EFFECT OF FATIGUE ON LOWER EXTREMITY SYMMETRY IN RECREATIONAL RUNNERS

Pan Hanyu

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Faculty of Sport and Health Sciences
University of Jyväskylä
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Supervisors: Janne Avela, Neil Cronin,
Krista Vohlakari

ABSTRACT

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Running is one of the most popular physical activities in the world. However, running is an impact activity that can lead to overuse injuries such as tibia stress fractures and patellofemoral pain syndrome. It is reported that both inter-limb asymmetries and fatigue are risk factors for non-contact injuries (Heil et al., 2020). Thus, this thesis aimed to examine the impact of running-induced fatigue on lower extremity symmetry in recreational runners. Twenty recreational runners were recruited in this study. Participants performed a fatigue protocol on a motorized treadmill, marker trajectories were collected before and after the fatigue protocol using a Vicon eight-camera motion capture system. OpenSim (Delp et al., 2007) was used for calculating hip, knee, and ankle joint kinematics.

Inter-limb asymmetries can be analyzed using discrete and continuous methods. Symmetry angle (SA) is the measurement to quantify the differences between the lower extremities. SA serves as a robust tool since it is a dimensionless metric without the reference number. A Wilcoxon signed-rank test was used to compare the SA for joint angles before and after fatigue protocol. In comparison, Statistical parametric mapping (SPM) as a continuous analysis was applied to the pre- and post-fatigue measurement to analyze the time series data.

After fatigue, the SA of the hip peak abduction angle significantly decreased by 7.9% after the running-induced fatigue protocol ($p=0.021$). Meanwhile, the SA of peak internal rotation after fatigue was significantly higher (9.8%) than that before fatigue ($p=0.023$). However, SPM results failed to show any difference in the pre- and post-fatigue measurements between the limbs. Although SA has been successfully proved as a reliable indicator to quantify symmetry, future studies should consider continuity by using time-series symmetry indices.

Key words: symmetry angle, statistical parametric mapping, running-induced fatigue, recreational runners

ABBREVIATIONS

DK	direct kinematic
DoF	degree of freedom
GA	gait asymmetry
RoM	range of motion
IK	inverse kinematic
OMC	optical motion capture system
PD	pelvic depth
PH	pelvic height
PW	pelvic weight
PiG	plug in gait
RI	ratio index
SA	symmetry angle
SI	symmetry index
SPM	statistical parametric mapping

CONTENTS

1 INTRODUCTION

This study aimed to examine the effect of running-induced fatigue on lower extremity symmetry in recreational runners. Running is a popular physical activity in the world and can be practiced by everyone regardless of their age and fitness levels. Running is also associated with reduced risks of chronic disease and all-cause mortality, indicating that it is a key component of public health programs (Lee et al., 2014). However, running is an impact activity where the stress of lower limbs predisposes the runners to injuries. For instance, it has been reported that up to 79% of runners may obtain an injury of lower extremity because of jogging during their lifetime (van Gent et al., 2007). Overuse injuries account for 50% to 75% of sports injuries, which are caused by repetitive actions. Factors contributing to running injuries including previous sports injury history, lack of long-standing running experience, participation in running competitions, and running long distances per week (Quan et al., 2021).

Exercise-induced fatigue, resulting from the application of physical loads such as training or competition, potentially increases predisposition for non-contact injuries (Heil et al., 2020). Acute fatigue changes the biomechanical response; therefore, measurement must be made again during or after training interventions to investigate the influence of fatigue. Lower extremity biomechanical response includes kinematics, kinetics, and muscle activity of the leg during running, as well as spatiotemporal gait parameters (Apte et al., 2021). Three literature reviews have previously investigated the effect of fatigue on running kinematics: Winter et al. (2017) focused on the effects of fatigue on kinematics and kinetics during overground running, Kim et al. (2018) examined the effects of fatigue on foot plantar pressure and related kinematic quantities, and Apte et al. (2021) explored the impact of fatigue, fatigue severity, and running surface on a total of 42 quantities. However, there is no consensus about how running fatigue will affect the lower extremity biomechanics.

Lower limb asymmetry refers to differences in function, strength, and other parameters between limbs, and has been associated with an increased risk of injury. A previous study reported a higher injury incident rate of healthy participants with a symmetry score above 15%, compared with participants with a symmetry score below 15%, thus indicating that bilateral lower limb asymmetry may be a potential cause of injury (Gao et al., 2022). Both inter-limb asymmetries and exercise-induced fatigue are regarded as risk factors for non-contact injuries (Heil et al., 2020). However, the relationship between the running fatigue and lower limb asymmetry still remains unclear.

Limb asymmetry has been evaluated using different indices, which can be categorized into discrete symmetry indices and time-series analysis (Steinmetzer et al., 2022). Discrete indices include Symmetry Angle (SA), Symmetry Index (SI), Gait Asymmetry (GA), and Ratio Index (RI). For example, SA measures the angle between the line of symmetry and a reference line (Zifchock et al., 2008), and is not subject to artificial inflation (Hanley & Tucker, 2018). Despite the discrete symmetry indices provide information related to inter-limb asymmetry, time-series analysis offers a more comprehensive understanding by assessing both the magnitude and waveforms of data. Statistical parametric mapping (SPM) has been used as a measurement of differences between limbs (Vial et al., 2023; Hughes-Oliver et al., 2019), which suggests that SPM can be a reliable tool to assess inter-limb asymmetry during running.

Biomechanical changes in running can be analyzed using a variety of methods, including optical motion capture systems (OMC), force plates, and inertial measurement units (Thorpe et al., 2017). Optical motion capture system has been considered the gold standard because of its high precision and accuracy. Thus, OMC is the preferred method for analyzing more complex movements and larger body segments, as seen in sports and rehabilitation settings. Vicon (Vicon, Oxford, UK) is one of the OMC systems that is frequently used in the literature. After data collection, Vicon Nexus utilizes the Plug-in-Gait (PiG) model to calculate joint kinematics by direct kinematics. Nowadays, software such as OpenSim and AnyBody apply musculoskeletal models for biomechanics analysis, which employ inverse kinematics for joint angle calculation. Compared to Vicon Nexus, OpenSim allows for additional information such as joint reaction forces, joint moments, muscle force, and muscle activation (Delp et al., 2007). Furthermore, OpenSim is a free, open-source software that offers more opportunities for biomechanics researchers to simulate and analyze human locomotion.

The purpose of this work is to detect the effect of running-induced fatigue protocol on lower extremities. The examination utilized SA and SPM to quantify the asymmetry between limbs, aiming to provide more information about lower limb kinematics asymmetry in both non-fatigue and fatigue situations.

2 RUNNING FATIGUE

Running is a high-impact activity that exposes the lower extremities to repetitive forces. Running may potentially lead to overuse-related injuries such as tibia stress fractures (Clansey et al., 2012). Fatigue exacerbates the body's ability for attenuating forces thus leading to an increase in running injury risk (Clansey et al., 2012). Monitoring lower limb loading can provide valuable insights to prevent bone stress injuries and other overuse injuries. Hence, runners and coaches can design training programs with precise strategies to prevent injuries.

Fatigue is a complex phenomenon that occurs during both high- and low-intensity exercise. The onset and severity of fatigue depending on the intensity, duration, and type of exercise (Millet and Lepers, 2004). Fatigue results in deterioration of the neuromuscular system in endurance athletes. It is characterized by reductions in movement control and alterations in running kinematics (Luo et al., 2019). Such biomechanical change may decrease not only running performance but also possibly enhance injury risks (Hreljac et al., 2000). For example, many studies have shown that fatigue may lead to changes in running biomechanics, such as joint angles, ground reaction forces, and step frequency (Winter et al., 2017; Kim et al., 2018).

2.1 Biomechanics of running gait

The gait cycle contains two main phases: the stance phase and the swing phase (Figure 1). The stance phase is defined as some part of the foot remains in contact with the foot while the swing phase is when on foot is not in contact with the ground. The stance and swing phases occupy 60% and 40% of the gait cycle, respectively. Running included an aerial phase where no limbs are touching the ground (Tongen and Wunderlich, 2010).

To date, there have been many discussions related to the quantitative analysis of running (Wang et al., 2003; Chen et al., 2012; Tao et al., 2012). Quantified parameters can be divided into four categories: spatiotemporal, kinematic, kinetic, and derivate parameters. More specifically, spatiotemporal data consists of contact time, flight time, cadence (step frequency), step length, vertical oscillation, etc. Kinematic characteristics describe the segmental or joint movement, often in the sagittal, coronal (frontal), and transverse planes. In addition, other metrics, such as leg stiffness, vertical stiffness, variability, and SA, are derived from spatiotemporal parameters or kinematic data.

In terms of the spatiotemporal parameters, step frequency refers to the number of steps or strides taken during a given time. It can be determined by considering the interval between consecutive initial foot contacts (Encarnación-Martnez et al., 2021). Ground contact time, or stance time, is the period between initial foot contact and toe-off for the same foot. This duration, along with the time between toe-off from one foot to initial contact of the other foot (flight time), helps define step/stride time, which is the interval between two consecutive heel strikes of the same foot or opposite foot (Figure 1).

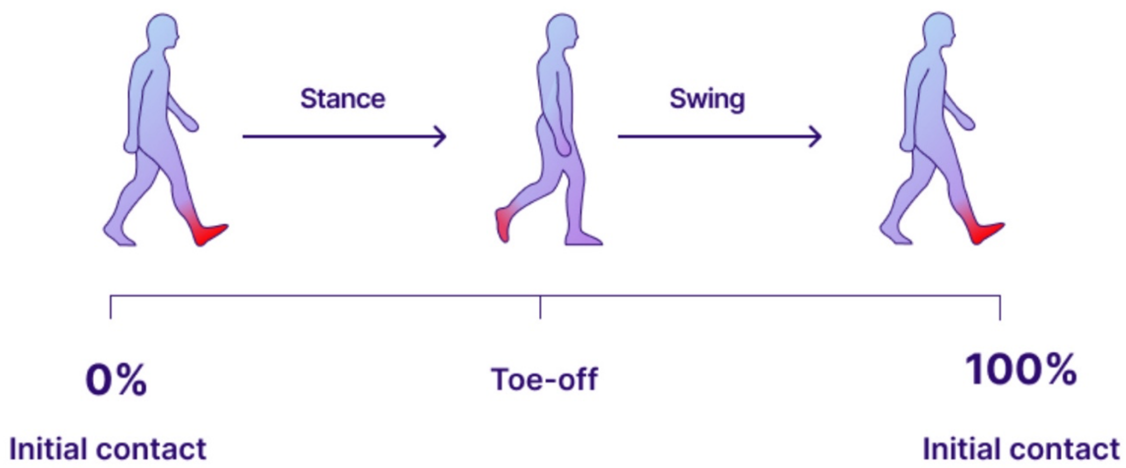

FIGURE 1. The gait cycle (Sanchez, 2022)

Step/stride length denotes the distance between successive points of initial contact of the same foot or opposite foot, while foot strike pattern, stride index, or foot strike angle refers to the moment, way, or angle when the foot first makes contact with the ground. Vertical speed and acceleration of the foot markers are used to identify foot strikes and toe-off events, respectively (Leitch et al., 2011). Gait events include the identification of any key occurrences, such as heel strike, toe-off, or mid-stance. Absorption time is the period between initial contact and maximum knee flexion, and propulsion time is the interval between maximum knee flexion and toe-off. Vertical oscillation, the up-and-down movement of an individual's center of mass during gait, can be estimated by calculating the mean vertical amplitude of the marker on the sacrum during the running trial (Maas et al., 2018).

Many studies have looked at the joint or segmental kinematics of the lower extremities (ankle, knee, and hip joints) and the torso (lumbar spine and thoracic spine joints) in sagittal, frontal, and transversal planes to include all significant degrees of freedom and constraints (Möhler et al., 2021). In the majority of studies, the maximal angle and range of motion (RoM) are extracted during the stance phase (Luo et al., 2019; Riazati et al., 2020; Dierks et al., 2010). RoM was determined by subtracting the maximum joint angle from the minimum joint angle for both the stance phase (right foot strike to right toe off) and the flight phase (right toe off to left foot strike). Because the RoM data clearly show the limits of mobility, they may be useful for evaluating adaptations to fatigue, particularly in terms of injuries. Increases in RoM might signify a higher risk of soft tissue damage since these tissues might be under more strain (Möhler et al., 2021).

Several systemic reviews have investigated the effects of running-induced fatigue on running kinematics. Zandbergen et al. (2023) assessed 33 articles and they found that after fatigue protocol, there was an increase in peak acceleration at the tibia and a reduction in leg stiffness. Peak acceleration is one of the parameters that quantify the amplitude of impact forces attenuation (Vanwanseele et al., 2020). High peak accelerations suggested an increased overuse injury risk since there is a high load applied on the body (Sheerin et al., 2019). One possible explanation for increased peak acceleration is that the coordination activation of the muscle around the hip, knee, and ankle joint decreases, resulting in a higher force impact. Moreover, knee flexion at initial contact and maximum knee flexion angle during the swing phase increased after running-induced fatigue. Compared to experienced runners, novice runners presented an increase in vertical center of mass displacements with fatigue. Experienced runners might be better at utilizing a more energy-efficient gait pattern with smaller vertical displacement of center of mass.

The results from Apte et al. (2021) are similar to those described above. Knee flexion angle at initial contact and peak tibial acceleration increased after fatigue protocol. The vertical stiffness was decreased across different fatigue protocols which is not supported by the meta-analyses result of Zandbergen et al. (2023). The authors explained it as the different fatigue protocols and other confounding factors. For example, there is a gap in individual differences in responses to running-induced fatigue. Various sports backgrounds, age, and gender might affect the biomechanical changes in running kinematics. Thus, including diverse subjects with varying backgrounds will help to clarify the role of individual variability in fatigue response.

2.2 Comparison of fatigue protocols in running research

Different fatigue protocols such as graded exercise tests and fixed-speed running are utilized in the studies. Koblbauer et al. (2014) and Gao et al. (2022) used the same graded exercise fatigue protocol in their studies. Specifically, all participants started at a speed of 6 km/h and increased their speed by 1 km/h every 2 min on a treadmill until the participant perceived a fatigue level of 13 according to the Borg scale (somewhat hard). The participants ran at this constant velocity until they reached a Borg category rating of 17 (very hard) or 90 percent of their maximum heart rate (HRmax), estimated at 220 - age. Once they got to this point, they continued to run for another 2 minutes before a self-selected cool-down. The maximal stages of exhaustion are also reached using the graded exercise test from Astrand by Radzak et al. (2017).

Another approach to fatigue protocols is fixed-speed running, with the speed being pre-determined by blood lactate values, VO_{2max}, or specific running performances. In the study by Möhler et al. (2021), the running speed to elicit fatigue was pre-determined by an incremental lactate threshold test. Monod and Scherrer (1965) created the concept of critical power, for predicting exhaustion speed. The fatigue speed was 110% of the velocity at four mmol/L Lactate level. And it was defined as the speed which the athlete could run fastest for 10 minutes. The same group of authors, Möhler et al. (2022), applied the testing speed at 13km/h and detected a fatiguing effect in amateur runners. This speed was selected based on previous research with novice runners. During that research, the researchers observed that the speed was appropriate to bring on the exhaustion that occurred after 5-7 minutes in male novice runners.

Except for lactate, Basile et al. (2017) applied VO_{2max} to determine the fatigue speed. They used a pre-determined velocity at 70% of VO_{2max} for 40 minutes to induce fatigue. Riazati et al. (2020) required participants to take an incremental treadmill test to identify their maximum steady state and VO_{2max}. After that, they conducted a high-intensity interval test at a speed that was 1km/h lower than the VO_{2max} speed.

A few studies also included specific running protocols to evaluate fatigue speed during the first visit. To establish their pace for the second session, Maas et al. (2018) required the participants to run a 3200-meter time trail during the first session. According to Willwacher et al. (2020), every participant was required to exert maximum effort during a 10 km treadmill run (equivalent to 105% of their season-best time for the same distance). Luo et al. (2019) required participants to run at a relatively high speed of 3.33 m/s until exhaustion.

Various fatigue protocols including different durations, running speeds, and termination criteria can lead to different internal loads, for example, exercise-induced fatigue. Previous studies suggest that short-term high-intensity exercises predominantly cause peripheral fatigue, while long-term activity tends to induce central fatigue (Perrey et al., 2010; Millet and Lepers, 2004). Thus, various fatigue protocols might stress the subjects differently.

2.3 Quantifying fatigue

Studies frequently use four measures to ensure desired fatigue level: lactate, questionnaires, heart rate, and VO2. Ratings of physical exhaustion (FAS) (Radzak & Stickley, 2020) and ratings of perceived exertion (RPEs), such as the Borg scale (6–20), are both employed in research.

The RPE questionnaire is often used in studies related to running fatigue. An RPE score of 17 (very hard) suggested a fatigue state equal to the long training run (Borgia et al., 2022). In the fatigue protocol that allowed participants to terminate the study at will, participants were asked to reach volitional fatigue or a predetermined RPE value (Borg, 1982). Zandbergen et al. (2023) suggested that using RPE scores as a stopping criterion did not result in significant fatigue effects. In other words, a fatigue protocol stopped via RPE rather than voluntary exhaustion may be insufficient to reach a subject-specific fatigue threshold for kinematics.

Lactate and heart rate are also incorporated into the graded exercise test protocol. Radzak & Stickley (2020) utilized a reference point of 8 mmol blood lactate level to indicate fatigue. Meanwhile, Luo et al. (2019) identify 90% of the maximum heart rate as an indicator of fatigue.

Different methods of assessing fatigue can influence the outcomes. RPE is a subjective perception of fatigue, while heart rate, VO_2, and lactate levels are objective physiological parameters. Although RPE was widely used as a reference for fatigue, a systemic review conducted by Zandbergen et al. (2023) found that five related studies found no significant effects of fatigue. Therefore, using RPE as a stopping criterion might not be sufficient to reach individual fatigue thresholds.

Fatigue is the phenomenon presented as the interaction between performance fatigue and perceived fatigue (Enoka & Duchateau, 2016). This process included both physiological and psychological aspects. Thus, a combination of questionnaires and physiological parameters (heart rate, lactate, etc.) might serve as a better quantification for fatigue.

2.4 Subject characteristics

In running-induced fatigue studies, the characteristics of the participants include various factors such as age, gender, body mass, height, maximal oxygen consumption (VO_{2max}), foot strike pattern, and shoe model. However, subjects are typically selected based on their running experience.

According to Maas et al. (2018), Möhler et al. (2022), and Koblbauer et al. (2014), novice runners can be defined as individuals who have little or no regular running background or have a limited weekly running distance. Maas et al. (2018) defined novice runners as those who do not have a history of competitive running or following a running training program, but they should be able to finish a 3,200 m time trial without pausing or walking. Similarly, Möhler et al. (2022) defined novice runners as individuals who do not engage in regular running activity (running no more than once per month). According to Koblbauer et al. (2014), novice runners were required to run less than 2-3 times per week for less than 10 km and/or 45 minutes per session but had the physical capacity to run at a self-selected pace for approximately 30 minutes and/or 5 km at a time.

Amateur or recreational runners are individuals who run regularly without the competitive or rigorous training schedule of professional athletes. Gao et al. (2022) defined amateur runners as those who run at least two to three times per week for less than 45 minutes or run less than 10 km. It's interesting to note that Koblbauer et al. (2014) define novice runners using similar standards. In contrast, Borgia et al. (2022) define recreational runners as individuals who routinely complete at least 15 miles per week. Luo et al. (2019) recruited fourteen male recreational runners who had been jogging at least 15 km per week for at least three months. Willwacher et al. (2020) included recreational runners who frequently trained and had a season-best time slower than 47:30 minutes for a 10-kilometer run. Mo & Chow (2019) recruited experienced runners who have been running regularly for 4-20 years with a minimum weekly running volume of 30-80 km. Finally, Bazuelo-Ruiz et al. (2018) examined recreational runners with a weekly volume of 20 to 40 km over 2 to 3 days a week.

Expert or competitive runners were defined based on their participation in running events and training. Competitive participants had at least three years of running experience, participation in running events, and an average weekly training capacity of 70 km in male runners and 50 km in female athletes (Maas et al.,2018). According to Möhler et al. (2022), expert runners should have a personal record of 10 km within 35 minutes, with a minimum weekly training mileage of 50 kilometers over the last eight weeks, coupled with a minimum two years running club membership. In addition to that, Willwacher et al. (2020) focused on competitive distance runners who can perform 10 km quicker than 37:30. These studies seem to suggest that an expert or competitive runner should have a significant amount of running experience or frequently competing in races.

Maas et al. (2018) investigated the differences in kinematic between novice and competitive runners. They found that novice runners increased forward trunk lean and hip abduction during mid-swing, while competitive runners demonstrated a decrease in hip abduction. This indicates that the effects of fatigue observed in experienced runners cannot be applied to novice runners. Increased forward trunk lean with fatigue in novice runners has been supported by previous studies (Koblbauer et al., 2014). This may be attributed to local fatigue of the trunk musculature or fatigue in the knee extensors. Forward trunk lean during running leads to an anterior displacement of the center of mass, which has been associated with a greater hip extensor moment and a lower knee extensor moment. The increase in hip abduction, ankle plantar flexion, and a decrease in knee flexion may be a compensation strategy to keep the foot off the ground while swinging the leg forward. These small but systematic changes in running kinematics could be explained by fatigue in the trunk, hip, and thigh musculature (Maas et al., 2018). It is also worth noting that the effects of fatigue are induced by a low to moderate intensity run.

Apte et al. (2021) found that runners with different skill levels could lead to confounding effects. Therefore, it is crucial to investigate the impact of experience level on kinematic changes with fatigue. Based on the review conducted by Zandbergen et al. (2023), the main kinematic changes resulting from fatigue include an increase in peak accelerations at the tibia, decreased leg stiffness, an increase in knee flexion at initial contact and maximum knee flexion, and an increase in vertical center of mass displacement (ΔCOMz) in novice runners. Differences in

ΔCOMz responses to fatigue may be because experienced runners adopted a more energy-efficient gait pattern resulting in smaller ΔCOMz. In contrast, novice runners may experience a larger ΔCOMz due to an increase in knee flexion resulting from more significant loss of knee extensor strength with fatigue.

Different lower-limb kinematic waveforms and coefficients of variation have been observed in runners with varying training volumes and athletes with different practice skill levels (Mo & Chow 2019). They found that expert and novice runners adapted to progressively induced fatigue via different coordination patterns of their lower limbs. During midstance, experienced runners showed more variability in hip-knee and shank-knee couplings, while novice runners displayed more variability in hip, knee, and thigh motion. Moreover, experienced runners showed more variability in their coordination, whereas novice runners exhibited more variability in their joints and body segments during long-term running. These differences in coordination and variability may contribute to improved running performance and lower injury risk in experienced runners.

2.5 Running environments

Testing environments are various through studies. Treadmills are commonly used for biomechanical evaluation. Different treadmill models are used in the studies (Luo et al., 2019; Möhler et al., 2022; Gao et al., 2022), including treadmills instrumented with force platforms or synchronized with motion capture systems for data collection. In most treadmill protocols, fatigue was commonly induced by using different speeds or intensities. Another important variable is inclinations, some studies had the inclinations set at 1%, while some applied a 0% inclination. Möhler et al. (2022) suggested that a 1% inclination can better simulate the energy demands and physiological responses of overground running. In contrast, Abt et al. (2011) set the treadmill inclination at 0% during the data collection procedures. However, it also should be acknowledged that some other studies conducted their experiments in overground running environments. For example, Maas et al. (2018) carried a 3,200 m time trial on a 400m running track. Borgia et al. (2022) used a 10-m runway with instrumented force platforms in a lab. Usually in overground running trials, the participants were allowed to run at self-selected or comfortable speeds. Kinematic data using optical motion capture systems were collected. Some researchers combined both the treadmill and overground running. Specifically, Gao et al. (2022) conducted fatigue protocol through the treadmill and the motion analysis data through overground running. Similarly, a treadmill was used for the fatigue protocol and an 18 m runway was utilized for kinematic data (Radzak & Stickley, 2020).

Riley et al., (2008) conducted a study to compare both kinematic and kinetic parameters between overground running and treadmill running. Motion capture data and ground reaction force were recorded through 15 consecutive gait cycles. While kinematic and kinetic trajectory features were similar between treadmill running and overground running, there were significant differences in knee kinematics and peak values of GRF. It indicated that the instrumented treadmill and overground running are comparable in kinematic and kinetic parameters but not the same. The authors also suggest that the surface of the treadmill should be stiff while belt speed is controlled adequately. Thus, it is essential to differentiate the treadmill and overground running, since the biomechanical responses may be different between the two environments (Van Hooren et al., 2020).

3 INTER-LIMB ASYMMETRIES

3.1 Factors affecting interlimb asymmetry

Interlimb asymmetry is defined as the difference in strength, function, and physical capacity between the limbs (Heil et al., 2020). Different sports characteristics might result in interlimb asymmetries. Ball sports inherently include many asymmetrical movements, ie. Football, baseball, and golf. Football players are required to complete high-intensity, intermittent and multidirectional movements such as jogging, sprinting, jumping, and changing direction in unpredictable competition environments (Ascensão et al., 2008). Players are exposed to high levels of unilateral force production during those movements. According to Mohr et al. (2003), those movements happened and changed approximately every five seconds in 1300 players. Asymmetry becomes common when players continue to perform these maneuvers during competition and react to unpredictable stimuli from opponents or ball trajectory (Hart et al., 2016). Furthermore, interlimb asymmetry also occurs when the non-kicking leg stabilizes the body and absorbs ground reaction forces during kicking (Bromley et al., 2021).

It is interesting to note that asymmetries can occur even in symmetric sports like running (Heil et al., 2020). Running is one of the common symmetric sports (swimming, cycling) with continuous and cyclic movement patterns. In running, lower limb dominance can be categorized according to the function of stabilization or mobilization. However, there is no agreement about the standard for defining lower limb dominants since both stabilization and mobilization would require finer neuromuscular control. In healthy populations, gait is generally considered symmetrical, however, observed discrepancies between limbs during walking suggest that one limb potentially favors braking and the other propulsion (Radzak et al., 2017). Research has indicated that the discrepancies in propulsion increase as the walking speed increase, suggesting a potential amplification of asymmetry in running (Seeley et al., 2008). In running, one side can compensate more for the other side which might lead to interlimb asymmetries (Levine et al., 2012). Hence, the lower limb asymmetry quantification is essential in gait analysis. The increased risk of non-contact injury of one lower limb is due to increased exposure to greater amounts of load (Paterno et al., 2010). If chronic use leads to the use of one limb more than the other, there is a resultant difference in strength, flexibility, range of motion, and neural development (Parrington & Ball, 2016). According to Barber et al. (1990), it is reported that a higher symmetry score (>15%) is associated with a higher injury rate compared to those with lower symmetry score (<15%).

Except for the sports background, various factors such as sex, injury history, fitness level, or anthropometric factors can also affect interlimb asymmetries (Heil et al., 2020). Sarabon et al. (2020) investigated how various strength outcomes, motor tasks and muscle groups influence interlimb asymmetry in professional and semi-professional athletes. The outcome was both unilateral and bilateral maximal voluntary contractions of the knee extensors and flexors. It was found that values of rate of torque development mostly showed greater bilateral deficits and interlimb asymmetries than those of MVC torque, mainly for the knee extensors and in unilateral tasks. Thus, valid assessment needs to be used carefully when measuring interlimb asymmetries.

In addition, Bredeweg et al. (2013) reported that after 9 weeks of training, a group of novice runners who sustained an injury showed increased asymmetry in ground contact time and decreased asymmetry in the impact force. However, after controlling for the number of BMI, age, sex, leg length, and training group, there were no relationship between interlimb asymmetry and injury.

Running speed has an effect on running mechanics and lower limb asymmetry (Wayner et al., 2023). Yet there is no consensus on the extent to which speed affects symmetry. For example, Stiffler-Joachim et al. (2021) examined lower extremity kinematic and kinetic running in healthy, Division-I collegiate athletes from football, basketball, soccer, track (mid-distance events), and cross-country. They included athletes who maintained different running speeds. The results indicate that asymmetries of joint kinematics are less than 3°, whereas joint power asymmetries are between 10% to 40%. However, there are minimal effects of gender and speed on interlimb asymmetry. In addition, Furlong and Egginton (2018) studied the effect of running speed on kinetic asymmetry. The speed was altered by ±20% of their preferred running speed. They found that the speed change had little effect on the asymmetry across joints, ranging between ±6%. Increasing running speed increases Achilles tendon stress but it is independent of ground reaction forces. Overall, running speed changes do not significantly affect interlimb asymmetry. Individuals respond differently in order to maintain interlimb symmetry.

3.2 Outcome measures for inter-limb asymmetry and sports injury

Interlimb-asymmetry may potentially increase the sports injury risk for both legs. The weaker side is not able to withstand the average load while the stronger side experiences excessive stress. Furthermore, inter-limb asymmetry may lead to unequal force absorption or frontal plane stability loss, which consequently results in sport-related injury (Guan et al., 2022). However, outcome measurements for inter-limb asymmetries are various in the publications. Low-limb strength tests including peak force in isometric/isokinetic strength tests are the most common. Steidl-Müller et al. (2018) recruited 285 high-level ski racers at 3 different age levels, aiming to investigate the limb symmetry index in strength and coordinate tasks. In youth ski racers, the LSI for one leg isometric/isokinetic press strength test was a significant risk factor for traumatic injury. This finding suggests the importance of an asymmetry monitoring system for youth athletes in order to prevent injury.

Jump tests including single leg hop and countermovement jump are also widely used in publications to examine lower-limb symmetry. According to previous studies, inter-limb asymmetry of more than 15% is associated with higher sports injury risk (Guan et al., 2022) Fort-Vanmeerhaeghe et al. (2022) observed there is a correlation between asymmetrical single leg countermovement jump height and non-contact lower-limb injuries in young healthy athletes. In particular, the average asymmetry for men and women among non-injured athletes was 9.7 ±8.3 and 7.7 ±5.6%, respectively. In contrast, values for men and women who were injured were 17.1 ±13.3 and 12.8 ±6.2%, respectively. The result is in alignment with the 15% threshold we discussed before. To further explain the observations, the authors suggested that the weaker limb is less capable of producing and absorbing force. During repeated high intensity activity, this deficiency likely predisposes the limb to injury since the weaker limb will surpass the "tolerance capacity" sooner compared to the stronger limb.

Smith et al. (2015) investigated 184 Division I athletes from eight different sports to explore the relationship between asymmetry and non-contact injury risk using Y Balance Test (YBT). YBT is a measurement based on the Star Excursion Balance Test, which measures dynamic balance in three different directions including anterior, posteromedial, and posterolateral directions. Asymmetry is defined as the distance difference between limbs in the anterior, posteromedial and posterolateral directions and the average reach distance normalized to the leg length (composite score). The results indicated that anterior asymmetry > 4 cm was significantly associated with noncontact injury risk.

YBT asymmetry has also been used to predict patellofemoral pain in male military recruits. the patellofemoral pain group had a more significant asymmetry than the controls in the posterolateral direction of the YBT. Furthermore, Nakagawa et al. (2020) have proposed that a mean YBT posterolateral asymmetry $\geqslant$ 4.08 cm was significantly associated with patellofemoral pain. Such activities as squatting, stair climbing, and running uphill/downhill increase the compressive force on such knee joints, thus increasing knee pain (Crossley et al., 2007). Based on these findings, it is important to know how running could contribute to gait asymmetry that predisposes runners to patellofemoral pain.

Plisky et al. (2006) carried out research among high school basketball players to establish whether the Star Excursion Balance Test could predict injury risk to the lower extremities. The athletes whose anterior reach difference between limbs was more than 4 cm are 2.5 times more likely to have an injury of lower extremity. In addition, the female players who reached less than 94% of the length of their limb with their composite reach were 6.5 times more likely to incur a lower limb injury. Also, Ruffe et. al (2019) found that male high school cross-country runners with a Lower-Quarter YBT posteromedial reach difference of ≥ 4.0 cm were five times more likely to incur a running-related injury. It is worth noting that basketball players had higher scores compared to the high school cross-country runners. It might be due to the weaker proximal muscles of runners while YBT is a measurement that requires proximal control and stability. This suggests that the YBT may have a sport-specific component. Sports emphasizing quadriceps and gluteal development are likely to result in higher YBT scores, since these muscles' movements are similar to those required during the test (Ruffe et al., 2019).

Among all outcome measurements, jumping test, YBT, and strength task contain different movement patterns, muscle contraction types and speeds. According to the specificity of training effects, the training exercises should mimic the sport task parameters in order to induce adaptations that may be transferred to the sport-specific movement (Duchateau & Baudry, 2010). The same may apply to the inter-limb asymmetry evaluation, athletes who have practiced the maneuver for a long period of time will receive higher scores. Therefore, care should be taken when selecting the outcome measurements to avoid movements that are unfamiliar to the athletes. Assessing inter-limb asymmetry during training or competition may also be an option.

3.3 The influence of exercise-induced fatigue on inter-limb asymmetries

Injuries in sports frequently occur towards the end of competitions. For example, 70.8% of injuries among amateur rugby players happened after halftime of the games (Gabbett, 2000). This phenomenon suggests that fatigue may be one of the risk factors for sports injury. According to Heil et al. (2020), fatigue induced by the physical loading of sports increases the non-contact injuries risks. This reduction in neuromuscular control after fatigue has been identified as a potential mechanism contributing to the high incidence of sports injury (Bishop et al., 2021). Specifically, such fatigue alters the kinematics which leads to high impact accelerations and further increases the injury risk (Leister et al., 2017).

Moreover, exercise-induced fatigue can lead to physiological alterations such as a reduction in muscle voluntary activation, activation pattern, and proprioception (Heil et al., 2020; Bell et al., 2016). Exercise-induced fatigue can reveal or exacerbate the preexisting asymmetries between limbs due to physiological changes. The detrimental alterations affect the execution of movements, potentially increasing the difference in movement parameters and thereby aggravating inter-limb asymmetries (Heil et al., 2020). Beck et al. (2018) found that asymmetry in running can impact running quality. Foot contact time asymmetry and mean ground reaction force asymmetry increase leads to higher metabolic costs. Therefore, evaluating exercise-induced fatigue on inter-limb asymmetry is essential to understand the potential mechanism behind non-contact injuries (Radzak et.al., 2017).

In running research, measurements of inter-limb asymmetry can be used for both injury risk prediction and quantification of low limb functional deficits. However, it is worth noting that not all differences between the spatiotemporal parameters necessarily reflect the inter-limb asymmetry (Hanley & Tucker, 2018). Hence, other outcomes such as kinetic or kinematic measurements should be considered as inter-limb asymmetry measurements. For example, asymmetries in vertical ground reaction forces can predict kinetic asymmetries since it is associated to the muscle strength and power. Assessment of landing force asymmetry can therefore provide valuable insights into muscle function and injury prediction (Bell et al., 2016).

Girard et al., (2017) investigated thirteen uninjured male athletes and quantified asymmetry using the Bilateral Leg Asymmetry percentage during sprinting repetition. This research suggested kinetics, kinematics, and vertical stiffness alterations after neuromuscular fatigue.

However, there was no significant interaction between the sprint repetitions and dominance for all parameters which indicates the dominant and non-dominant limbs may exhibit similar fatigue rates during treadmill running. Sprinting is inherently a short-duration activity; therefore, extending the duration, coupled with either shorter recovery periods or more intense loading patterns, might lead to more pronounced adverse effects of fatigue on dominance.

According to Ferris et al. (1998), stiffness is one of the important measurements related to the evaluation of lower limb neuromuscular modulation. Consistent with the spring-mass model, the leg acts similarly to a spring that absorbs shock during braking and releases energy during propulsion. The nervous system is able to maintain the desired center of mass on different running surface by adjusting the stiffness of the leg. In addition, research has shown that joint stiffness is associated with overuse injury because a more compliant joint reduces joint load more effectively compared to a stiffer joint (Hamill et al., 2009). Conversely, higher joint stiffness may help to stabilize limb movement and prevent joint instability (Brughelli & Cronin, 2008). The stiffness parameters were calculated from kinematic data (Morin et al., 2005). This method calculates leg and vertical stiffness using just a few fundamental biomechanics factors: body mass, forward velocity, leg length, flight time, and contact time. Möhler et al. (2021) studied vertical and leg stiffness in expert runners during middle-distance runs. The results indicated that after the fatigue protocol, both vertical and leg stiffness significantly decreased with a high effect size. The reduced stiffness made contact times longer and flight times shorter, which the authors explained as reduced running performance caused by the lower effectiveness of the stretch-shortening cycle that might potentially increase energy costs. Thus, stiffness is a key factor in stability and performance in sports.

Razak et al. (2017) studied the knee stiffness of 20 physically active individuals. It reveals a significant difference in vertical stiffness during running. knee internal rotation and knee stiffness became more asymmetrical with an increase of 14% and 5.3 % after fatigue. These results are supported by Gao et al. (2022), the knee stiffness SA in the coronal plane increased significantly by 13% after fatigue. Additionally, an increase in SA for the hip joint stiffness after fatigue suggests the absence of neuromotor activation from the central nervous system and a decrease in bilateral symmetry of hip muscle activity.

Kinetic asymmetry has been recognized as a potential risk factor for injury since it might expose bones to high-loading stress. Ali et al. (2016) conducted research on impact acceleration as a kinetic parameter due to its significant correlation with vertical ground reaction force, localized muscle fatigue, and the development of fatigue development. In general, there was a non-significant relationship between kinetic asymmetry and the metabolic stress parameters of blood lactate and RPE. In light of this fact, the author notes that kinetic asymmetry could not be used for predicting fatigue purposes for any of the study participants.

Surprisingly, Vial et al. (2023) found that kinematic parameters after fatigue running showed more symmetrical. Facts may suggest a protective strategy in which the loads are distributed more evenly between the limbs as fatigue progresses. Alternatively, after fatigue, the increase in workload for the nondominant leg may also increase the injury risk. In this case, the lower asymmetry may indicate that the non-dominant leg is at a further increased risk of injury, given that it bears a higher load. In contrast, male amateur runners showed increased SA of knee extension angles, knee abduction moment, and hip joint flexion moment after fatigue (Gao et al., 2022).

3.4 Inter-limb asymmetry indices

3.4.1 Symmetry angle (SA)

Steinmetzer et al. (2022) suggest there are two main areas in gait symmetry assessment, including discrete-based and time series-based. Usually, wearable devices are used in discrete-based assessment. The most common measures utilized in discrete-based methods include RI, SI, GA, and SA. In this case, the symmetry is derived from spatiotemporal parameters that include the stride length, duration, and duration of stance and swing phase.

Among these approaches, SA is a measurement that represents inter-limb asymmetries. It can also be used to assess the differences between limbs in various gait parameters. SA is a reliable tool for measuring spatiotemporal and kinetic variables because it is a dimensionless metric without a reference value. Also, SA is not subject to artificial inflation due to changes in the reference value (Hanley & Tucker, 2018). Moreover, SA is highly correlated with the symmetry index which suggests that SA is an effective replacement for the symmetry index (Błażkiewicz et al., 2014).

According to Zifchock et al. (2008), the asymmetry between left and right discrete values is often measured in the SA. Constructing a vector in the coordinate system, left and right side values build a vector that would result in producing an angle. Any angle deviating from 45° represents a certain degree of asymmetry (Figure 2).

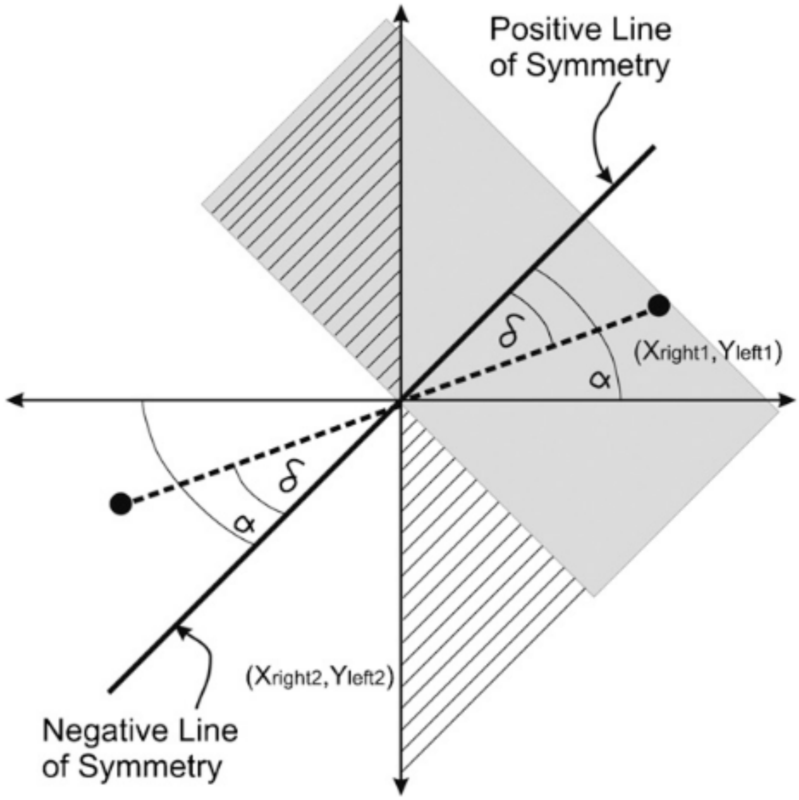

FIGURE 2. Quantification of the SA (Zifchock et al., 2008)

The formula for SA is as follows, X_{left} and X_{right} represent for the left and right side values, respectively:

$$SA = \frac{(45° - \arctan(\frac{Xleft}{Xright}))}{90°} \times 100\%$$

0% SA value means perfect symmetry, while 100% indicates equal but opposite direction values, which is completely asymmetry. If [45°-arctan (X_{left}/X_{right})] exceeds 90°, the subsequent formula needs to be replaced:

$$SA = \frac{(45° - \arctan\left(\frac{Xleft}{Xright}\right)) - 180°}{90°} \times 100\%$$

Gao et al. (2022) conducted a study to examine lower extremity SA among amateur runners. The study reported that, there was an increase in symmetry in internal rotation of the joints of the ankle, knee, and hip after fatigue; in contrast, the angle for the symmetry of external rotation of these joints decreased significantly. This change could be attributed to the body adapt a coordination mechanism in order to maintain the symmetry and stability of the overall lower limbs after fatigue. Also, Radzak et al. (2017) noted that variations in SA during rested and fatigued states for several gait variables. However, some variables remained asymmetrical both before and after the fatiguing protocol.

3.4.2 Symmetry index (SI)

SI was proposed by Robinson et al. (1987) to quantify the gait asymmetries. SI is also the most commonly used method in research about gait asymmetry:

$$SI=\frac{X_R-X_L}{0.5\times(X_R+X_L)}\times 100\%,$$

Where X_R represents for the right side and X_L represents for the left side. SI=0 indicates there is a perfect symmetry between left and right side while a positive value suggests that the X_R is higher than the X_L; a negative value indicates X_L is larger than the X_R.

Błażkiewicz et al. (2014) compared the four common approaches for quantifying gait asymmetry. Their results indicated there is a perfect agreement between SI and Ratio Index (RI) (Figure 3), leading them to propose that SI presents the most sensitive measurement of gait asymmetry based on spatial-temporal parameters in healthy subjects.

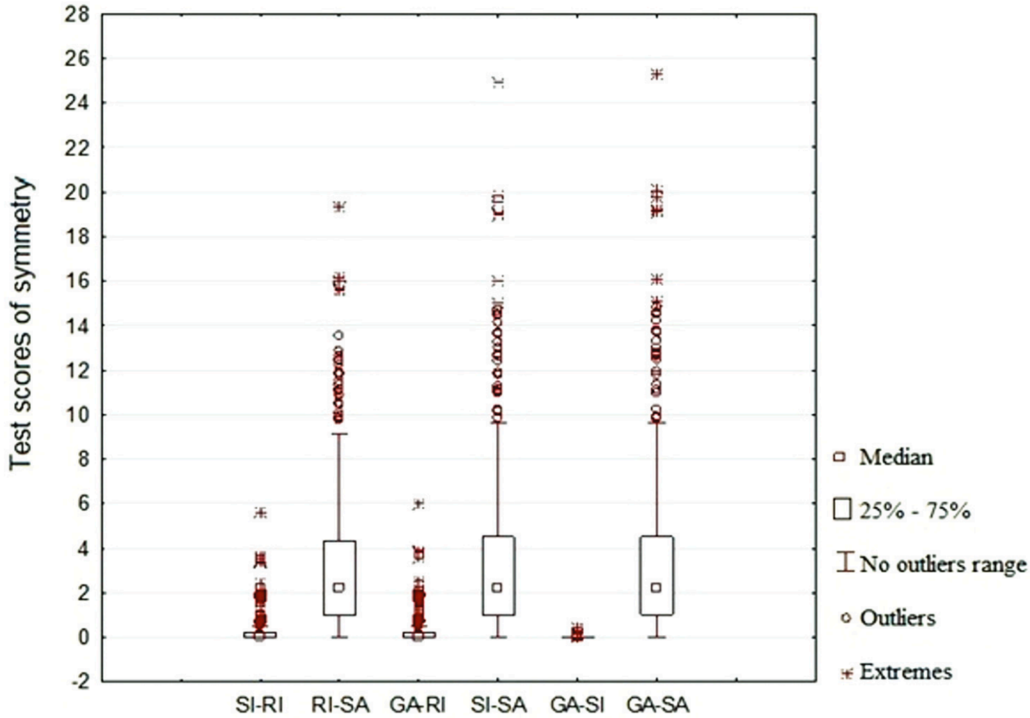

FIGURE 3. Representation of RI, SI, SA, and GA symmetry indicators featuring outliers and extreme data points (Błażkiewicz et al., 2014).

However, one of the limitations of SI is artificial inflation. According to Herzog et al. (1989), the upper and lower limits of SI vary from ±4% to over ±13000%. Calculation of SI requires a referenced number, the SI inflates when the reference number is close to zero. Also, it is difficult to choose a reference value for a healthy population since there is no injury side. Usually, the average for two sides is adopted as the reference value. Further, the SI lacks the information of the whole gait cycle (Shorter et al., 2008).

3.4.3 Gait asymmetry (GA)

GA (Plotnik et al., 2005) is calculated as a simply logarithmic transform of the RI:

$$GA=\ln(\frac{X_R}{X_L}) \times 100\% ,$$

Where X_R represents for the right side and X_L represents for the left side. The value of 0% indicates perfect symmetry while GA>100% suggests asymmetry. The value can be more than 100% and there is no upper limit. Native value is possible when $X_R < X_L$. It is worth to note that only when the ratio between $X_R < X_L$ is positive, the GA can be defined. When there are both positive and native values exist, it is not applicable to measure GA (Queen et al., 2020). Therefore, GA is not recommended as a standard symmetry parameter.

Plotnik et al. (2007) utilized the GA to study the gait of patients with Parkinson's disease based on the swing phase duration. Parkinson's disease has significantly higher GA compared to the elderly subjects. GA of healthy elderly subjects is higher than the young adults. The author did not provide any other information about the validity of GA.

Queen et al. (2020) tested the performance of four symmetry indices in ACL reconstruction patients. They suggested that RI and GA are unstable and can overestimate the value in some cases.

3.4.4 Ratio index (RI)

The Ratio Index (RI) is widely used to analyze gait asymmetry:

$$\mathrm{RI} = (1-\frac{X_R}{X_L}) \times 100\%,$$

Where X_R represents the right side and X_L represents the left side. RI=0 suggests perfect symmetry, while RI $\geqslant$ 100% indicates asymmetry. It is possible that RI is a negative value, and RI can exceed 100% with no upper limit (Queen et al., 2020).

The main disadvantages of RI are the low sensitivity, low asymmetry detection, and RI is unable to pinpoint the precise location of the asymmetry. Nevertheless, it is important to note that RI represents a reciprocal gait pattern although it is relatively straightforward. Both higher and lower values indicate asymmetries. In the review of the comparison of different symmetry indices, RI obtained from spatiotemporal parameters is in between the SA and SI (Błażkiewicz et al., 2014).

3.4.5 Statistical parametric mapping (SPM)

Nowadays, it is common to apply both discrete and continuous analysis to examine kinematics and spatiotemporal characteristics. Discrete analysis involves analyzing data at specific points or intervals in time. For example, in the analysis of gait, discrete analysis may involve identifying the times at which the foot strikes the ground and analyzing the joint angles and forces at those specific points in time. Discrete analysis can provide precise information about specific events and allow for comparisons between different time points but may miss important information about the overall movement or signal dynamics. Limb asymmetry has also been evaluated by all kinds of discrete indices, such as SA, GA, RI, SI. Those discrete asymmetry indices only consider a single time point of parameter, without taking into account both the magnitude and shape of a waveform (Vial et al., 2023).

Continuous analysis refers to a set of techniques for analyzing the entire continuum. Most continuous techniques do not discard data. Continuous analysis can capture the overall movement or signal dynamics but may require assumptions about the functional form of the data and may not provide precise information about specific events. Functional data analysis, dimensionality reduction, machine learning, and statistical parametric mapping (SPM) are all examples of continuous analysis. SPM can be used to examine time series data of joints because it is able to discover field regions that are significantly co-vary with the experimental design, making it superior to oversimplified discrete parameter studies (Pataky et al., 2013). SPM offers the benefits of considering the signal as a whole and presenting the findings in the original sampling space. For example, Vial et al. (2023) compared both kinematics and ground reaction force between limbs before and during a running-induced fatigue protocol. Surprisingly, the SPM test showed contradicted results to their hypotheses, the inter-limb asymmetry was higher before fatigue than during fatigue running. It is worth noting that Vial only included thirteen male semi-professional soccer players in their study. The sample size is relatively small. Also, their fatigue protocol consisted of three maximal 50-m sprinting, which might not be sufficient to induce an intense fatigue response.

In summary, SPM assumes the adjacent data points are related which is essential to the analysis of biomechanical time series data. Thus, SPM may serve as a valid method in the biomechanical field.

4 OPTICAL MOTION CAPTURE SYSTEM

Optical motion capture systems (OMC) are commonly used to capture kinematic data during running biomechanics research. These systems typically use multiple cameras to track markers placed on the participant's body, allowing for the measurement of joint angles and other biomechanical variables. Vicon (Vicon, Oxford, UK) is a motion capture technology that allows automated tracking and labeling. Their tracker is high-precision, approximately 0.017 mm for each marker. Vicon is frequently employed in gait analysis and is regarded as the "gold standard." Due to its great accuracy and reliability, Vicon has been one of these systems that is frequently utilized in the literature, with camera counts ranging from six to sixteen. The kinematic data could be collected at three different sampling rates in the Vicon: 100 Hz for the Maas et al. (2018) study, 500 Hz for the Riazati et al. (2020) study, and 1000 Hz for the Willwacher et al. (2020) study. The sampling rate depends on the research question and the level of detail necessary for the accurate capture of movement. Other OMCs being used in the studies include Qualysis (Qualysis, Inc., Sweden) (Mo & Chow 2019), Optotrak (Optotrak Certus®, Northern Digital Inc., Waterloo, Ontario) (Koblbauer et al., 2014), and Motion Analysis Corporation systems system (Motion Analysis Corporation; Santa Rosa, CA) (Brown et al., 2014).

Retroreflective markers are often placed on anatomical landmarks on the trunk, pelvis, and lower extremities to track motion in three dimensions, albeit the placement of markers varies between researches. Most studies used at least 20 markers (Mo & Chow 2019) in their marker sets, however, some used as many as 78 (Willwacher et al., 2020). The placement of markers was frequently bilateral (ie. the Plug-in gait model, Oxford foot model) and included landmarks such as the anterior and posterior superior iliac spines, the thigh, the lateral epicondyles of the femur, the lateral shank, the lateral malleoli, the base of the second metatarsal, and the calcaneus (Figure 4). Some investigations employed cluster markers on the thighs and shanks, and extra markers were put on shoes to track motion.

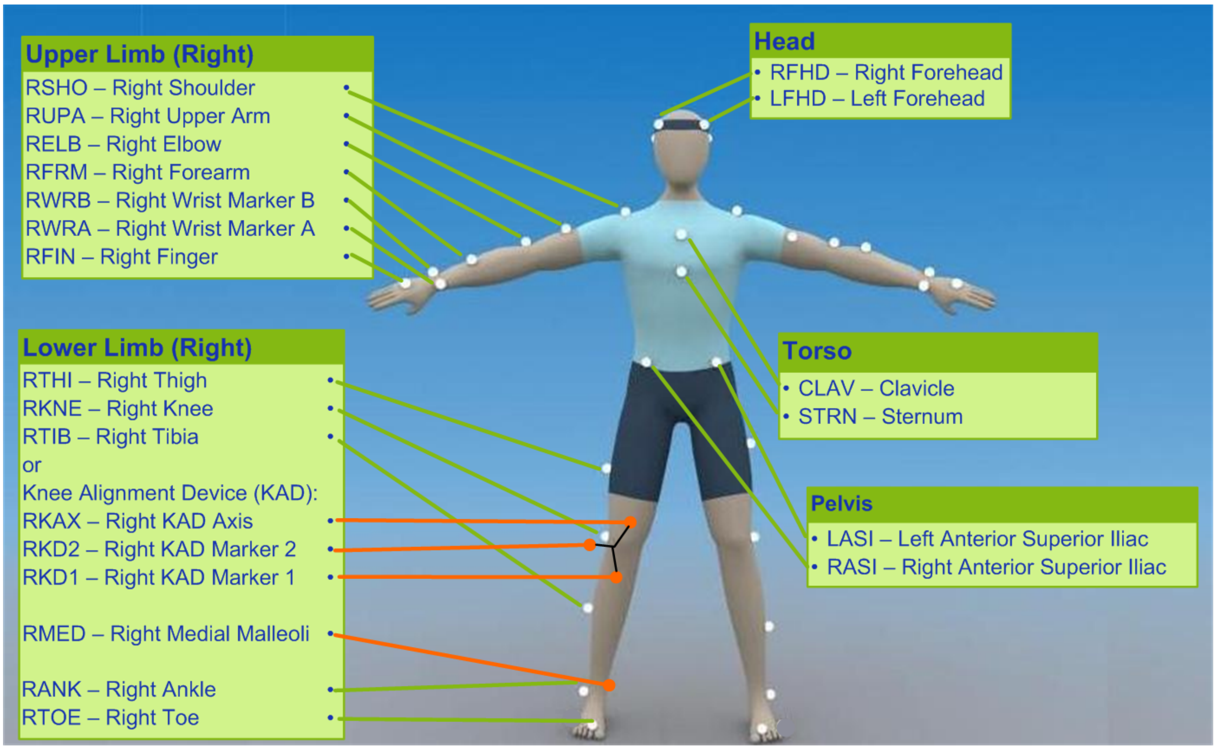

FIGURE 4. Plug-in-gait full body model (Vicon Documentation)

The low-pass Butterworth filter is the most commonly used filter in raw data analysis. This filter is widely applied in various research works for filtering kinematic and kinetic data related to human motion analysis (Luo et al., 2019; Chen et al., 2022; Möhler et al., 2022; Riazati et al., 2020; Willwacher et al., 2020; Radzak & Stickley 2020; Maas et al., 2018; Borgia et al., 2022; Dierks et al., 2010; Mo & Chow 2019; Koblbauer et al., 2014; Encarnación-Martínez et al., 2021; Abt et al., 2011; Brown et al., 2014). The order and cutoff frequency of the filter are usually user-defined and can be experiment-specific. A fourth-order low-pass Butterworth filter is most common. The cut-off frequencies range from 6 Hz to 20 Hz, depending on the specific application or data type.

In order to calculate kinematic parameters from motion capture data, a biomechanical model with individual anthropometry measurements is first created. The model is then fitted to the 3D marker trajectories using inverse kinematics algorithms or optimization techniques (least-squares optimization) by OpenSim or Vicon Nexus. Kinematic parameters such as joint angles, segment orientations, and velocities are derived from the model after fitting the model to the data. After that, the future analysis of the subject's motion and biomechanical experiment can be continued.

5 BASIC ABOUT OPENSIM

5.1 Direct kinematics (DK) vs. Inverse kinematics (IK)

Direct kinematics (DK) has been used in most conventional gait model analyses, while Inverse Kinematics (IK) has gained popularity in recent years due to its characteristics. Vicon Nexus utilizes the Plug-in-Gait (PiG) model to calculate joint kinematics by DK. While Musculoskeletal models, such as OpenSim (Delp et al., 2007) and AnyBody (Damsgaard et al., 2006) employ IK for joint angle calculation. DK assumes the experimental markers are rigidly attached to the bones and body segments. Joint angles are calculated directly from the 3D marker's position by Cardan angles based on the relationship between adjacent segment reference frames (Grood & Suntay, 1983). There are some limitations when fully relying on the marker position to calculate kinematics. The data can be invalid when markers are invisible by obstacles. The marker placement requires high accuracy otherwise the misplacements could lead to the wrong definition of the plane (Szczerbik & Kalinowska, 2011). Moreover, the length of the segment was changeable and varied up to 2 cm during walking (Baker et al., 2017). Also, DK gait models are limited to joint kinematics and kinetics (Horsak et al., 2018). In contrast, IK (or global optimization) employs a skeletal-joint model with rigidly attached markers, calculating joint kinematics by adjusting models (Kainz et al., 2016). When the optimal match between the model and experimental marker position is achieved, the marker location error will be accepted (Andersen et al., 2009). The estimated joint angles are considered reliable when the root-mean-squared marker error is less than 1 cm during the whole trial for most of the models and studies (Hicks et al., 2015). Lu & O'Connor (1999) pointed out that when compared to DK, static optimization (IK) is less sensitive to noise when calculating joint kinematics. They also found that the IK results were closet to the real values. Ziziene et al. (2022) compared the reliability between IK and DK in obese children and found no significant difference in joint kinematics.

Compared to DK, IK allows for providing additional kinetic information such as joint reaction forces, joint moments, muscle force, and muscle activation. This analysis can further identify the potential mechanism and provide strategies for people with movement disorders or obese individuals (Horsak et al., 2018). However, the kinetic parameters derived from Inverse Dynamics (ID) can only be achieved after obtaining kinematic parameters from IK (Figure 5). Thus, the accuracy of ID solely depends on the IK process (Ziziene et al., 2022). Therefore, while the DK may be sufficient for calculating kinematic parameters, IK is essential for acquiring extra kinetic information which is crucial for clinical procedures for patients (Ziziene et al., 2022).

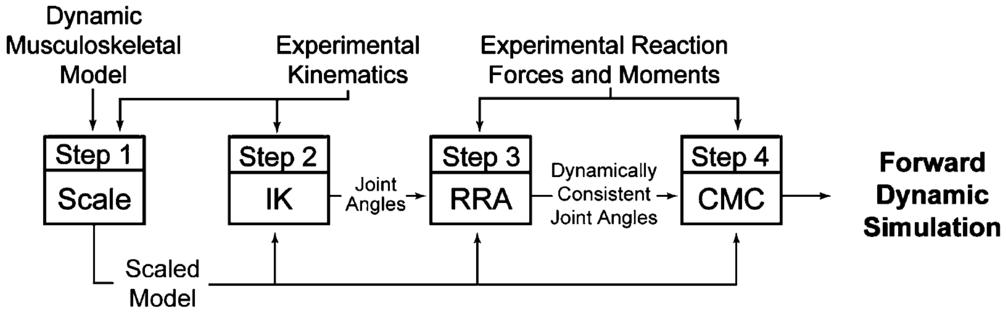

FIGURE 5. Flowchart for creating a muscle-driven simulation using OpenSim, RRA (residual reduction algorithm), CMC (computed muscle control) (Delp et al., 2007).

OpenSim is an open-source simulation software allowed to simulate dynamic movement, which is created by Delp et al. (2007). OpenSim gait2392 model (Delp et al., 1990) contains three degrees of freedom (DoF) at the hip, one DoF at the knee, and two rotational DoFs at the talocrural and subtalar joints. Roelker et al. (2017) compared the difference between four models on kinematics and kinetics. The results show the difference in coordination system between models changed joint kinematics consequently affecting the kinetic parameters (muscle moments, force, and activation). They conclude that the gait2392 is a sufficient model for studying walking in healthy young adults. The study conducted by Falisse et al. (2018) studied Human Body Model and the OpenSim gait2392 model. After comparing the kinematics and kinetic data, they found significant differences between joint kinematics, kinetic, and muscle forces. It can be explained by the hip and knee joint center location differences and the offset in pelvic reference frames. Specifically, OpenSim defines a neutral position as 0 degrees which is aligned with the anatomical position. The other model, for example, PiG defined the neutral position by the anterior and posterior iliac spine markers which lead to an offset of around 13% (Kainz et al., 2016).

Overall, the potential differences between Vicon Nexus and OpenSim can be attributed to the different computational methods (IK vs. DK), anatomical models (PiG vs. gait2392), and marker sets. Where different anatomical models include the difference in anatomical segment frames and joint constraints (rotational and/or translational DoF) (Kainz et al., 2016).

Constraints refer to a reduction in the six DoF of a segment. A ball-and-socket joint model is a pure rotation model since the physiological joint translation of the hip and ankle joint is less than 2 mm (de Asla et al., 2006). Hence, in a typical biomechanics model, the constraint joint is reduced to three DoF when it constrains translational movement between segments. Yamaguchi and Zajac (1989) developed a one DoF knee joint model when considering the movement of the tibiofemoral joint and the patellofemoral joint in the sagittal plane. Delp et al. (1990) used this planar knee model, defining how the femur, tibia, and patella move in relation to each other depending on the knee angle (Figure 6). Thus, the knee joint is analyzed as a hinged joint during gait2392 model. Joint constraint increases analysis reliability since it not only decreases the artifact of soft tissue but also reduces marker placement error (Flux et al., 2020).

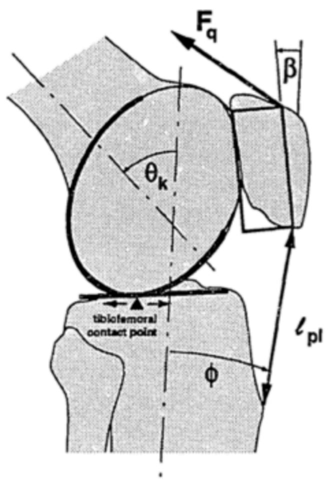

FIGURE 6. Geometry used to calculate knee moments and kinematics in the sagittal plane according to the Delp model (Delp et al., 1990).

5.2 Marker-based approach for scaling

Regardless of IK or DK methods, it is crucial to place markers accurately and reliably on anatomical landmarks. In IK method, the pose and marker position are iteratively adjusted until the generic models align best with the experimental markers. The musculoskeletal model can be generated from several scaling methods, including linear scaling, functional scaling, and statistical shape scaling (Kainz et al., 2017). Linear scaling method is commonly used to create a model based on participants' anthropometry since it is less time-consuming. Linear scaling adjusts the size of the generic model to match participants' anthropometry measurements by using the ratio between individual body segment size and the model (Delp et al., 2007). Generic model can be scaled using the surface markers placed on the anatomical landmarks or together with the virtual joint center markers (Hicks et al., 2015). The quality of results highly depends on the scaling process and experience of the examiner (Horsak et al., 2018).

The hip joint center is a crucial anatomical landmark for establishing local coordinate system of the thigh segment (Assi et al., 2016). However, the hip joint center cannot be directly obtained from the skin surface, a virtual marker is typically used to estimate the hip joint center (Kainz et al., 2017). There are two methods for predicting hip joint center, including predictive methods based on regression equations (Harrington et al., 2007) and functional methods (Pizaaz et al., 2004) based on dynamic calibration during measurements. Harrington regression equation results from regression analysis and leave-one-out cross-validation on the magnetic resonance imaging (MRI). The equation uses pelvic depth (PD) to predict anterior-posterior direction, pelvic width (PD), and leg length (LL) for the superior-inferior direction, and PD and PW for the mediolateral direction. Hence, they proposed a single linear regression equation (in mm) for the right hip considering the practicalities and the reliability:

$$\hat{X} = -0.24PD - 9.9$$
$$\hat{Y} = -0.30PW - 10.9$$
$$\hat{Z} = 0.33\,PW + 7.3$$

A systematic review conducted by Kainz et al. (2017) suggested that the Harrington regression equations and the geometric sphere fit were the most accurate predictive and functional methods.

Kainz et al. (2017) investigated the accuracy and reliability of different scaling methods (with or without joint center) for cerebral palsy participants and typically developed participants. MRI was used as the gold standard for locating anatomical landmarks. In joint center scaling methods, the Harrington regression equation was applied to estimate the hip joint center. While the knee joint center was determined as the midpoint between medial and lateral epicondyles markers, the ankle joint center was located at the midpoint between the medial and lateral malleolus markers. Additionally, different combinations of pelvic width, depth, and height were utilized to scale the pelvis. According to the results, the hip joint center-knee joint center method for scaling the thigh was proved more accurate than the KNEE-ASIS (Anterior superior iliac spine) approach in the cerebral palsy participants. When using PD for scaling, it resulted in the highest errors in comparison to the MRI measurement. Therefore, the pelvis should be scaled with width and height without pelvis depth. Compared to the methods of scaling with surface markers, scaling with joint center significantly increased the accuracy of thigh and shank segment length estimations. However, there is no significant difference between the methods in participants typically developed which may be due to the similarity between the static pose and neutral position of the generic model in typically developed participants. Moreover, the poor results from the KNEE-ASIS method may attributed to the variability in anterior pelvic tilt of cerebral palsy patients during static position.

6 RESEARCH QUESTIONS

Running is one of the most popular sports in the world. However, running-induced fatigue has been proven to be associated with a high injury rate (Van Gent et al., 2007). Both inter-limb asymmetries and exercise-induced fatigue are considered risk factors for these injuries (Heil et al., 2020). The relationship between running-induced fatigue and lower limb asymmetry remains unclear. Thus, the aim of this study was to investigate the inter-limb asymmetry using a time-series SPM test and a traditional discrete analysis of SA in recreational runners before and after the fatigue protocol.

Our research questions are as follows:

1） How does the running fatigue protocol affect the SA of the lower limbs?

2） Do the joint kinematic inter-limb asymmetries exist before and after fatigue protocol using SPM continuous analysis?

The hypotheses of the study are as follows:

1) Low limb SA change after running-induced fatigue protocol
2) Lower limbs would display significant joint kinematic asymmetries before and after the fatigue protocol by SPM test.

7 METHOD

7.1 Participants

This work is part of Krista Vohlakari's project: "Prediction of running economy utilizing IMUs". Initially, the project involved 60 subjects. However, due to time constraints, the sample was narrowed down to 20 recreational runners (11 females; 9 males; Age: 35.64±7.72 years; Weight: 69.31±11.31 kg; Height: 171.98±10.42 cm; Running speed: 10.56±1.38 km/h). The inclusion criteria of the sample required that subjects were 18 to 45 years old (male) or 18 to 55 years old (female); running regularly with a weekly volume of 30 to 80 km; able to run 10 kilometers in 1 hour without maximal performance. The exclusion criteria were acute musculoskeletal injury within 6 months of the measurements; acute illness such as flu or fever; chronic disorders which could affect running technique (eg. Neuromuscular problems); respiratory disorders (except asthma with medication for an endurance athlete); moderate or high risk of cardiovascular disorders; being pregnant. All subjects gave their written informed consent prior to participation in the study. All subjects were volunteers and had the freedom to withdraw from the study at any time without penalty.

7.2 Devices

A Vicon eight-camera motion capture system (Vicon Metrics Ltd., Oxford, United Kingdom) was used to collect the marker trajectories. The sampling frequency was 200Hz. A 43-marker set (Plug-In Gait full body model plus extra markers on medial side of elbows, knees, and ankles) was attached to the subjects for kinematic data collection (Figure 7). At the site of attachment, the skin was carefully shaved with a sterile razor to ensure optimal adhesion. Subsequently, a water-resistant kinesiology tape (Vivomed, United Kingdom) was employed to mildly stretch the skin and securely attach the markers. Vicon Nexus 2.11 software was used for the data acquisition.

Before each measurement, the Vicon system was calibrated. All unwanted reflections or detections from external sources were being masked to prevent interference. Calibration was achieved using a T-shaped wand with five markers, which was waved throughout the capture volume. This movement was to ensure that more than 1000 frames were caught up by the Vicon system. The calibration was successful until the feedback value was less than 1. The origin was set by positioning the cameras around the wand.

The motorized treadmill (OJK-1, Telineyhtymä, Kotka, Finland) was set at 1% grade so it could better represent overground running from a physiological perspective. A heart rate monitor (H10, Polar Electro-OY, Kempele, Finland) was attached to each subject's chest to record their HR during the whole measurement. Blood lactate levels were taken during each interval by using capillary blood samples from the fingertip. These samples were analyzed by a Biosen C-line glucose analyzer (EKF Diagnostics, Magdeburg, Germany) to assess blood lactate concentration.

A) B)

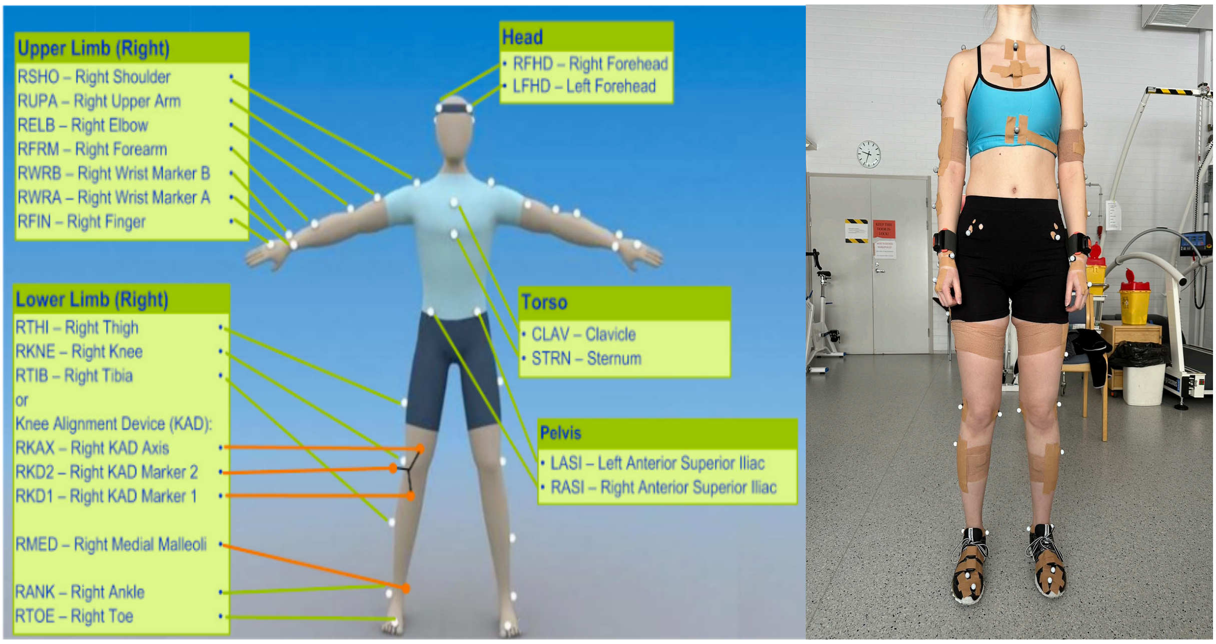

FIGURE 7. A) Plug-In Gait full body model. B) Plug-In Gait full body model plus extra markers on the medial sides of elbows, knees, and ankles.

7.3 Protocol

After subjects arrived at the laboratory, each subject was required to change into tight-fitting shorts with cutting holes on it, which were supplied by the laboratory. Subjects were instructed to wear their own shoes (Shoe brand and model were recorded before measurement). Anthropometric measurements (including height, body mass, leg length, knee width, ankle width, shoulder offset, elbow width, wrist width, and hand thickness) were measured and recorded by the training researcher before attaching the markers. Height and body mass were recorded with and without shoes. To ensure no marker displacement during the experiments, all markers were attached directly to the skin and further secured with athletic tape. However, foot and sternum markers were put on the subject's tops and shoes.

Once all markers were attached to the subject, a static calibration trial was conducted. First, subjects were required to perform a neutral stance with their feet shoulder-width apart, fully extended knees, maintaining a neutral hip alignment, and keeping the trunk upright. Subsequently, subjects needed to switch from a neutral position into an anatomical position with their palm facing the front. Followed by anatomical position, a 'motorcycle' pose was performed. Each posture was required to hold for at least 3 seconds. The joint angle was defined as 0 degrees based on the static calibration. After that, subjects moved on to a dynamic calibration. Subjects needed to move different body parts through the full range of motion during dynamic calibration. This process could help to identify the range of motion of different body parts and label markers in Vicon Nexus.

Subjects were scheduled for two separate visits: familiarization and experimental protocol. The objectives and protocols are outlined below.

7.3.1 Familiarization

The goal of the first visit was to find out the running speed when subjects reached a blood lactate concentration of aerobic threshold. This session included familiarization with the treadmill, the respiratory gas analysis mask, and the blood lactate taking process. First, the blood lactate was taken from the fingertip for baseline lactate measurement. The treadmill familiarization protocol consisted of a nine-minute run, which was divided into three 3-minute intervals based on each subject's self-selected moderate running speed. After each running interval, the treadmill was stopped for taking lactate and giving a Borg Rating of Perceived Exertion (RPE) scale (ranges from 6 to 20). After restart, the treadmill speed was increased by 1 km/h.

7.3.2 Experimental protocol

The experimental protocol was divided into four parts, including warm-up, pre-measurement, fatigue protocol, and post-measurement. Before pre-measurement, subjects conducted a 5-minute warm-up session and self-selected stretching. The baseline blood lactate was taken after the warm-up session. Following the warm-up, a one-minute pre-measurement phase started at the speed according to the subjects' aerobic threshold.

The fatigue protocol started with a relatively low speed, which was then increased by 1 km/h after every four minutes. The treadmill stopped after each four-minute interval to collect another fingertip blood sample. During each resting interval, subjects were also requested to give their perceived exertion using a Borg RPE 6-20 scale. The test was continued once the blood sample and RPE were taken. Subjects needed to run until they wanted to stop or could not maintain the speed.

After the fatigue protocol, subjects started a brief cool-down at a self-selected speed for approximately 3 minutes. The post-measurements were required to start within 5 minutes after the fatigue protocol, including a one-minute run at the same speed as the pre-measurements to ensure consistency in post-measurement. During both pre- and post-measurement Vicon motion analysis data was collected.

7.4 Vicon data pre-processing

Vicon Nexus (v2.11) was used for marker trajectory processing. Each Subject's static calibration trial was used to build up a model based on the Plug-In Gait full-body model (with extra markers). Markers were then automatically labelled using a pipeline. During each step, all markers were visually checked by researchers to make sure there was no mislabeling. Gaps were filled using the function in Vicon Nexus. Following the data process in Vicon Nexus, the 3D coordinates of all labelled markers were exported to C3D files. Matlab (R2022b) scripts were used to convert the C3D file into TRC files.

7.5 Inverse kinematics

OpenSim 4.5 was used for inverse kinematics data analysis, and the gait2392 model designed by Delp et al. (1990) was set as a generic model. A motorcycle position from the static calibration was used for the scaling process to scale the generic based on the subject's anthropometry data. Approximately 2 seconds of static posture was taken from static calibration for scaling. Scale factors were calculated from the pre-determined anatomical landmarks. All the markers' weights were set as a default value of 1 to ensure consistency. The scaling process kept iterated until the root mean square (RMS) error was reduced to less than 1 cm and the maximum error was reduced to less than 2 cm based on the OpenSim tutorial. Once the RMS met the requirements, the subject-specific model was saved and used to continue inverse kinematic process. Matlab (R2022b) scripts were used for the OpenSim inverse kinematic process, and once the kinematic results were generated, the error for each marker was generated during the inverse kinematics process. The root mean square (RMS) error of the kinematic process was less than 2 cm while the maximum error was limited to 4 cm (Figure 8).

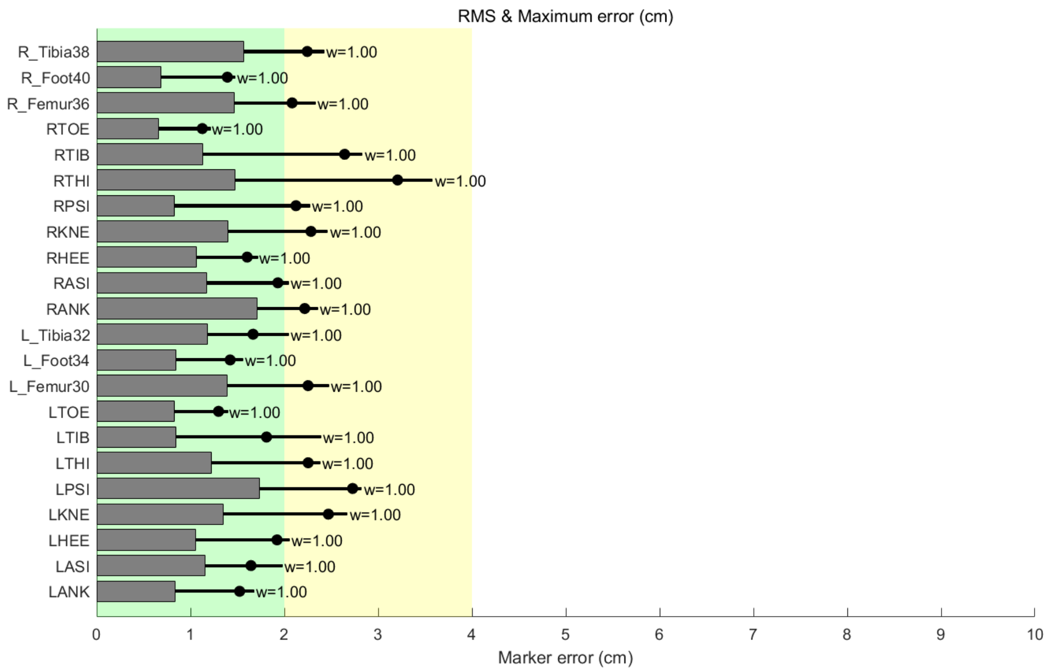

FIGURE 8. Demonstration of RMS & maximum marker error of participants (John, 2021)

7.6 Running analysis

All kinematic data were analyzed by Matlab (R2022b) software. The data was filtered by a low-pass fourth-order Butterworth filter with a cutoff frequency of 8 Hz (determined by residual analysis). All the running trials were visually checked to ensure the rearfoot running pattern. The initial contact phase was determined by identifying the point of minimal vertical displacement of the heel marker. The toe-off event was recognized by finding the minimum knee angle that happened after initial contact. The gait cycle recognition was completed with a pattern recognition algorithm (Fellin et al., 2010). After time normalization of the kinematic data, the gait cycles were analyzed by averaging the data over ten steps.

7.7 Symmetry angle (SA)

According to Zifchock et al. (2008), the relationship between the left and right discrete values is frequently measured by SA. The left and right side values create a vector which forms an angle in the coordinate system. Any angle deviating from 45° represents a certain degree of asymmetry.

The formula for SA is as follows, X_{left} and X_{right} represent the left and right side value, respectively:

$$SA = \frac{(45° - \arctan(\frac{Xleft}{Xright}))}{90°} \times 100\%$$

While 0% SA value means perfect symmetry, and 100% indicates equal but opposite direction values which is complete asymmetry. If [45°-arctan (X_{left}/X_{right})] exceeds 90°, the subsequent formula needs to be used:

$$SA = \frac{(45° - \arctan\left(\frac{Xleft}{Xright}\right)) - 180°}{90°} \times 100\%$$

Peak joint angles during stance phase were used for SA value. SA values were rectified to positive values for comparison. Mean and standard deviation of hip, knee and ankle SA before and after fatigue protocol were calculated.

7.8 Statistical analysis

First, the normality of data needed to be verified by the Shapiro-Wilks test. When the normality of data could not be assumed, a non-parametric statistic method was employed. Statistical analysis was performed in SPSS (IBM SPSS Statistics 26, IBM Corporation, USA). Wilcoxon signed-rank test was used to compare the SA for joint angle before and after fatigue protocol. One-dimensional Statistical Parameter Mapping (SPM_1d, paired sample t-test algorithm package) was then used to assess the joint kinematic between limbs across pre- and post-fatigue measurement in Matlab (R2022b). When SPM{t} values exceed the threshold, it indicates significant differences exist between legs in the corresponding part of the time series. A value of $p<0.05$ was considered significant for analysis.

8 RESULTS

8.1 Symmetry angle (SA)

As shown in Table 1, the SA of the hip peak abduction angle significantly decreased by 7.9% after the running-induced fatigue protocol (p=0.021). Meanwhile, the SA of peak internal rotation after fatigue was significantly higher (9.8%) than that before fatigue (p=0.023). For hip flexion, extension, adduction, and external rotation, there were no significant changes in SA after running-induce fatigue protocol.

TABLE 1: Changes in hip joint SA before and after fatigue.

Hip	**Symmetry angle (%)**				
Joint angle	**Pre- mean**	**±SD**	**Post- mean**	**±SD**	**Sig.**
Flex (+)	1.94	1.26	2.13	1.55	0.56
Ext (-)	6.37	7.27	6.96	7.31	0.25
Add (+)	11.40	8.40	10.65	8.62	0.30
Abd (-)	38.72	26.92	30.83	19.80	**0.021***
Intr (+)	38.48	27.42	48.28	31.85	**0.023***
Extr (-)	35.41	24.74	41.05	27.78	0.30

It can be seen in Table 2 that there were no significant differences found for knee SA before and after fatigue angle.

TABLE 2: Changes in knee joint SA before and after fatigue.

Knee	**Symmetry angle (%)**				
Joint angle	**Pre- mean**	**±SD**	**Post- mean**	**±SD**	**Sig.**
Flex (+)	1.65	1.28	1.56	1.08	0.55
Ext (-)	11.36	9.14	13.18	8.53	0.43

The changes in ankle SA before and after running-induced fatigue protocol are shown in Table 3. There were no significant changes in SA with fatigue protocol.

TABLE 3: Changes in Ankle joint SA before and after fatigue.

Ankle	Symmetry angle (%)				
Joint angle	**Pre- mean**	**±SD**	**Post- mean**	**±SD**	**Sig.**
Dors (+)	3.76	3.28	8.65	18.56	0.48
Plant (-)	4.99	3.56	6.04	5.26	0.60

8.2 SPM results of kinematics

8.2.1 Kinematics parameters before fatigue

The SPM_1d test results of the hip joint before fatigue are shown in Figure 9. There were no significant differences in lower limb joint angles before fatigue (p>0.05).

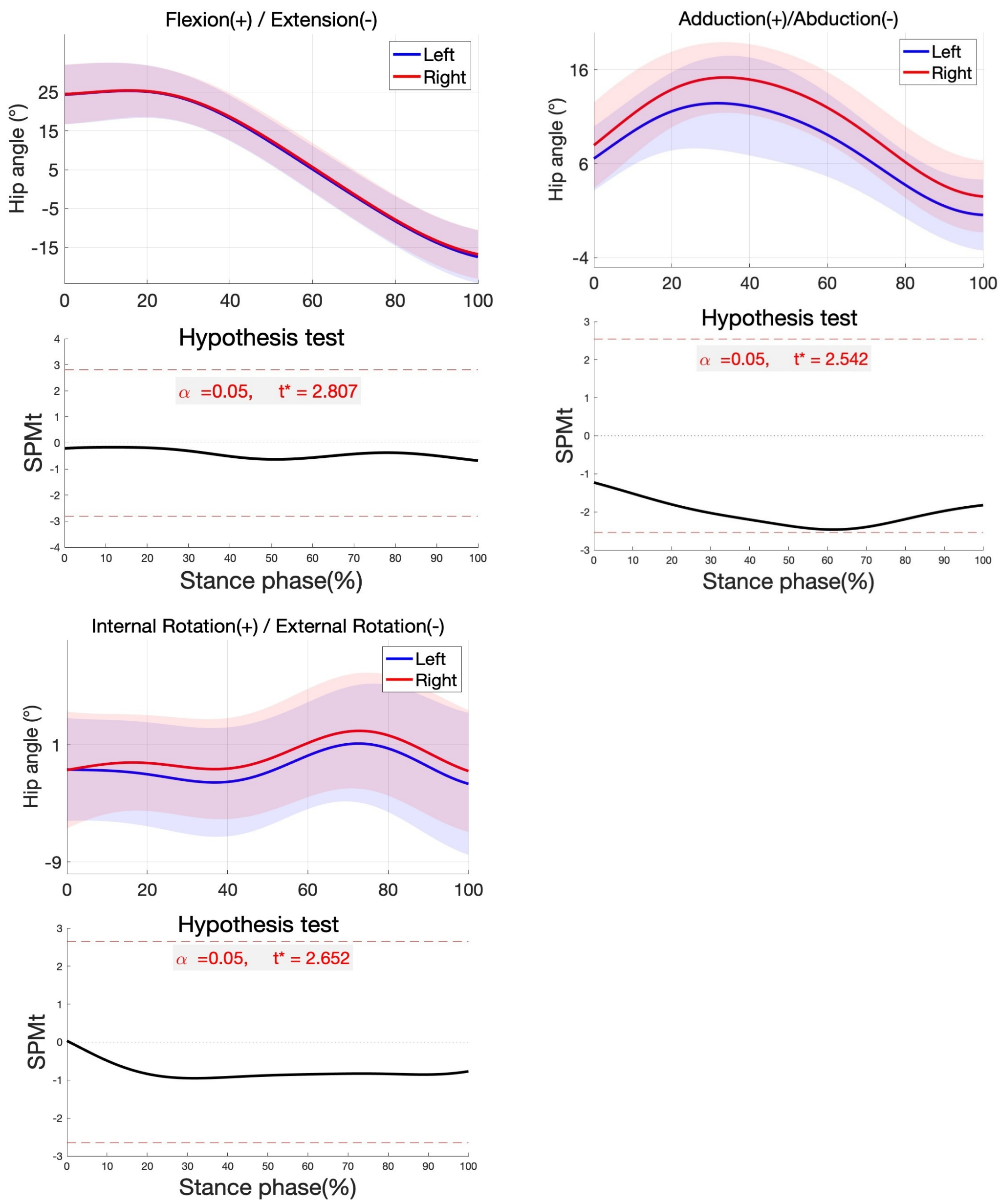

FIGURE 9. A comparison of hip joint kinematic data between the left (blue) and right (red) lower limbs throughout the stance phase by SPM_1d analysis before fatigue protocol.

The SPM_1d test results of the knee and ankle joint before fatigue are shown in Figure 10, and Figure 11, respectively. There were no significant differences in lower limb joint angles before fatigue (p>0.05).

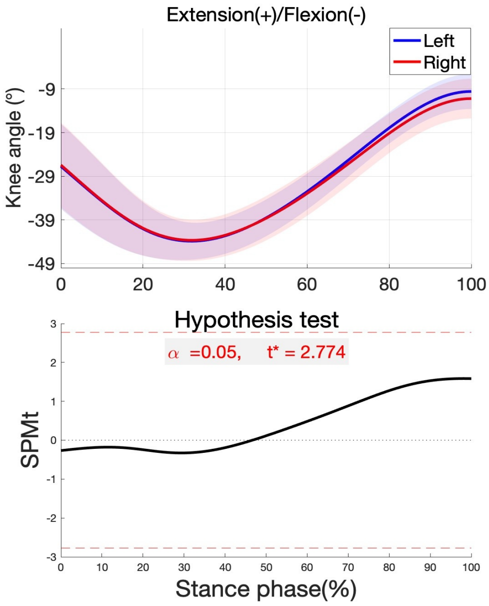

FIGURE 10. A comparison of knee joint kinematic data between the left (blue) and right (red) lower limbs throughout the stance phase by SPM_1d analysis before fatigue protocol.

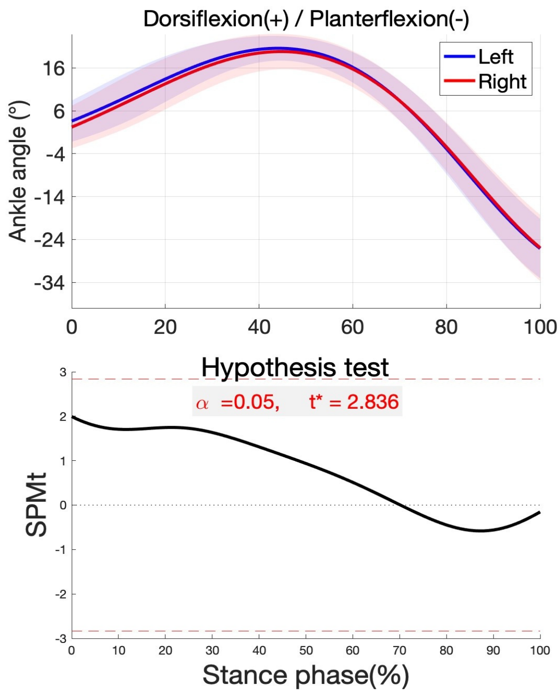

FIGURE 11. A comparison of ankle joint kinematic data between the left (blue) and right (red) lower limbs throughout the stance phase by SPM_1d analysis before fatigue protocol.

8.2.2 Kinematics parameters after fatigue

As presented in Figure 12, SPM_1d test results of the hip joint showed no significant difference in SA during the post-fatigue stage (p>0.05).

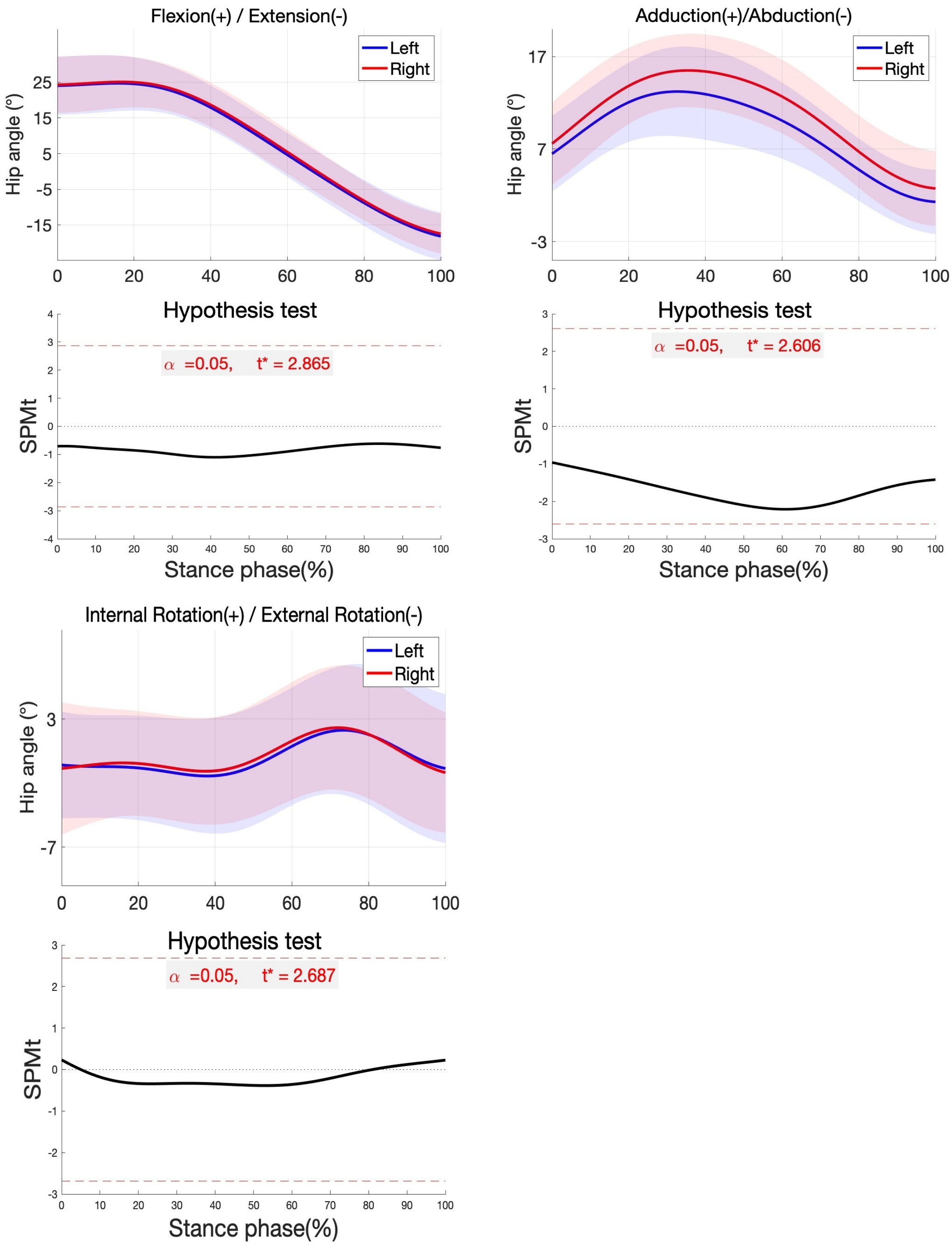

FIGURE 12. A comparison of hip joint kinematic data between the left (blue) and right (red) lower limbs throughout the stance phase by SPM_1d analysis after fatigue protocol.

As shown in Figure 13 and Figure 14, SPM_1d test indicates no significant change between limbs in both knee and ankle joint angles after fatigue.

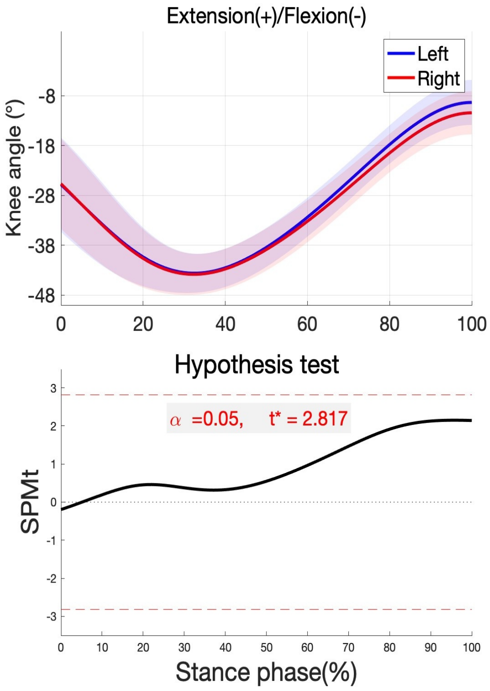

FIGURE 13. A comparison of knee joint kinematic data between the left (blue) and right (red) lower limbs throughout the stance phase by SPM_1d analysis after fatigue protocol.

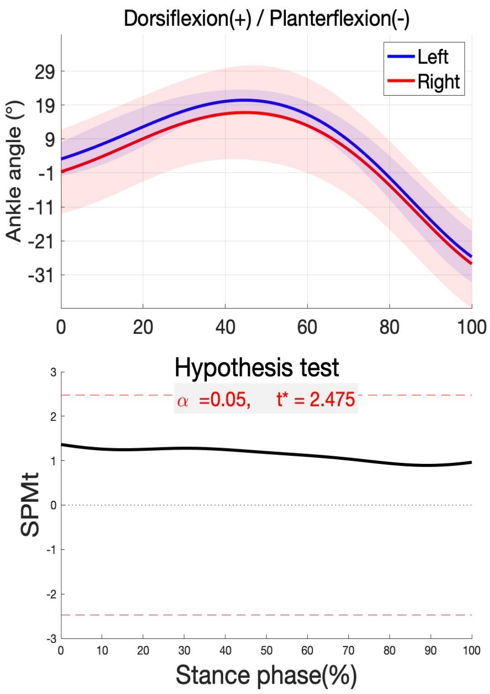

FIGURE 14. A comparison of ankle joint kinematic data between the left (blue) and right (red) lower limbs throughout the stance phase by SPM_1d analysis after fatigue protocol.

9 DISCUSSION

In consistent with the first hypothesis, the SA results showed that the asymmetry between limbs changed after the fatigue protocol. Specifically, the SA of the hip joint angle changed after the running-induced fatigue protocol. However, the SPM_1d test showed no significant differences between limbs in both pre- and post-fatigue measurements, which contrasts with the second hypothesis.

9.1 Symmetry angle (SA)

In the present study, it was observed that the SA of hip abduction decreased significantly by 7.9% after running-induced fatigue protocol, while the SA of hip internal rotation exhibited an increase of 9.8%. This finding agrees with Gao et al. (2022), who reported a reduction in the SA of peak hip abduction by 39% and significantly increased SA of hip internal rotation by 36%. The decrease in SA suggests that the hip abduction angle becomes more symmetrical after fatigue, which may be a potential compensation mechanism to maintain lower extremity stability after fatigue protocol (Gao et al., 2020a).

According to Neumann (2010), the hip abductor serves an important functional role during the single-limb support phase of gait. During this phase, the hip external adduction torque increases dramatically in the coronal phase once the contralateral limb leaves the ground. In response, the hip abductors generate an abduction torque to stabilize the pelvis. Moreover, Neumann also proposed that the same muscle group might need to generate a smaller internal rotation torque at the stance hip to align the pelvis with the movement direction of the contralateral "swing" limb. It is worth noting that both the gluteus medius and minimus muscles (potentially the tensor fascia latae) are able to produce both hip abduction and internal rotation torques at the same time. Thus, the different changes in the SA of hip abduction and internal rotation after fatigue may serve as a complex interaction of biomechanical adaptations. Such adaptations are likely to maintain stability and functional efficiency in response to the altered neuromuscular system from fatigue protocol, thus reducing the risk of injury.

Radzak et al. (2017) conducted a similar study to investigate the impact of fatigue on SA. However, their finding revealed no significant difference in SA before and after fatigue. Similar to our fatigue protocol, Radzak et al. adopted a Modified Astrand protocol graded exercise test to determine VO_{2max}. However, they did not report the sports background of the participants, and our study only recruited recreational runners. Variations in the background of participants might have led to different gait asymmetry results after fatigue protocol.

9.2 SPM_1d of kinematics

In line with the study of Brown et al. (2014), our study found no differences in hip, knee, or ankle kinematic variables between the left and right sides during pre- and post-fatigue measurements. While Brown et al. extracted peak kinematic values, we employed the SPM method to analyze the kinematics across time series.

Vial et al. (2023) examined the kinematics in semi-professional soccer players between the dominant and non-dominant legs in thirteen male soccer players using SPM. In contrast to our results, they found that the peak hip flexion angle in the non-dominant leg was significantly higher than in the dominant leg during the retraction-protraction transition point. This distinction might result from the difference between athletic demands and training regimens of soccer players and recreational runners. This inter-limb asymmetry in soccer players may be due to unilateral kicking and higher repetition movement with the dominant leg (DeLang et al., 2017). In addition, their study only included male participants while we recruited both genders, potentially contributing to the different outcomes.

Various fatigue protocols might have a potential impact on the results. In our study, fatigue protocol was a VO_{2max} test, which combines both aerobic and anaerobic metabolism. Vial et al. (2023) designed a fatigue protocol with three maximal 50-m sprinting which is a short-term high-intensity exercise. The speed of protocol has an influence on the running kinematics during and after fatigue protocol (Zandbergen et al., 2023). Verbitsky et al. (2018) required the participants to run on a treadmill at the anaerobic threshold speed for 30 min. They divided their participants into fatigue and non-fatigue groups based on their end-tidal CO_2 pressure. They found that the fatigue group showed a significant increase in peak tibial and peak sacral accelerations. This phenomenon did not present in the non-fatigue group. One participant from the non-fatigue group of the first visit conducted a second run at a higher speed on another day. Interestingly, this participant fell into the fatigue group this time and showed an increase in peak tibial accelerations. Their finding indicates that there might be a subject-specific threshold based on running duration and distance. According to Britannica (2024), fatigue protocol distance required a minimum distance of 3 km to set a lower limit for running-induced fatigue and meet the criteria for long-distance running. Thus, the sprinting fatigue protocol of Vial et al. might not be sufficient to induce running fatigue.

Previous studies have found that 15% served as a symmetry threshold to identify the inter-limb asymmetry (Barber et al., 1990; Kyritsis et al., 2016). Based on the criteria, SA of hip abduction, internal rotation, and external rotation all exhibited asymmetry before and after fatigue protocol. However, SPM did not identify any differences between limbs before and after fatigue protocol. This might be due to the threshold for significant differences in SPM is high since there is a multiple comparison correction in SPM (Hughes-Oliver et al., 2019). Although SPM has been successfully used in previous studies to identify inter-limb asymmetry of muscle activation, kinetics, and kinematics (Morais et al., 2022; Vial et al., 2023; Robinson et al., 2014). Our results did not indicate that asymmetry is non-existent in runners, it is possible that this research is not powered enough to identify the inter-limb asymmetry using SPM methods. Thus, a larger sample size might be required for SPM to detect all inter-limb asymmetry.

9.3 Limitations

Neuromuscular fatigue can be defined as peripheral fatigue and central fatigue. Peripheral fatigue suggests the impairment of muscle function, while central fatigue represents a reduced ability of the central nervous system to activate muscles. According to Carroll et al. (2017), central fatigue typically recovers within 2 min after short, high-intensity activity. However, the full recovery of muscle function takes a few hours. In our study, the subjects were required to start post-fatigue measurement within 5 min after fatigue protocol. The effects of central fatigue might have diminished over time, but the muscle function could still be incomplete. In our pilot study, we attempted to conduct immediate post-measurements. However, due to the intense nature of the $VO_{2\,max}$ test used in our fatigue protocol, participants were unable to perform these measurements immediately. Thus, we followed the same interval as Gao et al. (2022) and (Encarnación-Martínez et al., 2022) to allow a five-minute rest before post-measurement.

In this study, both the fatigue protocol and data collection were conducted using a treadmill. By using a treadmill, it is easier to maintain a constant speed and ensure a smooth surface, avoiding obstacles and inclinations on overground running. Consequently, treadmill running produces lower variability in movement patterns compared to outdoor running (Hanley &

Tucker, 2018). Although a previous study indicated treadmill running biomechanics analysis can be generalized to overground running, there remain controversies regarding the differences in running environments (Riley et al., 2008). For example, Van Hooren et al. (2020) included 33 studies related to running environments in a systematic review and meta-analysis. Compared to overground running, the sagittal foot-ground angle at foot strike, knee flexion range of motion, from footstrike to peak during stance, and vertical displacement of center of mass was lower in the treadmill, while other parameters such as knee flexion at foot strike and contact time were higher.

When considering the relationship between fatigue and the running environment, there is an interaction associated with foot strikes (Strohrmann et al., 2012). Fatigue in overground running leads to a reduction in running speed, increased running time, and decreased cadence and stride length. For expert athletes, they tend to adjust their pace when running overground (Dierks et al., 2010). However, we use the constant speed on the treadmill to detect the fatigue effect which can prevent the runners from modulating their pace. Thus, various running environments may have a potential effect on inter-limb asymmetry.

Another limitation of this study was that the dominance of lower limb was not recorded. However, previous studies found that the effect of dominance in explaining asymmetry is insignificant (Brown et al., 2014).

Finally, the SA is a measurement of extreme values that only capture a single time point (Vial et al., 2023). Although SPM applies to continuous data analysis, this study conducted an SPM pair t-test which only examines the difference between limbs. The interaction between timing and limbs for symmetry should be examined using SPM two-way ANOVA should be examined in future studies. Future studies should take the continuity into account by using different symmetry indices, such as SI_{Nigg} (Nigg et al., 2013).

10 CONCLUSION

This study investigated the effects of running-induced fatigue on lower extremity symmetry in recreational runners. This study adopted a discrete symmetry parameter and a time-series analysis for quantifying fatigue. SA quantifies the inter-limb symmetry using the peak joint angle. In line with the previous study (Gao et al., 2020), our study reveals that the SA in the lower limbs was significantly affected by fatigue. Particularly, the hip abduction SA significantly decreased after fatigue, while the SA of hip internal rotation increased following fatigue. An increase in asymmetry contributes to the high non-contact injury risks in fatigue states (Heil et al., 2020). However, hip abduction SA became more symmetrical, and this may suggest an interaction mechanism of biomechanics to maintain lower limb stability.

Statistical parametric mapping (SPM) provided information about the magnitude and shape of waveform through the data set. Nevertheless, the threshold for significance of SPM is higher which might lead to failure for asymmetry detection. Future studies should expand the sample size or take the continuity into account by using different symmetry indices, such as SI_{Nigg} (Nigg et al., 2013).

In conclusion, this study enhances the understanding of the kinematics consequences of fatigue in running. The strength coaches and physiotherapists can better design training and rehabilitation programs to reduce injury risk and improve running performance. By adopting different methods to quantify asymmetry, we can better design a research methodology for the future study.

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