoidance strategy, which is postulated to counteract knee instability. When passive ligamentous restraint is absent, heightened quadriceps and hamstring activity could signify a co-contraction between these two muscle groups. This co-contraction aims to stabilize the knee, as supported by findings from previous studies [158-160]. The elevated quadriceps muscle activity in novice runners suggests that they might need enhanced quadriceps strength to sustain knee joint stability during running compared experienced runners. Regarding the forces produced by ankle plantar flexors, GM, GL, and SO are known to generate force during running to support body weight and support in forward propulsion [161]. Notably, novice runners showed higher GM, GL, and SO forces during the push-off phase. This could imply that they demand more effort for propulsion than experienced runners. Conversely, the TA force was higher in experienced runners during the initial contact phase, but this trend diminished during the mid-stance and push-off phases. Given that TA is a primary contributor to both dorsiflexion and plantarflexion, and it plays a crucial role in facilitating the appropriate sagittal plane movement of the ankle joint, it is expected that TA would exhibit an increase in force output [98]. These findings provide additional insight into the mechanical differences between runners during running and may also provide runners with important clinical information regarding their susceptibility to injuries during running.

Knee reaction force plays a significant role in joint stability. Karamanidis et al. [162] advocate that compared to non-active subjects, the knee joint gearing of experienced runners is more advantageous. In our study, the knee joint reaction force between runners showed significant differences in all directions. The novice group showed higher medial knee reaction force during the mid-stance phase, the increased medial force has a good correlation with osteoarthritis since osteoarthritis often begins in the medial compartment of the knee [163]. Previous research has demonstrated that runners do not have a higher risk of osteoarthritis than non-runners [164]. The anterior-posterior force and superior-inferior force showed a similar trend of the difference caused by running experiences, which were observed higher in novice runners during initial contact, then showed a lower trend during the mid-stance phase. Knee joint instability is most prominent in the anterior-posterior plane in patients with anterior cruciate ligament rupture. Aghdam et al. [160] found significantly greater anterior shear force in anterior cruciate ligament rupture patients compared to the healthy group. The tibiofemoral compressive force was the main component during the walking stance phase, whereas the anterior-posterior shear component peaked at roughly 70% of the compressive peak during the running stance phase [165]. The peak force is frequently used as the primary outcome variable in studies on joint loading, however, the shape of biomechanical time series data can provide more comprehensive insights. Experienced runners showed that higher knee forces during mid to late stance may contribute more efficiently to propulsion. In this study, novice runners did not always have higher knee forces than experienced runners, which may suggest a difference in the biomechanics of running between the two groups. However, it cannot be directly inferred that novice runners are more susceptible to knee injuries than experienced runners.

In addition, studies [141,166] found that among healthy adult, muscles play an important role in determining knee joint loading. However, the correlation between selected muscles and knee reaction forces was not very strong in this study. Novice runners showed different correlations between muscle forces and knee reaction forces compared to experienced runners, which indicates that runners may adopt a biomechanical profile that gives uncertain patterns. Previous studies concluded that novice runners have a higher incidence of lower extremity injuries than experienced runners in both short-distance and long-distance groups. An explanation for why novice runners sustain more injuries is that their poor running mechanics place higher loads on their musculoskeletal system, particularly around the knee and at the tibia [167,168]. Years of expertise may potentially lower the risk of injury through improved musculoskeletal tissue tolerance to repetitive loads or better training methods that allow for adequate recovery periods [95]. However, in this study, there was no direct evidence that the novice group were more likely to be injured than the experienced group.

This study has limitations that must be considered. The present study estimated muscle forces and knee reaction forces based on modeling simulations, which should be regarded with caution. However, we compared experimentally normalized EMG signals to predicted muscle activity and confirmed that the model's muscles are active during physiological periods. Meanwhile, several smaller muscles that cross the knee, such as the tensor fascia latae, gracilis, and orsartorius, were omitted from the present model. These muscles have a smaller physiological cross-sectional area than the quadriceps, hamstrings, and gastrocnemius, hence they contribute less to the knee joint [169]. The impact of personality on the properties of muscle tissue should not be disregarded, despite the fact that the OpenSim model's muscular properties were not modified based on the subject population, which should be considered in future projects [170]. In recent years, muscle-driven simulations of joint reaction forces utilizing inverse kinematics have vastly developed and are thus considered a valuable tool for clinical analysis [98,138,171]. Even though musculoskeletal simulation techniques remain relatively new, it is possible to make significant advances in clinical biomechanics research by enhancing their accuracy and pursuing further advancements through more research. Another limitation of our study is the relatively small sample size, consisting only of male runners. In future research, we will recruit more runners to expand the sample size, including both men and women.

4.2 Joint biomechanics of a 5 km run and running experience

The purpose of this study was to analyze the biomechanical effects of a 5 km run between novice and experienced runners. Differences in lower limb kinematics and kinetics during a prolonged running session between novice runners and experienced runners were found. For the discrete variables obtained by a two-way repeated-measures ANOVA, the joint ROM showed differences between novice runners and experienced runners. For the PC modeling of waveforms, it was observed that the first four PCs accounted for the most variations, ranging from 86.52% to 96.16% for all biomechanical variables investigated, which is consistent with the literature [146,149,154]. PCs were a set of orthogonal waveform features obtained after principal component analysis of mixed biomechanical waveforms from multiple subjects. Typically, four PCs can be used to explain the main variation in a dataset. Using a PCA approach may offer unique insights into the underlying patterns of running biomechanical waveforms. These findings partially supported the hypothesis.

The runner's experience was expected to influence running performance and injury risks by altering lower extremity kinematics and kinetics. Consistent with this assumption, novice runners showed greater ankle Invert/Evert ROM, which was believed to be associated with RRIs [172]. Meanwhile, experienced runners had greater knee Ext/Flex ROM; this finding is in agreement with previous research [173]. The greater knee flexion ROM appears to be a protective adaptation in experienced runners, as previous studies have suggested that a greater knee flexion angle during stance can reduce the ground reaction force and attenuate shock impacts above the knee joint [174]. Our results showed that novice runners had greater hip Adduct/Abduct ROM; increased hip adduction has been identified as a potential risk factor for common running injuries such as iliotibial band syndrome [95]. After a 5 km run, the ROM of the knee Adduct/Abduct and hip adduct/abduct increased. The accumulated fatigue of hip abductor muscle-tendon units (tensor fasciae latae, gluteus medius, and gluteus minimus) may be causing the increase in hip adduction, and hip musculature is essential in overcoming substantial external hip adduction moments [175]. The stability of the hip joint may prevent RRIs to a certain extent. Similarly, Willwacher et al. [176] found clear changes in Adduct/Abduct and Int Rot/Ext Rot joint kinematics after a 10 km long-distance run.

Even though no statistical difference exists in standard discrete value analysis of the lower limb, PCA was capable of recognizing significant differences in the waveforms of joint angles between novice and experienced runners with the prolonged running session. PC1 captured the general magnitude differences in the data, PC2 primarily captured the differences in timing, and PC3 extracted differences in relative amplitudes [177]. PC1 and PC2 of the ankle Invert/Evert angle captured the significantly greater eversion angle in novice runners with respect to experienced runners, similar to the findings reported by Maas et al. [173]. The high eversion angle of the ankle joint has been linked to a higher risk of injury development in runners. It has been hypothesized that increased ankle eversion can lead to greater medial foot displacement, which is associated with increased tibial abduction [178,179]. Novice runners should be mindful of changes occurring in the ankle joint during running, particularly the eversion angle, and make necessary adjustments promptly. PC2 captured subtle shifts in the timing of the peak knee flexion angle, while PC3 reflected an increase in the knee flexion angle during early stance, specifically among novice runners. In hip Flex/Ext, PC2 and PC3 revealed that experienced runners exhibited a greater hip flexion angle than novice runners, which is consistent with a previous study [90]. The increased knee internal rotation angle, increased hip adduction angle, and increased hip internal rotation angle have been associated with RRIs, especially iliotibial band syndrome, which were reflected in PC1 of the knee Int Rot/Ext Rot, hip Adduct/Abduct, and hip Int Rot/Ext Rot among novice runners [180-182]. Compensatory femoral internal rotation caused by excessive tibial internal rotation during stance may lead to knee stress injuries [179,183]. These kinematic changes in novice runners may indicate a lack of control over running technique, while experienced runners may exhibit greater control.

There were also significant differences between the experience levels of runners for the kinetic variables. The peak ankle inversion moment and peak internal rotation moment of novice runners were greater than those of experienced runners. An increase in ankle inversion moment indicates that novice runners may have increased demands on the ankle varus muscles, including the anterior tibialis and posterior tibialis, which play a role in eccentrically supporting the plantar arch during the stance phase [150]. At the hip joint, the extension moment of novice runners was reduced, and the abduction moment was increased. The lack of running experience may be related to an imbalance of hip muscles. The decreases in ankle plantar-flexion moment and knee extension moment were noticed during post-5 km running. These changes in biomechanics after a prolonged running session were consistent with previous research [180].

PC1 and PC2 of the ankle Invert/Evert moment captured the differences in magnitude and amplitude between the two groups, reporting a significantly greater inversion moment in novice runners compared to experienced runners throughout the entire stance phase. PC1 and PC3 of the ankle Int Rot/Ext Rot moment captured the differences between experienced runners and novice runners, showing that novice runners have a greater ankle internal rotation moment than experienced runners. The greater moment can reflect an increase in antagonistic activity and, thus, may indicate increased joint load. PC2 and PC3 extracted phase shift and amplitude differences in the knee Ext/Flex moment, while PC2 extracted phase shift differences in the knee Adduct/Abduct moment. This time delay would decrease the loading rate during the initial stance to midstance, which has been considered a risk factor for overuse running injuries. Given the relatively modest variance explained, it was difficult to distinguish the influence expressed by PC4. The increased knee internal rotation moment throughout the entire stance phase may be an unintended effect of running, as it has been linked to the progression of knee osteoarthritis during gait [184]. In the hip joint, PC1 captured the magnitude difference in the flexion moment, which was consistent with the hip Flex/Ext ROM.

Kinetic differences between pre-5 km running and post-5 km running were also reflected in PC scores of joint moments, especially in the ankle and knee joints. The reduced plantarflexion moment may be due to the decrease in energy absorption caused by sustained running, and the decreased knee extension moment during the middle stance and later stance may indicate that runners have weak extensor muscles after a 5 km run.

While the traditional two-way repeated-measures ANOVA can only perform statistical analysis on discrete values, PCA can perform dimensionality reduction analysis on the entire time series curve. PCA captured differences in the magnitude and amplitude of lower extremity biomechanical waveforms by retaining at least 90% of the available information [185]. Using single-component reconstruction, the lower limb joint angles, joint moments, and GRFs collected by PCA can be interpreted visually. In fact, this method may offer a robust and clinically relevant interpretation. In the current study, PCs generated from lower extremity kinematics and kinetics were shown to be indicators of running experience effects and prolonged running effects. Results from our study suggest that running experience may influence the running mechanics of runners, especially those commonly associated with RRIs. Biomechanical changes during post-5 km running might be associated with a fatigued state and may help to understand potential alterations due to overuse injuries [127,186].

Several limitations in this study should be acknowledged. Firstly, the running biomechanics differences in this study may have been affected by running speed, as we collected gait data at the preferred running speed of runners rather than a uniform speed to ensure a more natural gait pattern. Abbasi et al. [187] suggested that gait coupling patterns changed as running speed varied. Orendurff et al. [109] found that running speed affects lower limb joint biomechanics, especially in maximal kinematic and kinetic variables of the hip, knee, and ankle joints. However, a few studies [115] indicated that running speed does not have a significant influence on the lower limb biomechanical asymmetry of runners. In order to gain a better understanding of this aspect, future research will focus on determining the influence of various running speeds on lower limb biomechanics. Secondly, we investigated how a prolonged running session influences gait data, thus using a 5 km run protocol rather than a fatigue run protocol. Different runners have different reactions to the 5 km run; most novice runners have reached an exerted fatigue state after a 5 km run, while experienced runners have not. Thirdly, due to limitations in laboratory and experimental equipment, we conducted our data collection overground, whereas the 5 km running was performed on a treadmill. It is important to note that running on different surfaces can potentially introduce biomechanical differences to some extent, which should be avoided in future research to ensure more accurate and consistent findings [188]. Furthermore, I investigated only male runners; as gender differences exist in running biomechanics, our findings may not apply to female runners. Future studies could perform this kind of analysis on female runners. These limitations should be considered in future studies.

4.3 Ground reaction force of running speed and runners' gender

The primary purpose of this study was to explore gait pattern differences in GRFs between male and female runners across seven running speeds. We hypothesized that specific GRF characteristics would vary between genders across different speeds and that GRFs would correlate with running speed. The primary finding of our study was that female runners exhibit higher propulsive and vertical forces than male runners at all tested speeds. The findings indicated that distinct running patterns for male and female runners are identifiable through GRFs at each speed, particularly during the later stance phase. We also observed that running speed significantly influences GRFs for all runners, with all genders exhibiting increased trends in early and mid-late stance as speed increased. The stepwise regression analysis revealed that certain discrete GRF variables could predict running speed, thereby providing partial support for our hypothesis.

Distinction in the incidence rates of specific injuries among male and female runners has indicated the necessity to distinguish running mechanics [5,189,190]. The differences in propulsive force between males and females at each running speed suggest that female runners may require more effort to accelerate the body to maintain forward momentum, in order to keep the same speed as male runners. Previous studies also found that propulsive force in females was higher when compared with males at the same speed during running [102,191]. Females have a larger braking force during the first half of stance at running speeds of 10 km/h and 11 km/h, which indicated that the mass center of females accelerated more backward than males at slower speeds. Faster running speeds require a higher amount of propulsive force, but not necessarily a lower amount of braking force [192]. Previous studies have analyzed gender differences on parameters extracted from vertical GRF [102,104], but none have prospectively conducted time series curve analysis on GRF throughout the stance phase. A novel finding in this study was that females have higher vertical GRF during the later stance phase at each running speed compared with males. Higher vertical GRF may be considered an inevitable result of needing a higher percentage of available strength to propel the body towards toe-off [193]. Female runners exhibited a greater peak vertical impact force at faster running speeds, which may induce potential shock increases in the musculoskeletal system and thus lead to RRIs [194,195]. This may provide a potential explanation for the higher patellofemoral pain and tibial stress fracture rates among female runners [101,196].

In 2016, Yokoyama et al. [197] identified three running speed categories: slow (2.7–2.9 m/s), moderate (3.5–3.7 m/s), and fast (4.4–4.5 m/s) for experienced runners. In this study, we opted for a speed range of 10–16 km/h, corresponding to 2.78–4.44 m/s. This selection spans the spectrum from slow to fast running, facilitating a more comprehensive examination of the effect of running speed on gait mechanisms. Furthermore, the incremental difference of 1 km/h (0.28 m/s) between each chosen running speed allows for a more detailed investigation of the impact of speed on GRFs. As running speed increases, male and female runners exhibit different GRF characteristics. The results of our study demonstrated that running speed had a significant effect on propulsive force during the second half of stance in both females and males. Runners typically exhibit a forward inclination of the trunk, with foot contact striking the ground behind the body's center of mass. Consequently, from a biomechanical perspective, the aim is to maximize the propulsive component of GRF to maintain faster running speeds [108]. Additionally, male runners exhibited increased braking force at higher speeds, suggesting greater impact during the braking phase of high-speed running, aligning with previous findings [198]. The runner-ground interaction during the braking phase is crucial, playing a significant role in lower extremity injury risk [33,194,199]. At initial ground contact, the lower extremity experiences rapid loading with forces exceeding 1.5 times the runner's body weight [43,112]. With increased speed, runners displayed an increased peak vertical impact force, producing greater external loads on their bodies. The forefoot underwent considerable loading. Previous studies investigated running speed as it is related to GRFs and found similar correlations [22,200]. Interestingly, we observed no significant differences in peak vertical active force across speeds. The relationship between peak vertical impact force and lower limb injuries, however, remains a topic of controversy [201,202].

The current results also showed significant correlations (i.e., Pearson correlation and stepwise multiple linear regression) between running speed and the GRF variables, which are compatible with the findings by Breine et al. [200] and Fukuchi et al. [203]. The GRF variables chosen in this study together explained approximately 90% of the variance associated with increases in running speed. Key contributors and predictors of higher running speeds for all genders included peak propulsive force, vertical impulse, peak vertical impact force and VALR. Schache et al. [108] suggested that, to achieve higher running speeds, runners tend to exert greater force against the ground rather than increasing the frequency of their strides. This conclusion is also consistent with the higher values we recorded for the propulsive force. Consistent with our hypothesis, variations in vertical GRF were responsive to changes in running speed, indicating the necessity for the legs to generate more vertical force to attain faster speeds. Notably, braking impulse and peak braking force emerged as significant factors only in male runners. This could be explained by the fact that male runners, having relatively larger body weights, experience greater gravity and inertia effects during the braking phase, which emphasizes the importance of the braking phase in their running mechanics [204].

This study verified whether running speeds influence the GRF on overground running and whether these likely influences depended on gender differences. However, several limitations must be acknowledged. The selected running speeds are based on absolute values, not relative to each runner's physiological capabilities. We chose absolute speeds to quantify the impact of speed more accurately on a runner's GRF and minimize potential biomechanical differences that could arise from differences in relative speeds. Furthermore, the intervals between the selected running speeds are relatively small. Utilizing speeds based on each runner's physiological capabilities could have introduced confounding variables into our experimental results. Nevertheless, it is important to acknowledge that this constitutes a limitation of the present study. Notably, even at identical speeds, runners may experience differing physiological intensities [205]. Another limitation is that the order of running speeds was not randomized. This decision was primarily made to prevent fatigue effects by minimizing the total duration of time spent in the laboratory, which was kept under two hours [109]. Randomizing the running speeds could be challenging and potentially unsafe, especially when attempting to achieve high running speeds without first gradually progressing through lower speeds [136]. It is also important to consider that the observed changes at higher speeds might result from both the external force exerted during ground contact and muscle force production in anticipation of or in response to surface interaction [19,43,108]. The knee joint is the most susceptible to injury during running [195]. While the current study only focuses on GRF, future research will integrate the biomechanics of the knee joint with GRF to investigate the impact of running speed on runners. Moreover, our findings are based on data from healthy runners and may not reflect GRF pattern changes in runners with RRI. Future research should include runners with RRI to investigate gender and speed influences on RRI risk factors within this population.

5 CONCLUSIONS AND FUTURE WORKS

5.1 Muscle force and knee reaction force between novice and experienced runners

This study explored the differences in muscle forces and knee joint reaction forces between novice and experienced runners using biomechanical modeling and simulation. Specifically, it examined how knee joint loading and muscle forces interact during running. The findings revealed distinct running mechanics between the two groups, with novice runners exhibiting higher knee joint loading and muscle forces throughout much of the stance phase compared to experienced runners. These elevated forces in novice runners may be linked to a greater risk of RRIs, as higher knee joint loading is commonly associated with increased injury susceptibility. The study highlights the importance of understanding the mechanical differences between novice and experienced runners, especially regarding knee joint loading. Novice runners appear to generate more stress on their lower extremities, which may place them at a higher risk for RRIs. However, the degree to which these altered mechanics impact performance or increase injury risk remains unclear. Furthermore, the threshold levels of muscle forces and joint reaction forces that predict injury risk have not yet been established. Future research should focus on longitudinal studies that investigate whether specific knee joint variables, such as loading patterns or muscle forces, can be used as reliable predictors of RRI risk. Understanding these variables could help in developing injury prevention strategies tailored to the biomechanical profiles of novice runners, potentially improving their performance while minimizing injury risks.

5.2 Joint biomechanics of a 5 km run and running experience

The present study aimed to compare the running mechanics of novice and experienced runners during a 5 km run by employing both traditional discrete variables and PCA with single-component reconstruction for waveform analysis. The results demonstrated that PCA offers valuable insights beyond those provided by conventional methods, highlighting distinct biomechanical patterns that may be related to injury mechanisms. Specifically, running experience was shown to significantly influence lower limb biomechanics. Novice runners displayed more pronounced variations in joint angles and moments compared to their experienced counterparts. These differences may contribute to a higher risk of lower limb injuries in novice runners, as the greater biomechanical variability could lead to increased joint loading and stress. Understanding these patterns is essential for the development of targeted training programs and injury prevention protocols that cater to runners with varying levels of experience. Moreover, this study underscores the importance of employing advanced biomechanical analysis techniques, such as PCA, to capture subtle but meaningful differences in running mechanics. Future research could take a longitudinal approach, prospectively examining runners across different experience levels to determine whether specific biomechanical variables can be predictive of performance enhancements and injury risks. By identifying these variables, researchers and coaches could develop more personalized training regimens that not only improve running performance but also mitigate the likelihood of injury, particularly for novice runners as they progress toward more advanced training stages.

5.3 Ground reaction force of running speed and runners' gender

This study investigated the effects of increased running speeds (10–16 km/h) on GRFs in both male and female runners on an overground runway. The findings revealed notable adaptations in GRFs with speed increases, which were observed consistently across genders. Despite the biomechanical differences between male and female runners, both groups displayed a similar trend in GRF adaptations. The most significant changes occurred during the early and late stance phases, with vertical and anterior-posterior GRFs increasing proportionally as running speed increased. Interestingly, the study also highlighted that female runner exhibited higher propulsive and vertical forces compared to their male counterparts during the late stance phase at all running speeds. This suggests that females may exert more effort to maintain the same running speed as males, potentially due to differences in muscle strength or running mechanics. These findings shed light on the gender-specific biomechanical factors influencing running performance, particularly in relation to GRFs, which are critical for understanding the loading patterns experienced by runners. The results contribute valuable insights into the movement patterns associated with GRFs during running, which could help inform injury prevention and performance optimization strategies. Future research should further explore the relationship between RRIs and gender-specific GRF patterns. This will deepen our understanding of how gender differences in biomechanics might influence injury risk and provide a foundation for developing tailored interventions to mitigate injury risks for male and female runners alike.

NEW SCIENTIFIC THESIS POINTS

1st Thesis point

It is possible that increased experience leads to improved running mechanics and fewer injuries, however, the reason behind this is still unclear. Indicators of RRIs or running biomechanics related to higher knee joint loading are frequently investigated during running research. Based on my experiments, the lower limb muscle forces and knee joint loading of runners were estimated by musculoskeletal modeling based on OpenSim (Figure 32). The results showed that novice runners and experienced runners have different running mechanisms, mainly novice runners showed significantly bigger knee loading and muscle forces than the experienced group in most of the stance phases. Considering the proposed relationship between knee joint loading, muscles, and RRIs, the novice group may be more prone to lower extremity injuries due to increased loading during running compared to experienced runners. However, the evidence is not direct that novice runners are at greater risk for RRI.

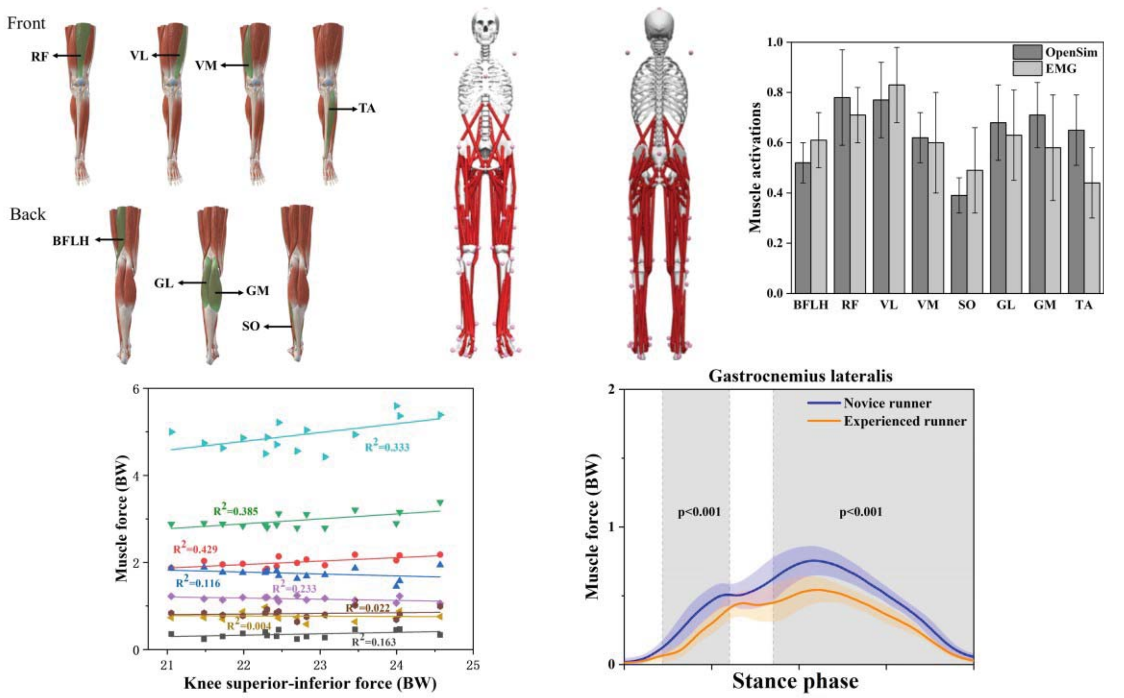

Figure 1 Muscle force differences between the novice and experienced runners

Related articles to the first thesis point:

1. Kang, Z., **Jiang, X.** (2024). The effect of running experience on muscle forces and knee joint reaction forces during running. International Journal of Biomedical Engineering and Technology. **IF: 0.7, Q4**

2nd Thesis point

Studying the biomechanics of runners with different running experiences before and after long-distance running can improve our understanding of the relationship between faulty running mechanics and injury. Based on my experiments (biomechanical data were collected from 15 novice and 15 experienced runners), which I used both PCA with single-component reconstruction and a two-way repeated-measures ANOVA was conducted to explore the effects of runner and a 5 km run (Figure 33). I found that novice runners exhibited greater changes in joint angles and joint moments than experienced runners regardless of the prolonged running session, and those patterns may relate to lower limb injuries. I also found that the reduced ankle plantarflexion moment may be due to the decrease in energy absorption caused by sustained running, and the decreased knee extension moment during the middle stance and later stance may indicate that runners have weak extensor muscles after a 5 km run. The results of this study suggest that the PCA approach can provide unique insight into running biomechanics and injury mechanisms. The findings from the study could potentially guide training program developments and injury prevention protocols for runners with different running experiences.

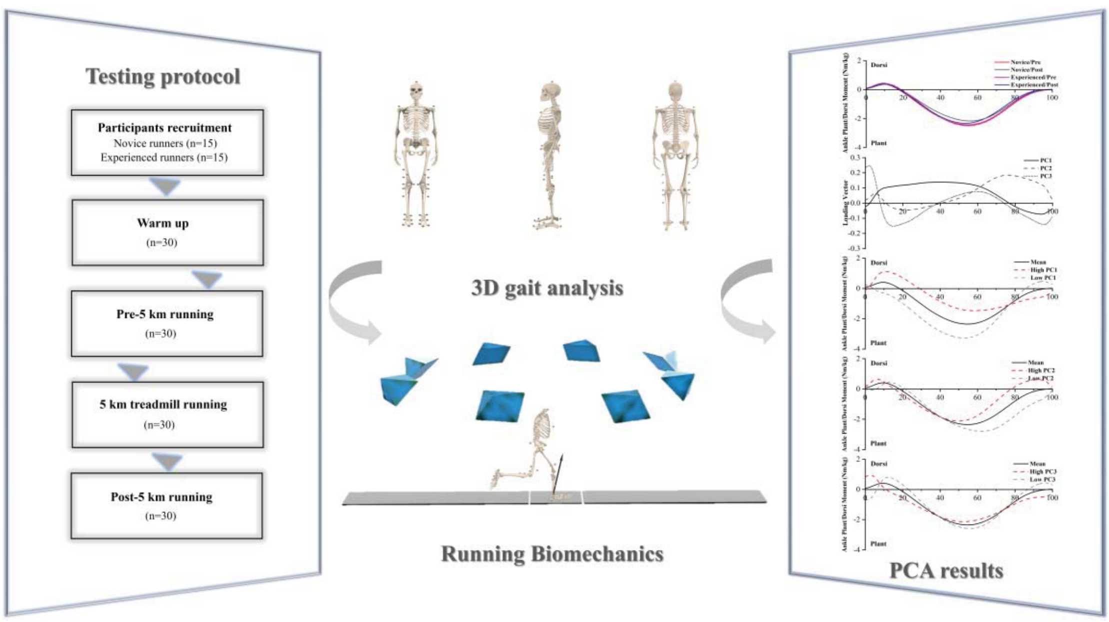

Figure 33 The overview of running biomechanical study on runners

Related articles to the second thesis point:

1. **Jiang, X.**, Xu, D., Fang, Y., Bíró, I., Baker, J. S., Gu, Y. (2023). PCA of Running Biomechanics after 5 km between Novice and Experienced Runners. Bioengineering, 10(7), 876. **IF: 3.8, Q3**

3rd Thesis point

Based on my experiments, I investigated the gait pattern differences between males and females while running at different speeds and verified the relationship between GRFs and running speed among both males and females (Figure 34). GRF data were collected from forty-eight participants (thirty male runners and eighteen female runners) while running on an overground runway at seven discrete speeds: 10, 11, 12, 13, 14, 15 and 16 km/h. The ANOVA results showed that running speed had a significant effect (p < 0.05) on GRFs, propulsive and vertical forces increased with increasing speed. An independent t-test also showed significant differences (p < 0.05) in vertical and anterior-posterior GRFs at all running speeds, specifically, female runners demonstrated higher propulsive and vertical forces than males during the late stance phase of running. Pearson correlation and stepwise multiple linear regression showed significant correlations between running speed and the GRF variables. These findings suggest that female runners require more effort to keep the same speed as male runners. This study may provide valuable insights into the underlying biomechanical factors of the movement patterns at GRFs during running.

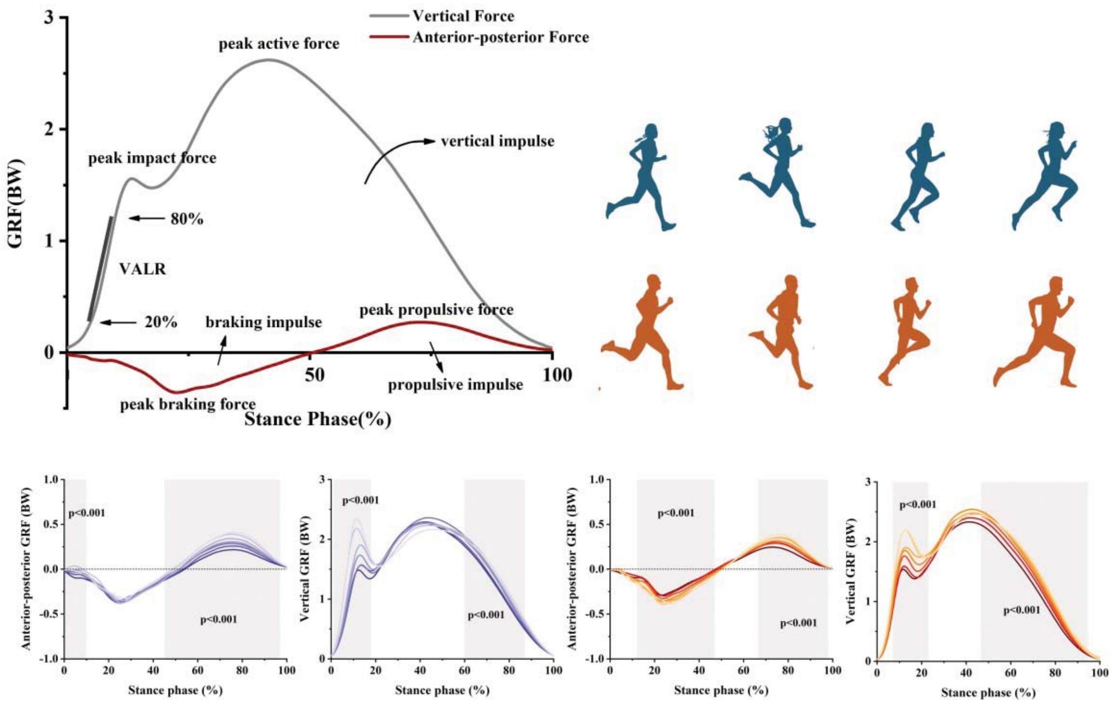

Figure 34 GRF changes between female and male runners across 7 running speeds

Related articles to the third thesis point:

1. **Jiang, X.**, Bíró, I., Sárosi, J., Fang, Y., Gu, Y. (2024). Comparison of ground reaction forces as running speed increases between male and female runners. Frontiers in Bioengineering and Biotechnology, 12, 1378284. **IF: 4.3, Q1**

LIST OF PUBLICATIONS

Articles related to this thesis

1. **Jiang, X.**, Xu, D., Fang, Y., Bíró, I., Baker, J. S., & Gu, Y. (2023). PCA of Running Biomechanics after 5 km between Novice and Experienced Runners. Bioengineering, 10(7), 876. **IF: 3.8, Q3**
2. **Jiang, X.**, Bíró, I., Sárosi, J., Fang, Y., Gu, Y. (2024). Comparison of ground reaction forces as running speed increases between male and female runners. Frontiers in Bioengineering and Biotechnology, 12, 1378284. **IF: 4.3, Q1**
3. Kang, Z., **Jiang, X.** (2024). The effect of running experience on muscle forces and knee joint reaction forces during running. International Journal of Biomedical Engineering and Technology. **IF: 0.7, Q4**

International conference related to this thesis

1. **Jiang, X.**, & Bíró, I. (2023, April). Muscle force estimation in running gait analysis after a prolonged running session via OpenSim. In 2023 4th International Conference on Computer Engineering and Application (ICCEA) (pp. 498-502). IEEE.
2. **Jiang, X.**, & Bíró, I. (2023, May). The effects of shoes with a triple density midsole on lower limb kinematics and kinetics in male recreational runners. In 2023 IEEE 17th International Symposium on Applied Computational Intelligence and Informatics (SACI) (pp. 000661-000666). IEEE.
3. **Jiang, X.**, Zhang, Q., & Bíró, I. (2024, May). Effects of Long-Distance Running on Lower Limb Joint Kinematics in Recreational Runners. In 2024 IEEE 18th International Symposium on Applied Computational Intelligence and Informatics (SACI) (pp. 000095-000098). IEEE.

Other publications

1. **Jiang, X.**, Sárosi, J., Bíró, I. (2024). Characteristics of lower limb running-related injuries in trail runners: a systematic review. Physical Activity and Health, 8(1). CS

(Scopus CiteScore): 4.4, Q1

2. **Jiang, X.**, Zhou, H., Quan, W., Hu, Q., Baker, J. S., & Gu, Y. (2021). Ground reaction force differences between bionic shoes and neutral running shoes in recreational male runners before and after a 5 km run. International Journal of Environmental Research and Public Health, 18(18), 9787. **CS: 7.3, Q1**
3. **Jiang, X.**, Yang, X., Zhou, H., Baker, J. S., & Gu, Y. (2021). Prolonged running using bionic footwear influences lower limb biomechanics. Healthcare, 9 (2), 236. **IF: 2.4, Q2**
4. **Jiang, X. Y.**, Li, S. D., Teo, E. C., & Zhou, Z. X. (2021). The effect of high-intensity intermittent training on the acute gait plantar pressure in healthy young adults. Journal of Biomimetics, Biomaterials and Biomedical Engineering, 49, 21-32. **IF: 0.5, Q4**
5. **Jiang, X.**, Chen, H., Sun, D., Baker, J. S., & Gu, Y. (2021). Running speed does not influence the asymmetry of kinematic variables of the lower limb joints in novice runners. Acta of Bioengineering and Biomechanics, 23(1). **IF: 0.8, Q3**
6. Xu, D., Zhou, H., Quan, W., **Jiang, X.**, Liang, M., Li, S., ... & Gu, Y. (2024). A new method proposed for realizing human gait pattern recognition: Inspirations for the application of sports and clinical gait analysis. Gait & Posture, 107, 293-305. **IF:2.2, Q1**
7. Xu, D., Zhou, H., **Jiang, X.**, Li, S., Zhang, Q., Baker, J. S., & Gu, Y. (2022). New insights for the design of bionic robots: adaptive motion adjustment strategies during feline landings. Frontiers in Veterinary Science, 9, 836043. **IF:2.6, Q1**
8. Xu, D., **Jiang, X.**, Cen, X., Baker, J. S., & Gu, Y. (2020). Single-leg landings following a volleyball spike may increase the risk of anterior cruciate ligament injury more than landing on both-legs. Applied Sciences, 11(1), 130. **IF:2.5, Q1**

REFERENCE

1. Mandolesi, L.; Pol1. Mandolesi, L.; Polverino, A.; Montuori, S.; Foti, F.; Ferraioli, G.; Sorrentino, P.; Sorrentino, G. Effects of physical exercise on cognitive functioning and wellbeing: biological and psychological benefits. Frontiers in psychology 2018, 9, 509.

2. Lee, D.-c.; Brellenthin, A.G.; Thompson, P.D.; Sui, X.; Lee, I.-M.; Lavie, C.J. Running as a key lifestyle medicine for longevity. Progress in cardiovascular diseases 2017, 60, 45-55.

3. Oswald, F.; Campbell, J.; Williamson, C.; Richards, J.; Kelly, P. A scoping review of the relationship between running and mental health. International journal of environmental research and public health 2020, 17, 8059.

4. Messier, S.P.; Martin, D.F.; Mihalko, S.L.; Ip, E.; DeVita, P.; Cannon, D.W.; Love, M.; Beringer, D.; Saldana, S.; Fellin, R.E. A 2-year prospective cohort study of overuse running injuries: the runners and injury longitudinal study (TRAILS). The American journal of sports medicine 2018, 46, 2211-2221.

5. Van Gent, R.; Siem, D.; van Middelkoop, M.; Van Os, A.; Bierma-Zeinstra, S.; Koes, B. Incidence and determinants of lower extremity running injuries in long distance runners: a systematic review. British journal of sports medicine 2007, 41, 469-480.

6. Lieberman, D.E.; Venkadesan, M.; Werbel, W.A.; Daoud, A.I.; D'andrea, S.; Davis, I.S.; Mang'Eni, R.O.; Pitsiladis, Y. Foot strike patterns and collision forces in habitually barefoot versus shod runners. Nature 2010, 463, 531-535.

7. Lun, V.; Meeuwisse, W.; Stergiou, P.; Stefanyshyn, D. Relation between running injury and static lower limb alignment in recreational runners. British journal of sports medicine 2004, 38, 576-580.

8. Buist, I.; Bredeweg, S.W.; Bessem, B.; Van Mechelen, W.; Lemmink, K.A.; Diercks, R.L. Incidence and risk factors of running-related injuries during

preparation for a 4-mile recreational running event. British journal of sports medicine 2010, 44, 598-604.

9. Buist, I.; Bredeweg, S.W.; Lemmink, K.A.; Van Mechelen, W.; Diercks, R.L. Predictors of running-related injuries in novice runners enrolled in a systematic training program: a prospective cohort study. The American journal of sports medicine 2010, 38, 273-280.

10. Jiang, X.; Chen, H.; Sun, D.; Baker, J.S.; Gu, Y. Running speed does not influence the asymmetry of kinematic variables of the lower limb joints in novice runners. Acta of Bioengineering and Biomechanics 2021, 23.

11. Blair, S.N.; Kohl, H.W.; Goodyear, N.N. Rates and risks for running and exercise injuries: studies in three populations. Research Quarterly for Exercise and Sport 1987, 58, 221-228.

12. Xiang, L.; Mei, Q.; Wang, A.; Shim, V.; Fernandez, J.; Gu, Y. Evaluating function in the hallux valgus foot following a 12-week minimalist footwear intervention: A pilot computational analysis. Journal of Biomechanics 2022, 132, 110941.

13. Gu, Y.; Lu, Y.; Mei, Q.; Li, J.; Ren, J. Effects of different unstable sole construction on kinematics and muscle activity of lower limb. Human movement science 2014, 36, 46-57.

14. Hannigan, J.; Pollard, C.D. Differences in running biomechanics between a maximal, traditional, and minimal running shoe. Journal of science and medicine in sport 2020, 23, 15-19.

15. Buist, I.; Bredeweg, S.W.; Van Mechelen, W.; Lemmink, K.A.; Pepping, G.-J.; Diercks, R.L. No effect of a graded training program on the number of running-related injuries in novice runners: a randomized controlled trial. The American journal of sports medicine 2008, 36, 33-39.

16. Ferber, R.; Davis, I.M.; Williams Iii, D.S. Gender differences in lower extremity mechanics during running. Clinical biomechanics 2003, 18, 350-357.

17. Chumanov, E.S.; Wall-Scheffler, C.; Heiderscheit, B.C. Gender differences in walking and running on level and inclined surfaces. Clinical biomechanics 2008, 23, 1260-1268.

18. Xie, P.-P.; István, B.; Liang, M. Sex-specific differences in biomechanics among runners: A systematic review with meta-analysis. Frontiers in Physiology 2022, 13, 994076.

19. Hamner, S.R.; Delp, S.L. Muscle contributions to fore-aft and vertical body mass center accelerations over a range of running speeds. Journal of biomechanics 2013, 46, 780-787.

20. Stoquart, G.; Detrembleur, C.; Lejeune, T. Effect of speed on kinematic, kinetic, electromyographic and energetic reference values during treadmill walking. Neurophysiologie Clinique/Clinical Neurophysiology 2008, 38, 105-116.

21. Nilsson, J.; Thorstensson, A. Ground reaction forces at different speeds of human walking and running. Acta Physiologica Scandinavica 1989, 136, 217-227.

22. Hollis, C.R.; Koldenhoven, R.M.; Resch, J.E.; Hertel, J. Running biomechanics as measured by wearable sensors: effects of speed and surface. Sports biomechanics 2021.

23. Jiang, X.; Bíró, I.; Sárosi, J.; Fang, Y.; Gu, Y. Comparison of ground reaction forces as running speed increases between male and female runners. Frontiers in Bioengineering and Biotechnology 2024, 12, 1378284.

24. Xu, D.; Zhou, H.; Quan, W.; Jiang, X.; Liang, M.; Li, S.; Ugbolue, U.C.; Baker, J.S.; Gusztav, F.; Ma, X. A new method proposed for realizing human gait pattern recognition: Inspirations for the application of sports and clinical gait analysis. Gait & Posture 2024, 107, 293-305.

25. Sallis, R.; Franklin, B.; Joy, L.; Ross, R.; Sabgir, D.; Stone, J. Strategies for promoting physical activity in clinical practice. Progress in cardiovascular diseases 2015, 57, 375-386.

26. Shiroma, E.J.; Sesso, H.D.; Moorthy, M.; Buring, J.E.; Lee, I.M. Do moderate - intensity and vigorous - intensity physical activities reduce mortality rates to the same extent? Journal of the American Heart Association 2014, 3, e000802.

27. Mei, Q.; Gu, Y.; Xiang, L.; Yu, P.; Gao, Z.; Shim, V.; Fernandez, J. Foot shape and plantar pressure relationships in shod and barefoot populations. Biomechanics and modeling in mechanobiology 2020, 19, 1211-1224.

28. Dugan, S.A.; Bhat, K.P. Biomechanics and analysis of running gait. Physical Medicine and Rehabilitation Clinics 2005, 16, 603-621.

29. Novacheck, T.F. The biomechanics of running. Gait & posture 1998, 7, 77-95.

30. Ounpuu, S. The biomechanics of running: a kinematic and kinetic analysis. Instructional course lectures 1990, 39, 305-318.

31. Puleo, J.; Milroy, P. Running anatomy; Human Kinetics: 2018.

32. Ounpuu, S. The biomechanics of walking and running. Clinics in sports medicine 1994, 13, 843-863.

33. Jiang, X.; Zhou, H.; Quan, W.; Hu, Q.; Baker, J.S.; Gu, Y. Ground reaction force differences between bionic shoes and neutral running shoes in recreational male runners before and after a 5 km run. International Journal of Environmental Research and Public Health 2021, 18, 9787.

34. Kapri, E.; Mehta, M.; Singh, K. Biomechanics of running: An overview on gait cycle. Int. J. Phys. Educ. Fit. Sports 2021, 10, 1-9.

35. Dicharry, J. Kinematics and kinetics of gait: from lab to clinic. Clinics in sports medicine 2010, 29, 347-364.

36. Frigo, C.; Rabuffetti, M.; Kerrigan, D.; Deming, L.; Pedotti, A. Functionally oriented and clinically feasible quantitative gait analysis method. Medical and Biological Engineering and Computing 1998, 36, 179-185.

37. Nester, C.; Jones, R.K.; Liu, A.; Howard, D.; Lundberg, A.; Arndt, A.; Lundgren, P.; Stacoff, A.; Wolf, P. Foot kinematics during walking measured using bone and surface mounted markers. Journal of biomechanics 2007, 40, 3412-3423.

38. Song, Y.; Cen, X.; Chen, H.; Sun, D.; Munivrana, G.; Bálint, K.; Bíró, I.; Gu, Y. The influence of running shoe with different carbon-fiber plate designs on internal foot mechanics: A pilot computational analysis. Journal of Biomechanics 2023, 153, 111597.

39. Zhou, H.; Xu, D.; Quan, W.; Ugbolue, U.C.; Zhou, Z.; Gu, Y. Can the entire function of the foot be concentrated in the forefoot area during the running stance phase? A finite element study of different shoe soles. Journal of Human Kinetics 2024, 92, 5.

40. Giannini, S. Gait analysis: methodologies and clinical applications; IOS press: 1994.

41. Cavanagh, P.R. The biomechanics of lower extremity action in distance running. Foot & ankle 1987, 7, 197-217.

42. Nagel, A.; Fernholz, F.; Kibele, C.; Rosenbaum, D. Long distance running increases plantar pressures beneath the metatarsal heads: a barefoot walking investigation of 200 marathon runners. Gait & posture 2008, 27, 152-155.

43. Logan, S.; Hunter, I.; Hopkins, J.T.; Feland, J.B.; Parcell, A.C. Ground reaction force differences between running shoes, racing flats, and distance spikes in runners. Journal of sports science & medicine 2010, 9, 147.

44. Smoliga, J.M.; Myers, J.B.; Redfern, M.S.; Lephart, S.M. Reliability and precision of EMG in leg, torso, and arm muscles during running. Journal of Electromyography and Kinesiology 2010, 20, e1-e9.

45. Kyröläinen, H.; Avela, J.; Komi, P.V. Changes in muscle activity with increasing running speed. Journal of sports sciences 2005, 23, 1101-1109.

46. DeVita, P. The selection of a standard convention for analyzing gait data based on the analysis of relevant biomechanical factors. Journal of biomechanics 1994, 27, 501-508.

47. Murley, G.S.; Landorf, K.B.; Menz, H.B.; Bird, A.R. Effect of foot posture, foot orthoses and footwear on lower limb muscle activity during walking and running: a systematic review. Gait & posture 2009, 29, 172-187.

48. Fields, K.B.; Sykes, J.C.; Walker, K.M.; Jackson, J.C. Prevention of running injuries. Current sports medicine reports 2010, 9, 176-182.

49. Lysholm, J.; Wiklander, J. Injuries in runners. The American journal of sports medicine 1987, 15, 168-171.

50. Macera, C.A.; Pate, R.R.; Powell, K.E.; Jackson, K.L.; Kendrick, J.S.; Craven, T.E. Predicting lower-extremity injuries among habitual runners. Archives of internal medicine 1989, 149, 2565-2568.

51. Hreljac, A. Impact and overuse injuries in runners. Medicine & Science in Sports & Exercise 2004, 36, 845-849.

52. Ferber, R.; Hreljac, A.; Kendall, K.D. Suspected mechanisms in the cause of overuse running injuries: a clinical review. Sports health 2009, 1, 242-246.

53. Hreljac, A. Etiology, prevention, and early intervention of overuse injuries in runners: a biomechanical perspective. Physical Medicine and Rehabilitation Clinics 2005, 16, 651-667.

54. Hreljac, A.; Marshall, R.N.; Hume, P.A. Evaluation of lower extremity overuse injury potential in runners. Medicine & Science in Sports & Exercise 2000, 32, 1635-1641.

55. Dempster, J.; Dutheil, F.; Ugbolue, U.C. The Prevalence of Lower Extremity Injuries in Running and Associated Risk Factors: A Systematic Review. Physical Activity & Health (2515-2270) 2021, 5.

56. Saragiotto, B.T.; Yamato, T.P.; Hespanhol Junior, L.C.; Rainbow, M.J.; Davis, I.S.; Lopes, A.D. What are the main risk factors for running-related injuries? Sports medicine 2014, 44, 1153-1163.

57. Clement, D.; Taunton, J.; Smart, G.; McNicol, K. A survey of overuse running injuries. The Physician and sportsmedicine 1981, 9, 47-58.

58. Taunton, J.E.; Ryan, M.B.; Clement, D.; McKenzie, D.C.; Lloyd-Smith, D.R.; Zumbo, B.D. A retrospective case-control analysis of 2002 running injuries. British journal of sports medicine 2002, 36, 95-101.

59. Jiang, X.; Sárosi, J.; Bíró, I. Characteristics of lower limb running-related injuries in trail runners: a systematic review. Physical Activity and Health 2024, 8.

60. Rolf, C. Overuse injuries of the lower extremity in runners. Scandinavian journal of medicine & science in sports 1995, 5, 181-190.

61. Williams Iii, D.S.; McClay, I.S.; Hamill, J. Arch structure and injury patterns in runners. Clinical biomechanics 2001, 16, 341-347.

62. Gu, Y.; Li, J.; Ren, X.; Lake, M.J.; Zeng, Y. Heel skin stiffness effect on the hind foot biomechanics during heel strike. Skin Research and Technology 2010, 16, 291-296.

63. Bertelsen, M.; Hulme, A.; Petersen, J.; Brund, R.K.; Sørensen, H.; Finch, C.; Parner, E.T.; Nielsen, R. A framework for the etiology of running - related injuries. Scandinavian journal of medicine & science in sports 2017, 27, 1170-1180.

64. Willwacher, S.; Kurz, M.; Robbin, J.; Thelen, M.; Hamill, J.; Kelly, L.; Mai, P. Running-related biomechanical risk factors for overuse injuries in distance runners: a systematic review considering injury specificity and the potentials for future research. Sports Medicine 2022, 52, 1863-1877.

65. Ceyssens, L.; Vanelderen, R.; Barton, C.; Malliaras, P.; Dingenen, B. Biomechanical risk factors associated with running-related injuries: a systematic review. Sports medicine 2019, 49, 1095-1115.

66. Johnson, C.D.; Tenforde, A.S.; Outerleys, J.; Reilly, J.; Davis, I.S. Impact-related ground reaction forces are more strongly associated with some running injuries than others. The American journal of sports medicine 2020, 48, 3072-3080.

67. Van der Worp, H.; Vrielink, J.W.; Bredeweg, S.W. Do runners who suffer injuries have higher vertical ground reaction forces than those who remain injury-free? A systematic review and meta-analysis. British journal of sports medicine 2016, 50, 450-457.

68. Zadpoor, A.A.; Nikooyan, A.A. The relationship between lower-extremity stress fractures and the ground reaction force: a systematic review. Clinical biomechanics 2011, 26, 23-28.

69. Pohl, M.B.; Mullineaux, D.R.; Milner, C.E.; Hamill, J.; Davis, I.S. Biomechanical predictors of retrospective tibial stress fractures in runners. Journal of biomechanics 2008, 41, 1160-1165.

70. Kluitenberg, B.; van der Worp, H.; Huisstede, B.M.; Hartgens, F.; Diercks, R.; Verhagen, E.; van Middelkoop, M. The NLstart2run study: Training-related factors associated with running-related injuries in novice runners. Journal of Science and Medicine in Sport 2016, 19, 642-646.

71. Stefanyshyn, D.J.; Stergiou, P.; Lun, V.M.; Meeuwisse, W.H.; Worobets, J.T. Knee angular impulse as a predictor of patellofemoral pain in runners. The American journal of sports medicine 2006, 34, 1844-1851.

72. Li, S.; Zhang, Y.; Gu, Y.; Ren, J. Stress distribution of metatarsals during forefoot strike versus rearfoot strike: A finite element study. Computers in biology and medicine 2017, 91, 38-46.

73. Mei, Q.; Gu, Y.; Xiang, L.; Baker, J.S.; Fernandez, J. Foot pronation contributes to altered lower extremity loading after long distance running. Frontiers in physiology 2019, 10, 573.

74. Anderson, L.M.; Martin, J.F.; Barton, C.J.; Bonanno, D.R. What is the effect of changing running step rate on injury, performance and biomechanics? A systematic review and meta-analysis. Sports Medicine-Open 2022, 8, 112.

75. Knobloch, K.; Yoon, U.; Vogt, P.M. Acute and overuse injuries correlated to hours of training in master running athletes. Foot & ankle international 2008, 29, 671-676.

76. Arnold, M.J.; Moody, A.L. Common running injuries: evaluation and management. American family physician 2018, 97, 510-516.

77. Gefen, A. Biomechanical analysis of fatigue-related foot injury mechanisms in athletes and recruits during intensive marching. Medical and Biological Engineering and Computing 2002, 40, 302-310.

78. Gijon-Nogueron, G.; Fernandez-Villarejo, M. Risk factors and protective factors for lower-extremity running injuries: a systematic review. Journal of the American Podiatric Medical Association 2015, 105, 532-540.

79. Doherty, C.; Bleakley, C.; Delahunt, E.; Holden, S. Treatment and prevention of acute and recurrent ankle sprain: an overview of systematic reviews with meta-analysis. British journal of sports medicine 2017, 51, 113-125.

80. Jiang, X.; Yang, X.; Zhou, H.; Baker, J.S.; Gu, Y. Prolonged running using bionic footwear influences lower limb biomechanics. In Proceedings of Healthcare; p. 236.

81. Sherry, M. Examination and treatment of hamstring related injuries. Sports health 2012, 4, 107-114.

82. Heiderscheit, B.C.; Sherry, M.A.; Silder, A.; Chumanov, E.S.; Thelen, D.G. Hamstring strain injuries: recommendations for diagnosis, rehabilitation, and injury prevention. journal of orthopaedic & sports physical therapy 2010, 40, 67-81.

83. Ali, K.; Leland, J.M. Hamstring strains and tears in the athlete. Clinics in sports medicine 2012, 31, 263-272.

84. Linton, L.; Valentin, S. Running with injury: A study of UK novice and recreational runners and factors associated with running related injury. Journal of science and medicine in sport 2018, 21, 1221-1225.

85. Nielsen, R.O.; Buist, I.; Sørensen, H.; Lind, M.; Rasmussen, S. Training errors and running related injuries: a systematic review. International journal of sports physical therapy 2012, 7, 58.

86. Van der Worp, M.P.; Ten Haaf, D.S.; van Cingel, R.; de Wijer, A.; Nijhuis-van der Sanden, M.W.; Staal, J.B. Injuries in runners; a systematic review on risk factors and sex differences. PloS one 2015, 10, e0114937.

87. Hulme, A.; Nielsen, R.O.; Timpka, T.; Verhagen, E.; Finch, C. Risk and protective factors for middle-and long-distance running-related injury. Sports Medicine 2017, 47, 869-886.

88. Schmitz, A.; Russo, K.; Edwards, L.; Noehren, B. Do novice runners have weak hips and bad running form? Gait & posture 2014, 40, 82-86.

89. Boyer, K.A.; Silvernail, J.F.; Hamill, J. The role of running mileage on coordination patterns in running. Journal of applied biomechanics 2014, 30, 649-654.

90. Quan, W.; Ren, F.; Sun, D.; Fekete, G.; He, Y. Do Novice Runners Show Greater Changes in Biomechanical Parameters. Applied Bionics and Biomechanics 2021.

91. Nakayama, Y.; Kudo, K.; Ohtsuki, T. Variability and fluctuation in running gait cycle of trained runners and non-runners. Gait & posture 2010, 31.

92. Maas, E.; Bie, J.D.; Vanfleteren, R.; Hoogkamer, W.; Vanwanseele, B. Novice runners show greater changes in kinematics with fatigue compared with competitive runners. Sports Biomech 2017, 17, 1-11.

93. Giandolini, M.; Vernillo, G.; Samozino, P.; Horvais, N.; Edwards, W.B.; Morin, J.-B.; Millet, G.Y. Fatigue associated with prolonged graded running. European journal of applied physiology 2016, 116, 1859-1873.

94. Apte, S.; Prigent, G.; Stöggl, T.; Martínez, A.; Snyder, C.; Gremeaux-Bader, V.; Aminian, K. Biomechanical response of the lower extremity to running-induced acute fatigue: a systematic review. Frontiers in physiology 2021, 12, 646042.

95. Agresta, C.E.; Peacock, J.; Housner, J.; Zernicke, R.F.; Zendler, J.D. Experience does not influence injury-related joint kinematics and kinetics in distance runners. Gait & Posture 2018, 61, 13-18.

96. Middelkoop, M.V.; Kolkman, J.; Ochten, J.V.; Bierma-Zeinstra, S.M.A.; Koes, B.W. Risk factors for lower extremity injuries among male marathon runners. Scand J Med Sci Sports 2010, 18, 691-697.

97. Nielsen, R.O.; Bertelsen, M.L.; Parner, E.T.; Srensen, H.; Rasmussen, S. Running more than three kilometers during the first week of a running regimen may be associated with increased risk of injury in obese novice runners. International Journal of Sports Physical Therapy 2014, 9, 338-345.

98. Sinclair, J.; Selfe, J. Sex differences in knee loading in recreational runners. Journal of Biomechanics 2015, 48, 2171-2175.

99. Xie, P.-P.; István, B.; Liang, M. Sex-specific differences in biomechanics among runners: A systematic review with meta-analysis. Frontiers in Physiology 2022, 13.

100. Besson, T.; Macchi, R.; Rossi, J.; Morio, C.Y.M.; Kunimasa, Y.; Nicol, C.; Vercruyssen, F.; Millet, G.Y. Sex Differences in Endurance Running. Sports Medicine 2022, 52, 1235-1257.

101. Almonroeder, T.G.; Benson, L.C. Sex differences in lower extremity kinematics and patellofemoral kinetics during running. Journal of Sports Sciences 2017, 35, 1575-1581.

102. Bruno; Bazuelo-Ruiz; Juan; Durá-Gil; Nicolás; Palomares; Enrique; Medina; Salvador. Effect of fatigue and gender on kinematics and ground reaction forces variables in recreational runners. Peerj 2018.

103. Isherwood, J.; Wang, H.; Sterzing, T. Running biomechanics and running shoe perception of Chinese men and women. Footwear Science 2021, 13, 55-67.

104. Sinclair, J.; Greenhalgh, A.; Edmundson, C.; Hobbs, S. Gender Differences in the Kinetics and Kinematics of Distance Running: Implications for Footwear Design. 2012.

105. Peyré-Tartaruga, L.A.; Coertjens, M. Locomotion as a Powerful Model to Study Integrative Physiology: Efficiency, Economy, and Power Relationship. Frontiers in Physiology 2018, 9.

106. Folland, J.P.; Allen, S.J.; Black, M.I.; Handsaker, J.C.; Forrester, S.E. Running Technique is an Important Component of Running Economy and Performance. Med Sci Sports Exerc 2017, 49, 1412.

107. Daniels, J.; Daniels, N. Running economy of elite male and elite female runners. Medicine & Science in Sports & Exercise 1992, 24, 483???489.

108. Schache, A.G.; Dorn, T.W.; Williams, G.P.; Brown, N.A.; Pandy, M.G. Lower-limb muscular strategies for increasing running speed. journal of orthopaedic & sports physical therapy 2014, 44, 813-824.

109. Orendurff, M.S.; Kobayashi, T.; Tulchin-Francis, K.; Tullock, A.M.H.; Villarosa, C.; Chan, C.; Kraus, E.; Strike, S. A little bit faster: Lower extremity joint kinematics and kinetics as recreational runners achieve faster speeds. Journal of biomechanics 2018, 71, 167-175.

110. Farris, D.J.; Sawicki, G.S. The mechanics and energetics of human walking and running: a joint level perspective. Journal of The Royal Society Interface 2012, 9, 110-118.

111. Floría, P.; Sánchez-Sixto, A.; Harrison, A.J.; Ferber, R. The effect of running speed on joint coupling coordination and its variability in recreational runners. Human Movement Science 2019, 66, 449-458.

112. Grabowski, A.M.; Kram, R. Effects of velocity and weight support on ground reaction forces and metabolic power during running. Journal of applied biomechanics 2008, 24, 288-297.

113. Schache, A.G.; Blanch, P.D.; Dorn, T.W.; Brown, N.A.; Rosemond, D.; Pandy, M.G. Effect of running speed on lower limb joint kinetics. Medicine & Science in Sports & Exercise 2011, 43, 1260-1271.

114. Mercer, J.A.; Bezodis, N.E.; Russell, M.; Purdy, A.; DeLion, D. Kinetic consequences of constraining running behavior. Journal of sports science & medicine 2005, 4, 144.

115. Girard, O.; Morin, J.-B.; Ryu, J.; Read, P.; Townsend, N. Running velocity does not influence lower limb mechanical asymmetry. Frontiers in sports and active living 2019, 1, 36.

116. Muñoz-Jimenez, M.; Latorre-Román, P.; Soto-Hermoso, V.; García-Pinillos, F. Influence of shod/unshod condition and running speed on foot-strike patterns, inversion/eversion, and vertical foot rotation in endurance runners. Journal of sports sciences 2015, 33, 2035-2042.

117. Nielsen, R.O.; Nohr, E.A.; Rasmussen, S.; Sørensen, H. Classifying running‐related injuries based upon etiology, with emphasis on volume and pace. International journal of sports physical therapy 2013, 8, 172.

118. Petersen, J.; Nielsen, R.O.; Rasmussen, S.; Sørensen, H. Comparisons of increases in knee and ankle joint moments following an increase in running speed from 8 to 12 to 16 km· h− 1. Clinical Biomechanics 2014, 29, 959-964.

119. Keller, T.S.; Weisberger, A.; Ray, J.L.; Hasan, S.; Shiavi, R.G.; Spengler, D.M. Relationship between vertical ground reaction force and speed during walking, slow jogging, and running. Clinical biomechanics 1996, 11, 253-259.

120. Edwards, W.B.; Taylor, D.; Rudolphi, T.J.; Gillette, J.C.; Derrick, T.R. Effects of running speed on a probabilistic stress fracture model. Clinical Biomechanics 2010, 25, 372-377.

121. Shipway, R.; Holloway, I. Running free: Embracing a healthy lifestyle through distance running. Perspectives in public health 2010, 130, 270-276.

122. Gallo, R.A.; Plakke, M.; Silvis, M.L. Common leg injuries of long-distance runners: anatomical and biomechanical approach. Sports health 2012, 4, 485-495.

123. Van Mechelen, W. Running injuries: a review of the epidemiological literature. Sports medicine 1992, 14, 320-335.

124. Kim, H.K.; Mirjalili, S.A.; Fernandez, J. Gait kinetics, kinematics, spatiotemporal and foot plantar pressure alteration in response to long-distance running: Systematic review. Human movement science 2018, 57, 342-356.

125. Millet, G.Y.; Morin, J.-B.; Degache, F.; Edouard, P.; Feasson, L.; Verney, J.; Oullion, R. Running from Paris to Beijing: biomechanical and physiological consequences. European journal of applied physiology 2009, 107, 731-738.

126. Weist, R.; Eils, E.; Rosenbaum, D. The influence of muscle fatigue on electromyogram and plantar pressure patterns as an explanation for the incidence of metatarsal stress fractures. The American journal of sports medicine 2004, 32, 1893-1898.

127. Mizrahi, J.; Verbitsky, O.; Isakov, E.; Daily, D. Effect of fatigue on leg kinematics and impact acceleration in long distance running. Human movement science 2000, 19, 139-151.

128. Quan, W.; Ren, F.; Xu, D.; Gusztav, F.; Baker, J.S.; Gu, Y. Effects of fatigue running on joint mechanics in female runners: A prediction study based on a partial least squares algorithm. Frontiers in Bioengineering and Biotechnology 2021, 9, 746761.

129. Gao, Z.; Fekete, G.; Baker, J.S.; Liang, M.; Xuan, R.; Gu, Y. Effects of running fatigue on lower extremity symmetry among amateur runners: From a biomechanical perspective. Frontiers in physiology 2022, 13, 899818.

130. Blagrove, R.; Hayes, P. The science and practice of middle and long distance running; Routledge: 2021.

131. Hajiloo, B.; Anbarian, M.; Esmaeili, H.; Mirzapour, M. The effects of fatigue on synergy of selected lower limb muscles during running. Journal of Biomechanics 2020, 103, 109692.

132. Mizrahi, J.; Verbitsky, O.; Isakov, E. Fatigue-related loading imbalance on the shank in running: a possible factor in stress fractures. Annals of biomedical engineering 2000, 28, 463-469.

133. Nummela, A.; Heath, K.; Paavolainen, L.; Lambert, M.I.; Gibson, A.S.C.; Rusko, H.; Noakes, T. Fatigue during a 5-km running time trial. International journal of sports medicine 2008, 29, 738-745.

134. Hettinga, F.J.; Edwards, A.M.; Hanley, B. The science behind competition and winning in athletics: using world-level competition data to explore pacing and tactics. Frontiers in Sports and Active Living 2019, 1, 11.

135. Zhou, H.; Xu, D.; Quan, W.; Liang, M.; Ugbolue, U.C.; Baker, J.S.; Gu, Y. A pilot study of muscle force between normal shoes and bionic shoes during men walking and running stance phase using opensim. In Proceedings of Actuators; p. 274.

136. Aljohani, M.; Kipp, K. Use of self-organizing maps to study sex-and speed-dependent changes in running biomechanics. Human Movement Science 2020, 72, 102649.

137. Xu, D.; Jiang, X.; Cen, X.; Baker, J.S.; Gu, Y. Single-leg landings following a volleyball spike may increase the risk of anterior cruciate ligament injury more than landing on both-legs. Applied Sciences 2020, 11, 130.

138. Delp, S.L.; Anderson, F.C.; Arnold, A.S.; Loan, P.; Habib, A.; John, C.T.; Guendelman, E.; Thelen, D.G. OpenSim: open-source software to create and analyze dynamic simulations of movement. IEEE transactions on biomedical engineering 2007, 54, 1940-1950.

139. Roldán, E.; Reeves, N.D.; Cooper, G.; Andrews, K. In vivo mechanical behaviour of the anterior cruciate ligament: a study of six daily and high impact activities. Gait & posture 2017, 58, 201-207.

140. Zajac, F.E. Muscle and tendon: properties, models, scaling, and application to biomechanics and motor control. Critical reviews in biomedical engineering 1989, 17, 359-411.

141. Richards, C.; Higginson, J. Knee contact force in subjects with symmetrical OA grades: differences between OA severities. Journal of biomechanics 2010, 43, 2595-2600.

142. Kluitenberg, B.; Bredeweg, S.W.; Zijlstra, S.; Zijlstra, W.; Buist, I. Comparison of vertical ground reaction forces during overground and treadmill running. A validation study. BMC musculoskeletal disorders 2012, 13, 1-8.

143. Blackmore, T.; Willy, R.W.; Creaby, M.W. The high frequency component of the vertical ground reaction force is a valid surrogate measure of the impact peak. Journal of Biomechanics 2016, 49, 479-483.

144. Samaan, C.D.; Rainbow, M.J.; Davis, I.S. Reduction in ground reaction force variables with instructed barefoot running. Journal of Sport and Health Science 2014, 3, 143-151.

145. Munro, C.F.; Miller, D.I.; Fuglevand, A.J. Ground reaction forces in running: a reexamination. Journal of biomechanics 1987, 20, 147-155.

146. Quan, W.; Zhou, H.; Xu, D.; Li, S.; Baker, J.S.; Gu, Y. Competitive and recreational running kinematics examined using principal components analysis. In Proceedings of Healthcare; p. 1321.

147. Brandon, S.C.; Graham, R.B.; Almosnino, S.; Sadler, E.M.; Stevenson, J.M.; Deluzio, K.J. Interpreting principal components in biomechanics: representative extremes and single component reconstruction. Journal of Electromyography and Kinesiology 2013, 23, 1304-1310.

148. Jiang, X.; Xu, D.; Fang, Y.; Bíró, I.; Baker, J.S.; Gu, Y. PCA of Running Biomechanics after 5 km between Novice and Experienced Runners. Bioengineering 2023, 10, 876.

149. Sheppard, P.S.; Stevenson, J.; Graham, R. Sex-based differences in lifting technique under increasing load conditions: A principal component analysis. Applied ergonomics 2016, 54, 186-195.

150. Braga, U.M.; Mendonça, L.D.; Mascarenhas, R.O.; Alves, C.O.; Renato Filho, G.; Resende, R.A. Effects of medially wedged insoles on the biomechanics of the lower limbs of runners with excessive foot pronation and foot varus alignment. Gait & posture 2019, 74, 242-249.

151. Wrigley, A.T.; Albert, W.J.; Deluzio, K.J.; Stevenson, J.M. Principal component analysis of lifting waveforms. Clinical Biomechanics 2006, 21, 567-578.

152. Foch, E.; Milner, C.E. The influence of iliotibial band syndrome history on running biomechanics examined via principal components analysis. Journal of biomechanics 2014, 47, 81-86.

153. Lee, M.; Roan, M.; Smith, B. An application of principal component analysis for lower body kinematics between loaded and unloaded walking. Journal of biomechanics 2009, 42, 2226-2230.

154. Yu, L.; Mei, Q.; Xiang, L.; Liu, W.; Mohamad, N.I.; István, B.; Fernandez, J.; Gu, Y. Principal component analysis of the running ground reaction forces with different speeds. Frontiers in bioengineering and biotechnology 2021, 9, 629809.

155. Portney, L.G.; Watkins, M.P. Foundations of clinical research: applications to practice; Pearson/Prentice Hall Upper Saddle River, NJ: 2009; Vol. 892.

156. Seth, A.; Hicks, J.L.; Uchida, T.K.; Habib, A.; Dembia, C.L.; Dunne, J.J.; Ong, C.F.; DeMers, M.S.; Rajagopal, A.; Millard, M. OpenSim: Simulating musculoskeletal dynamics and neuromuscular control to study human and animal movement. PLoS computational biology 2018, 14, e1006223.

157. Zeni, J.A.; Rudolph, K.; Higginson, J.S. Alterations in quadriceps and hamstrings coordination in persons with medial compartment knee osteoarthritis. Journal of Electromyography and Kinesiology 2010, 20, 148-154.

158. Hurd, W.J.; Snyder‐Mackler, L. Knee instability after acute ACL rupture affects movement patterns during the mid‐stance phase of gait. Journal of Orthopaedic Research 2007, 25, 1369-1377.

159. Hart, J.M.; Ko, J.-W.K.; Konold, T.; Pietrosimione, B. Sagittal plane knee joint moments following anterior cruciate ligament injury and reconstruction: a systematic review. Clinical biomechanics 2010, 25, 277-283.

160. Aghdam, H.A.; Haghighat, F.; Rezaie, M.; Kavyani, M.; Karimi, M.T. Comparison of the knee joint reaction force between individuals with and without acute anterior cruciate ligament rupture during walking. Journal of Orthopaedic Surgery and Research 2022, 17, 250.

161. Swinnen, W.; Hoogkamer, W.; Delabastita, T.; Aeles, J.; De Groote, F.; Vanwanseele, B. Effect of habitual foot-strike pattern on the gastrocnemius medialis muscle-tendon interaction and muscle force production during running. Journal of applied physiology 2019, 126, 708-716.

162. Karamanidis, K.; Arampatzis, A. Aging and running experience affects the gearing in the musculoskeletal system of the lower extremities while walking. Gait & posture 2007, 25, 590-596.

163. Stensgaard Stoltze, J.; Rasmussen, J.; Skipper Andersen, M. On the biomechanical relationship between applied hip, knee and ankle joint moments and the internal knee compressive forces. International Biomechanics 2018, 5, 63-74.

164. Miller, R.H.; Edwards, W.B.; Brandon, S.; Morton, A.M.; Deluzio, K.J. Why don't most runners get knee osteoarthritis? A case for per-unit-distance loads. Medicine and science in sports and exercise 2014, 46, 572-579.

165. D’Lima, D.D.; Patil, S.; Steklov, N.; Chien, S.; Colwell Jr, C.W. In vivo knee moments and shear after total knee arthroplasty. Journal of biomechanics 2007, 40, S11-S17.

166. Winby, C.R.; Lloyd, D.G.; Besier, T.F.; Kirk, T.B. Muscle and external load contribution to knee joint contact loads during normal gait. Journal of biomechanics 2009, 42, 2294-2300.

167. Kluitenberg, B.; van Middelkoop, M.; Diercks, R.; van der Worp, H. What are the differences in injury proportions between different populations of runners? A systematic review and meta-analysis. Sports medicine 2015, 45, 1143-1161.

168. Tonoli, D.C.; Cumps, E.; Aerts, I.; Verhagen, E.; Meeusen, R. Incidence, risk factors and prevention of running related injuries in long-distance running: a systematic review. Sport & Geneeskunde 2010, 43.

169. Besier, T.F.; Fredericson, M.; Gold, G.E.; Beaupré, G.S.; Delp, S.L. Knee muscle forces during walking and running in patellofemoral pain patients and pain-free controls. Journal of biomechanics 2009, 42, 898-905.

170. Scarton, A.; Jonkers, I.; Guiotto, A.; Spolaor, F.; Guarneri, G.; Avogaro, A.; Cobelli, C.; Sawacha, Z. Comparison of lower limb muscle strength between diabetic neuropathic and healthy subjects using OpenSim. Gait & posture 2017, 58, 194-200.

171. Weinhandl, J.T.; Bennett, H.J. Musculoskeletal model choice influences hip joint load estimations during gait. Journal of biomechanics 2019, 91, 124-132.

172. Vtasalo, J.T.; Kvist, M. Some biomechanical aspects of the foot and ankle in athletes with and without shin splints. The American Journal of Sports Medicine 1983, 11, 125-130.

173. Maas, E.; De Bie, J.; Vanfleteren, R.; Hoogkamer, W.; Vanwanseele, B. Novice runners show greater changes in kinematics with fatigue compared with competitive runners. Sports biomechanics 2018, 17, 350-360.

174. Hamill, J.; Derrick, T.R.; Holt, K.G. Shock attenuation and stride frequency during running. Human movement science 1995, 14, 45-60.

175. Brown, A.M.; Zifchock, R.A.; Lenhoff, M.; Song, J.; Hillstrom, H.J. Hip muscle response to a fatiguing run in females with iliotibial band syndrome. Human movement science 2019, 64, 181-190.

176. Willwacher, S.; Sanno, M.; Brüggemann, G.-P. Fatigue matters: An intense 10 km run alters frontal and transverse plane joint kinematics in competitive and recreational adult runners. Gait & posture 2020, 76, 277-283.

177. O'Connor, K.M.; Bottum, M.C. Differences in cutting knee mechanics based on principal components analysis. Medicine and science in sports and exercise 2009, 41, 867-878.

178. Dierks, T.A.; Davis, I.S.; Hamill, J. The effects of running in an exerted state on lower extremity kinematics and joint timing. Journal of biomechanics 2010, 43, 2993-2998.

179. Kuhman, D.J.; Paquette, M.R.; Peel, S.A.; Melcher, D.A. Comparison of ankle kinematics and ground reaction forces between prospectively injured and uninjured collegiate cross country runners. Human movement science 2016, 47, 9-15.

180. Ferber, R.; Noehren, B.; Hamill, J.; Davis, I. Competitive female runners with a history of iliotibial band syndrome demonstrate atypical hip and knee kinematics. journal of orthopaedic & sports physical therapy 2010, 40, 52-58.

181. Noehren, B.; Hamill, J.; Davis, I. Prospective evidence for a hip etiology in patellofemoral pain. Medicine and science in sports and exercise 2013, 45, 1120-1124.

182. Giandonato, J.A.; Tringali, V.M.; Thoms, R.C. Improving Mental Health through Physical Activity: A Narrative Literature Review. Physical Activity & Health (2515-2270) 2021, 5.

183. Lieberman, D.E.; Warrener, A.G.; Wang, J.; Castillo, E.R. Effects of stride frequency and foot position at landing on braking force, hip torque, impact peak force and the metabolic cost of running in humans. Journal of Experimental Biology 2015, 218, 3406-3414.

184. Davis, E.M.; Hubley-Kozey, C.L.; Landry, S.C.; Ikeda, D.M.; Stanish, W.D.; Wilson, J.L.A. Longitudinal evidence links joint level mechanics and muscle activation patterns to 3-year medial joint space narrowing. Clinical Biomechanics 2019, 61, 233-239.

185. Mantovani, G.; Lamontagne, M.; Varin, D.; Cerulli, G.G.; Beaulé, P.E. Comparison of total hip arthroplasty surgical approaches by Principal Component Analysis. Journal of biomechanics 2012, 45, 2109-2115.

186. Riazati, S.; Caplan, N.; Matabuena, M.; Hayes, P.R. Fatigue induced changes in muscle strength and gait following two different intensity, energy expenditure matched runs. Frontiers in bioengineering and biotechnology 2020, 8, 360.

187. Abbasi, A.; Yazdanbakhsh, F.; Tazji, M.K.; Ataabadi, P.A.; Svoboda, Z.; Nazarpour, K.; Vieira, M.F. A comparison of coordination and its variability in lower extremity segments during treadmill and overground running at different speeds. Gait & posture 2020, 79, 139-144.

188. Van Hooren, B.; Fuller, J.T.; Buckley, J.D.; Miller, J.R.; Sewell, K.; Rao, G.; Barton, C.; Bishop, C.; Willy, R.W. Is motorized treadmill running biomechanically comparable to overground running? A systematic review and meta-analysis of cross-over studies. Sports medicine 2020, 50, 785-813.

189. Boyer, K.A.; Freedman Silvernail, J.; Hamill, J. Age and sex influences on running mechanics and coordination variability. Journal of Sports Sciences 2017, 35, 2225-2231.

190. Hollander, K.; Rahlf, A.L.; Wilke, J.; Edler, C.; Steib, S.; Junge, A.; Zech, A. Sex-Specific Differences in Running Injuries: A Systematic Review with Meta-Analysis and Meta-Regression. Sports Medicine 2021, 51, 1011-1039.

191. Stickley, C.D.; Andrews, S.N.; Parke, E.A.; Hetzler, R.K. The Effectiveness of Scaling Procedures for Comparing Ground Reaction Forces. Journal of Biomechanics 2018, S0021929018304524-.

192. Morin, J.-B.; Slawinski, J.; Dorel, S.; de villareal, E.S.; Couturier, A.; Samozino, P.; Brughelli, M.; Rabita, G. Acceleration capability in elite sprinters and ground impulse: Push more, brake less? Journal of Biomechanics 2015, 48, 3149-3154.

193. Cavanagh, P.R.; Kram, R. Stride length in distance running: velocity, body dimensions, and added mass effects. Medicine and science in sports and exercise 1989, 21, 467-479.

194. Davis, I.S.; Bowser, B.J.; Mullineaux, D.R. Greater vertical impact loading in female runners with medically diagnosed injuries: a prospective investigation. British Journal of Sports Medicine 2016, 50.

195. Vannatta, C.N.; Heinert, B.L.; Kernozek, T.W. Biomechanical risk factors for running-related injury differ by sample population: A systematic review and meta-analysis. Clinical Biomechanics 2020, 75, 104991.

196. Wright, A.A.; Taylor, J.B.; Ford, K.R.; Siska, L.; Smoliga, J.M. Risk factors associated with lower extremity stress fractures in runners: a systematic review with meta-analysis. British Journal of Sports Medicine 2015, 49, 1517-1523.

197. Yokoyama, H.; Ogawa, T.; Kawashima, N.; Shinya, M.; Nakazawa, K. Distinct sets of locomotor modules control the speed and modes of human locomotion. Scientific Reports 2016, 6, 36275.

198. Hollis, C.R.; Koldenhoven, R.M.; Resch, J.E.; Hertel, J. Running biomechanics as measured by wearable sensors: effects of speed and surface. Sports Biomechanics 2021, 20, 521-531.

199. Williams, L.R.; Standifird, T.W.; Creer, A.; Fong, H.B.; Powell, D.W. Ground reaction force profiles during inclined running at iso-efficiency speeds. Journal of Biomechanics 2020, 113, 110107.

200. Breine, B.; Malcolm, P.; Galle, S.; Fiers, P.; Frederick, E.C.; De Clercq, D. Running speed-induced changes in foot contact pattern influence impact loading rate. European Journal of Sport Science 2019, 19, 774-783.

201. Matijevich, E.S.; Branscombe, L.M.; Scott, L.R.; Zelik, K.E. Ground reaction force metrics are not strongly correlated with tibial bone load when running across speeds and slopes: Implications for science, sport and wearable tech. PLOS ONE 2019, 14, e0210000.

202. Napier, C.; MacLean, C.L.; Maurer, J.; Taunton, J.E.; Hunt, M.A. Kinetic risk factors of running-related injuries in female recreational runners. Scandinavian Journal of Medicine & Science in Sports 2018, 28, 2164-2172.

203. Fukuchi, R.K.; Fukuchi, C.A.; Duarte, M. A public dataset of running biomechanics and the effects of running speed on lower extremity kinematics and kinetics. PeerJ 2017, 5, e3298.

204. Chang, Y.-H.; Huang, H.-W.C.; Hamerski, C.M.; Kram, R. The Independent Effects of Gravity and Inertia on Running Mechanics. Journal of Experimental Biology 2000, 203, 229-238.

205. Mo, S.; Lau, F.O.Y.; Lok, A.K.Y.; Chan, Z.Y.S.; Zhang, J.H.; Shum, G.; Cheung, R.T.H. Bilateral asymmetry of running gait in competitive, recreational and novice runners at different speeds. Human Movement Science 2020, 71, 102600.

ABBREVATION

3D: three-dimensional
Adduct/Abduct: adduction/abduction
ANOVA: analysis of variance
BFLH: biceps femoris long head
BMI: body mass index
BW: body weight
CMC: computed muscle control
CNS: central nervous system
COM: center of mass
COP: center of pressure
Dorsi/Plant: dorsiflexion/plantarflexion
EMG: electromyography
Ext/Flex: extension/flexion
Flex/Ext: flexion/extension
GRF: ground reaction force
GL: gastrocnemius lateralis
GM: gastrocnemius medialis
Int Rot/Ext Rot: internal rotation/external rotation
Invert/Evert: inversion/eversion
MRI: magnetic resonance images
MVC: maximal voluntary contractions
PCA: principal component analysis
RF: rectus femoris
RMS: root mean square
ROM: range of motion
RRI: running-related injury
SO: soleus
SPM1d: one-dimensional statistical parametric mapping
TA: tibialis anterior

LIST OF TABLES

LIST OF FIGURES

ACKNOWLEDGEMENTS

As I sit down to write this section of my dissertation, it signals the nearing end of my PhD journey, a significant milestone in my academic and personal life. Reflecting on the past two years at Óbuda University, I find myself filled with deep gratitude and appreciation for all the support, guidance, and encouragement I have received throughout this challenging yet rewarding journey.

First and foremost, I would like to express my sincere gratitude to my primary supervisor, Prof. Dr. habil. Bíró István. His unwavering support, both academically and personally, has been a cornerstone of my doctoral life. Not only is he a professor to be respected and admired for his brilliance in academics, but he is also one of the kindest and most sincere individuals I have encountered. His genuine care for his students and his willingness to go above and beyond to provide guidance has made a profound impact on my journey. I am incredibly fortunate to have had him as my supervisor.

I also extend my heartfelt thanks to my co-supervisor, Prof. Dr. Yaodong Gu, who has been my mentor not only during my PhD but also throughout my master's studies. His dedication to helping me overcome numerous academic challenges has been invaluable. His continuous support and mentorship have significantly contributed to my growth as a researcher. The profound influence of his guidance is something I will always cherish.

In addition, I owe a debt of gratitude to several professors and colleagues, without whose assistance and support, it would have been difficult for me to complete my studies successfully. I would like to thank Prof. Dr. Zoltán Rajnai, Prof. Dr. Lívia Cvetityánin, Dr. Fekete Gusztáv, Dr. József Sárosi, and Dr. János Simon. These professors and mentors have shown great care for my academic progress and have always been willing to provide help when needed. Their kindness and wisdom have been instrumental in my achievements, and I am deeply appreciative of their guidance.

I am also grateful to my colleagues, especially Dr. Wenjing Quan, Dr. Yao Meng, Dr. Xuanzhen Cen, Dr. Qiaolin Zhang, Dr. Hairong Chen, and all the PhD students I have worked with. Many tasks in research require collaboration, and I am incredibly thankful for their assistance and the countless discussions that have helped me grow intellectually. Without their support, completing numerous tasks and overcoming obstacles would have been much more difficult.

Secondly, I would like to extend my deepest appreciation to the institutions and organizations that have supported me throughout my academic journey. The Doctoral School on Safety and Security Sciences at Óbuda University, Faculty of Engineering at the University of Szeged, Research Academy of Grand Health & Faculty of Sports Science at Ningbo University, the Stipendium Hungaricum Programme, Tempus Public Foundation, and the China Scholarship Council (CSC) have provided me with both the facilities and financial support necessary for my research. I am incredibly grateful for the opportunities and resources they have afforded me, which have allowed me to pursue my research at the highest level.

Lastly, but certainly not least, I must express my deepest gratitude to my family. Without their unconditional support, I would not have made it this far. To my father, Mr. Chunfang Jiang, and my mother, Mrs. Xialian Jiang, thank you for believing in me and for your endless encouragement. Your love and faith in me have been the driving force behind my perseverance and determination. To my dearest friends, Mengjie Lin and Peiwei Liang, your friendship and unwavering support over the years have been invaluable. You have been by my side through thick and thin, and your belief in me has been a source of strength. I am forever grateful to you both, and I love you dearly.

As I reach the culmination of this chapter of my life, I am filled with a profound sense of gratitude for everyone who has been a part of this journey. This dissertation is not only a reflection of my work but also of the collective efforts of all those who have supported and believed in me along the way. Thank you all from the bottom of my heart.