Stride-to-Stride Variability While Prolonged Running on a Treadmill and Over-Ground

by

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Statement of Originality

I, IQBAL, Zaheen Ahmed, hereby declare that I am the sole author of the thesis and the material presented in this thesis is my original work except those indicated in the acknowledgement. I further declare that I have followed the University's policies and regulations on Academic Honesty, Copyright and Plagiarism in writing the thesis and no material in this thesis has been submitted for a degree in this or other universities.

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Abstract

Long-distance running offers both health benefits and an increased risk of running-related injuries due to the strain on joints. Understanding how the body responds to different running environments is crucial for preventing injuries in long-distance runners. Due to the limitations of traditional gait monitoring systems there is a research gap in long-term gait monitoring in natural settings to understand how gait patterns changes with duration and surface of running and runner's body response different demands of running environments and surfaces during prolonged running outside controlled environments in laboratories. This doctoral thesis focuses on analyzing the stride-to-stride variability during prolonged running, specifically investigating how stride variability changes with running duration and surface type (treadmill vs. over-ground) during sagittal plane motions using inertial measurement units. Eleven runners (2 females) were instructed to run on treadmill and over-ground track and 7 inertial measurement units (APDM Inc., Portland, OR, USA) were used to measure stride time and lower limb joint angles throughout 30-min running at the preferred speed. Coefficient of variation was used to calculate variability of stride time and lower limb joint angles in different phases of gait cycle. Coordination and its variability were calculated using continuous relative phase for knee-hip, ankle-knee, and ankle-hip joint couplings of the dominant side in overall gait cycle as well as stance and swing phases. A two-way 2 x 2 repeated-measures mixed-design analysis of variance was used to compare the mean and variability of all the parameters between initial and final 5 minutes of running and the two running surfaces (treadmill and over-ground) with statistical significance set at p less than 0.05. Post-hoc analysis was conducted using the Bonferroni correction. Key results indicate no significant differences in stride time or its variability between two running durations or surfaces. However, higher variability was observed in different joint angles during over-ground running and in the initial duration of the run, aligning with the study's hypothesis. Analysis of lower limb joint coordination revealed significant differences in joint coordination at specific gait phases, with increased coordination observed in the final duration of running and during fatigue-induced conditions. There were no significant differences in coordination variability between two running durations or surfaces. In conclusion, this study underscores the significance of investigating gait organization and control during long-distance running in natural settings outside the laboratory. The comprehensive approach taken in this research provides valuable insights into the interaction between the body and the environment in explaining variations in spatial-temporal gait parameters. This study provides insights into gait motor control under fatigue and tailor interventions for injury prevention and performance improvement for various groups of runners. This study also supports the potential for future research to utilize inertial measurement units in outdoor settings for long-term data collection on running biomechanics, with larger-scale studies involving diverse populations and running conditions to enhance the understanding of human movement complexities.

Keywords: Stride-to-stride variability; long-distance running; inertial measurement units; treadmill and over-ground running; running kinematics

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List of abbreviations

1. ApEn: Approximate entropy
2. CRP: Continuous relative phase
3. CRPv: Continuous relative phase variability
4. CV: Coefficient of variation
5. DFA: Detrended fluctuation analysis
6. IMU: Inertial measurement units
7. RRI: Running-related injuries
8. SampEn: Sample entropy
9. SD: Standard deviation
10. SSV: Stride-to-stride variability

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Chapter 1: Introduction

1.1 Rationale and Justifications of This Study

Running is a childhood acquired skill that involves fast cyclical movement of the lower limbs and consists of multiple strides, i.e. consecutive strikes of the same foot (Dugan & Bhat, 2005; Malina et al., 2004). Due to the associated health benefits, long-distance running, like full marathon, has become a popular activity around the world (Fields et al., 2010; Lee et al., 2014; Williams, 2009). Runners with higher number of marathon participation have been shown to suffer lesser joint pain, and those who have completed at least 5 marathon races exhibited lower prevalence of arthritis (Ponzio et al., 2018). Runners with greater experience of running long distances develop better understanding of extreme fatigue condition that cannot achieved under normal circumstances. Such experience increases fatigue tolerance by enhancing their body capacity by in response to the dynamic external as well as internal environment. External environment here refers to outside running environment including temperature, humidity, ruining surface, etc. while internal environment refers to different physiological and other changes happening inside the body as a response to running.

With more and more people choosing to run as a part of recreational activity and means to attain fitness, incidence and prevalence of running-related injuries (RRI) have also remarkably increased (Kluitenberg et al., 2015). Lower limbs are the most common sites for RRIs and the incidence rates of overuse injuries, like iliotibial band syndrome and patellofemoral pain, and ranges from 11% to 92.4% (Kluitenberg et al., 2015; Lopes et al., 2012; Tschopp & Brunner, 2017; Van Gent et al., 2007). In order to prevent the development of RRIs, it is important to understand running biomechanics. Based on cognitive theory to motor control, various studies have investigated the biomechanical factors, like running pattern and shoes, that may lead to development of RRIs (Breine et al., 2017; Lieberman et al., 2010; Van der Worp et al., 2016; Zhang et al., 2017). There are limited studies about how running mechanics are related to RRIs

(Ferber et al., 2009). To understand this relation further, dynamical systems theory has been cited that suggests stride-to-stride variability (SSV) is an important aspect of running mechanics and can provide insights into RRIs (Bartlett et al., 2007; Davids et al., 2003b; Hamill et al., 2012). According to this theory, SSV is considered a natural and adaptive characteristic of human locomotion.

SSV is the small fluctuations that occur between strides during walking or running (Jordan et al., 2007a). It reflects the flexibility and adaptability of the sensorimotor system to the dynamic internal (inside body) and external (outside) environments during movement (Chau, 2001; Davids et al., 2003a). There are two movement levels while running, the movement at execution level, and the movement at outcome level. (Jordan et al., 2006; Mo & Chow, 2019). Lower limb coordination variability corresponds to movement at execution level while stride time variability corresponds to movement at outcome level. Coordination and its variability have become popular among researchers due its relation with various factors associated with runners like gender, age, status of injury, level of skill, type and speed of running (Boyer et al., 2017; Floría et al., 2019; Hafer et al., 2016; Miller et al., 2008; Nakayama et al., 2010; Silvernail et al., 2015). Despite this, there is a lack of clarity about interpretation of variability data across different running conditions especially under real-world situations. This is important as runners are expected to modify their pattern within and between running sessions depending on their aim, running technique and surface (Floría et al., 2019). However, fewer studies have explored how lower limb joint angles, their coordination and variability vary between treadmill and over-ground surfaces while prolonged running. Stride time has been regarded as the ‘final output’ of the locomotor system and its long-range correlations reflect the adaptability of the locomotor system while gait regulation in response to the constraints of dynamic environments (Hausdorff, 2007; Jordan et al., 2006). Healthy persons have been shown to have stable stride time during walking and treadmill running but with some patterned fluctuations (Jordan et al., 2007a, 2007b). Such variability is functional and dependent on the nature of task performed (Hamill et al., 2012). Stride time variability has also been shown among runners after injury and fatigue which affects neuromuscular output during running (Meardon et al., 2011). As the runners need to reliably, effectively and safely reorganize their body response to the dynamic environment and stress while prolonged running on different surfaces, it is important to study the differences of such variability under these conditions.

SSV at both the levels of movement have been shown to be related to developing RRIs, but they have been investigated separately. Ideally, the impact load over all the structures of joints should be evenly distributed during each stride but higher SSV while running decreases the total load applied on a single structure that may lead to injury (Bartlett et al., 2007; Bertelsen et al., 2017; Gabbett, 2016; Hamill et al., 2012). During each stride while running, the body experiences ground reaction forces that are transmitted through various structures, including bones, muscles, and ligaments, which are quite significant. When there is a higher SSV, it means that there is greater variability in the way forces are distributed across the structures of the joints. This variability can result in certain structures, such as bones, muscles, or ligaments, being subjected to higher forces than others. This can increase the load on these structures and potentially lead to overuse injuries or stress fractures. Higher SSV has also shown to negatively affect running performance which could be due to lower energy efficiency (Belli et al., 1995; Morgan et al., 1989). On the other hand, as compared to healthy counterparts, lower SSV has been shown in patients with iliotibial band syndrome, patellofemoral pain syndrome, chronic ankle instability, and other similar injuries (Brindle et al., 2020). Hence, performance of motor tasks like running can be affected by both high and low variability (Davids et al., 2003a; Hamill et al., 2012; Srinivasan et al., 2015). Overall, maintaining a balanced and consistent distribution of forces during running is important for minimizing the risk of injury. Higher or lower SSV can disrupt this balance, leading to an uneven load distribution and potentially increasing the risk of injury. Although a link between SSV and RRIs has been widely suggested (Hamill et al., 1999; Lilley et al., 2018; Miller et al., 2008), the understanding of SSV while prolonged running and as a function of running surface is limited.

The biomechanical effects of SSV on joint and tissue loading can be explained by various factors. It can lead to inconsistent joint alignment and stability, increasing forces on specific structures. Inconsistent muscle activation patterns due to SSV can result in uneven force distribution across joints, increasing stress on tissues. It can also impact shock absorption, causing inconsistent force distribution and potentially increasing load on structures. Additionally, SSV can alter gait mechanics, affecting force distribution and joint loading. Further research is needed to provide detailed insights in biomechanics and running gait analysis. SSV has been shown to be affected by various factors like running speed, fatigue, injury and running experience (Fuller et al., 2017; Maas et al., 2018; Mo & Chow, 2018b). It has been shown to be affected by progressive fatigue induced by running a relatively shorter distance (Fuller et al., 2017; Meardon et al., 2011; Mo & Chow, 2018b). Other inconsistent findings have shown that as compared to non-fatigued conditions, there are no SSV differences during fatigued conditions (Fuller et al., 2017; Mann, Malisoux, Urhausen, et al., 2015). However, no studies have yet investigated the effects of fatigue during long-distance running. Multiple studies have indicated that the interaction between running experience and progressive fatigue can influence movement patterns and SSV during running (Maas et al., 2018; Mo & Chow, 2018b; Strohrmann et al., 2012). However, it is worth noting that in these studies, "running experience" refers to the number of years an individual has been running, rather than their subjective experience of running itself. It would be intriguing to investigate the impact of a runner's experience in managing progressive fatigue during long-distance running and how it may affect SSV. This could provide valuable insights into the role of mental and physical fatigue management in relation to SSV.

Recording of kinetic and kinematic data has been traditionally done using multi-camera motion capture system, force plates and treadmills inside the laboratories, which cannot reflect true picture of the on-field outdoor environment. Continuous recording for long duration in a natural sports setting, like prolonged running, has been another challenge for the researchers. Inertial measurement units (IMU) based wearable sensors have helped to overcome these challenges and provided a cost-effective alternative to expensive laboratory-based instruments for unobtrusive on-field activity monitoring (Bussmann et al., 2009; Weiss et al., 2013). IMUs are among the core elements in wearable technology. Being a portable system for estimating human gait kinetics and kinematics, they have been widely used to investigate multiple gait parameters during various dynamic tasks like running (Weiss et al., 2013). Some researchers have proposed the future application of stride time in the prediction of RRIs, as the stride time can now be easily measured using IMUs (Meardon et al., 2011; Mo & Chow, 2018a; Norris et al., 2014). Linear and non-linear analytical methods are two different approaches used to measure SSV in gait analysis. These methods provide a quantitative assessment of gait parameters. Linear analytical methods primarily focus on measuring the kinetic (ground reaction forces, joint moments, etc.) and kinematic properties (joint angles, segment orientation, etc.) of movement. On the other hand, non-linear analytical methods, such as detrended fluctuation analysis (DFA), analyze SSV and complexity at the movement outcome level (Hamill et al., 2012; Jordan et al., 2006; Zhang et al., 2017). These methods consider the small fluctuations between strides that were initially excluded from analysis using linear methods. Non-linear analytical methods have added more meaning to gait analysis by capturing the subtle variations and complexity in movement patterns (Bartlett et al., 2007; Hamill et al., 2012; Hausdorff, 2007; Jordan et al., 2006; Mullineaux et al., 2004; Schwartz et al., 2004). However, there are limited number of studies that have specifically investigated variability in stride time variability while prolonged running, particularly in a natural setting outside laboratory. This suggests that more research is needed to understand the impact of prolonged running on SSV in real-world conditions.

Treadmills have been widely used by athletes and their coaches for training and assessing running performance (Bishop et al., 2017; Cappa et al., 2014; Jones & Doust, 1996). Research and training using treadmills have been popularly used as they require limited space to record wide range of physiological and performance parameters under controlled indoor environment as compared to on-field running (Morin & Sève, 2011). Physiological response and perceived effort required while running on treadmill have been traditionally thought to replicate the body demands and response while running over-ground (Edwards et al., 2017; Taunton et al., 2003). However, researchers comparing such parameters while running on treadmill and over-ground have revealed an opposite picture prompting further research on how running surface and environment can affect running performance (Miller et al., 2019). The main reason is that that indoor treadmill running cannot reliably reflect the natural pattern of gait while over-ground running adopted on outdoor tracks (McCrory et al., 2022). Researchers have also reported various clinical and biomechanical differences in motor control while comparing walking over treadmill and over-ground. Although some of them reported some similar spatiotemporal parameters between treadmill and over-ground, others reported individual differences with increased cadence and shorter strides over treadmill (Alton et al., 1998; Murray et al., 1985; Riley et al., 2007). Such comparisons are more important for rehabilitation settings where walking gait training for non-ambulatory patients is done on a treadmill that is gradually changed to over-ground as the natural variability of the sensory-motor system is altered. Combs-Miller et al. reported greater walking speed and gait symmetry among chronic stroke patients post two weeks of over-ground gait training as compared to training using treadmill for similar duration (Combs-Miller et al., 2014). Such findings raise further concern about adaptability of training on treadmill for over-ground running. There are some key physical differences between a treadmill and a regular running track (Mitchell et al., 2023). First is the surface, a treadmill has a continuous moving belt usually made of rubber or synthetic material, whereas a running track has a solid surface, often made of rubber, asphalt, or other synthetic material. Treadmills typically have a smaller size running surface compared to running tracks. Although the dimensions of a treadmill vary, but the average size is around 20 inches wide and 55-60 inches long. In contrast, a standard running track is 400 meters in length and has multiple lanes, each measuring 1.22 meters wide. Treadmills usually have adjustable incline settings that allow runners to simulate uphill running by increasing the slope. On the other hand, running tracks are usually flat, with no incline. Treadmills often have cushioned surfaces that provide better shock absorption, reducing the impact on joints and muscles. Running tracks, especially those made of rubber, also offer some level of shock absorption, but it may not be as pronounced as on a treadmill. Treadmills have electronic controls that allow users to adjust the speed according to their preference. Most treadmills offer a wide range of speeds, typically up to 12-15 miles per hour. Running tracks, on the other hand, do not have speed control mechanisms as they rely on the runner's own effort to determine the pace. There are some environmental and accessibility factors to consider. When running on a treadmill, there is no need to worry about weather conditions such as rain, wind, or extreme temperatures. However, when running on a regular track, these factors can significantly impact the running experience. Treadmills are readily available in gyms, fitness centers, and even for personal use at home. Running tracks, on the other hand, may require access to a specific facility or public space with a designated track. The choice between the two ultimately depends on personal preference, convenience, and specific training goals.

Despite considerable research on physiological and perceptual responses and performance among runners there are fewer studies that have studied SSV after prolonged running in a natural setting or as a function of running surface. One of the reasons for this is the limitation of traditional motion capture systems that cannot be used outside the laboratory. However, with the recent developments in portable technology in the recent years and availability of wearable sensors have enabled gait analysis in actual daily life and sports settings (Bussmann et al., 2009; Weiss et al., 2013). IMUs can be utilized to differentiate the motor behavior associated with running on different surfaces under different environment, i.e. indoor over treadmill and outdoor over track, especially while running for prolonged duration. Such quantification of gait and underlying motor behavior can help to further understand dynamic stability, risk of fall and development of injury while performance of different tasks under different conditions. Such gait monitoring can also provide more information to design more injury prevention and development of effective rehabilitation and training protocols (Dobkin & Dorsch, 2011).

The research gap regarding monitoring gait variability can be summarized in the following way. There is a lack of research focusing on long-term gait monitoring to understand how gait patterns change over time and their relationship with injury occurrence, running duration and surface. Longitudinal studies are needed to capture gait parameters in real-life settings and provide insights into injury prevention and rehabilitation strategies. While gait analysis often involves the use of various sensors (e.g., accelerometers, gyroscopes, pressure sensors), there is a need for research on integrating multiple sensors to obtain comprehensive gait data. This would enable a more accurate assessment of gait parameters and facilitate the development of personalized rehabilitation and training protocols. Research is needed to explore the potential of real-time feedback systems based on gait monitoring data. Providing immediate feedback to individuals about their gait mechanics could help correct faulty movement patterns, reduce injury risk, and optimize rehabilitation and training outcomes. There is a need for objective outcome measures that can be derived from gait monitoring data. Currently, most assessments rely on subjective measures, such as self-report questionnaires or clinician observations. Developing objective measures based on gait monitoring would provide more accurate and reliable data for injury prevention and rehabilitation protocols. Research is needed to investigate the effectiveness of individualized approaches in injury prevention and rehabilitation based on gait monitoring data. Understanding how specific gait parameters vary across individuals and their relationship with injury risk can help tailor interventions to address each person's unique needs. There is a growing interest in using machine learning and artificial intelligence techniques to analyze gait monitoring data. Research is needed to explore the potential of these approaches in identifying injury risk factors, predicting injury occurrence, and optimizing rehabilitation and training protocols. Addressing these research gaps would enhance our understanding of gait mechanics, improve injury prevention strategies, and facilitate the development of more effective rehabilitation and training protocols.

1.2 Objectives

As it is apparent from the literature review that there are inconsistent findings about the effect of various factors associated with running on SSV. Although the running experience as the years of running practice and progressive fatigue have been shown to affect SSV and movement pattern while running (Maas et al., 2018; Mo & Chow, 2018b; Strohrmann et al., 2012), the effect of experience while prolonged running outside the laboratory has not been studied. Additionally, the need to investigate the differences in variability of stride time and coordination while prolonged running indoors on treadmill and outdoors over-ground has been highlighted. Keeping these gaps in the existing literature in mind, this study was conducted with following objectives:

a. To conduct a literature review on running gait variability to identify various factors that can affect it and different methods and parameters used in the literature to quantify it
b. To investigate the effect of duration of running on the stride variability while long-distance running among healthy runners

c. To investigate the effect of surface of running (treadmill and over-ground) on the stride variability while long-distance running among healthy runners

1.3 Hypotheses

a. SSV while long-distance running varies with running duration
b. There are comparable differences in the SSV while long-distance running on treadmill and over-ground
c. As compared to treadmill, there is higher SSV while long-distance running over-ground
d. There will be significant interaction between duration and surface while long-distance running

1.4 Significance and Implications of the Study

The study aims to provide valuable insights for researchers to better understand and apply assessments and training techniques for running indoors on treadmill and outdoors over-ground. Results of this study shall lead to the development of strategies to prevent RRIs and improve rehabilitation and performance outcomes. It is one of the first study to investigate SSV in stride time, lower limb joint angles and joint coordination in sagittal plane movements while long-distance running as a function of running duration and surface outside the laboratory. This would help to further understand regulation of gait motor control with progressive fatigue. This is one of the first study to simultaneously investigate SSV at the movement execution and outcome levels, i.e. lower limb coordination and stride time variability respectively.

The long-term impact of this research stems from the potential association of SSV during running and development of RRIs. Outcome of this study would be helpful to further understand how SSV could be used to benefit different runners. The results of this study would help to tailor interventions specific to the needs of runners such as elite athletes or recreational runners. Understanding how SSV changes over time during prolonged running can provide valuable insights into the biomechanics of running and potentially enhance performance. The results of this study would not only help to formulate easier to implement and effective interventions for the prevention of RRIs but also be a reference for interpreting SSV while prolonged running to enlighten runners about performance enhancement and future determination of optimal gait variability. Results of this study will provide insights of prolonged running into the stride time variability and help to predict RRIs in future studies using artificial intelligence.

This thesis consists of seven chapters focusing on the effect of prolonged running on SSV assessed using IMUs. Overall outline of the thesis is presented in Figure 1.1. CONSORT flow chart and details regarding study design has been presented in Appendix B.

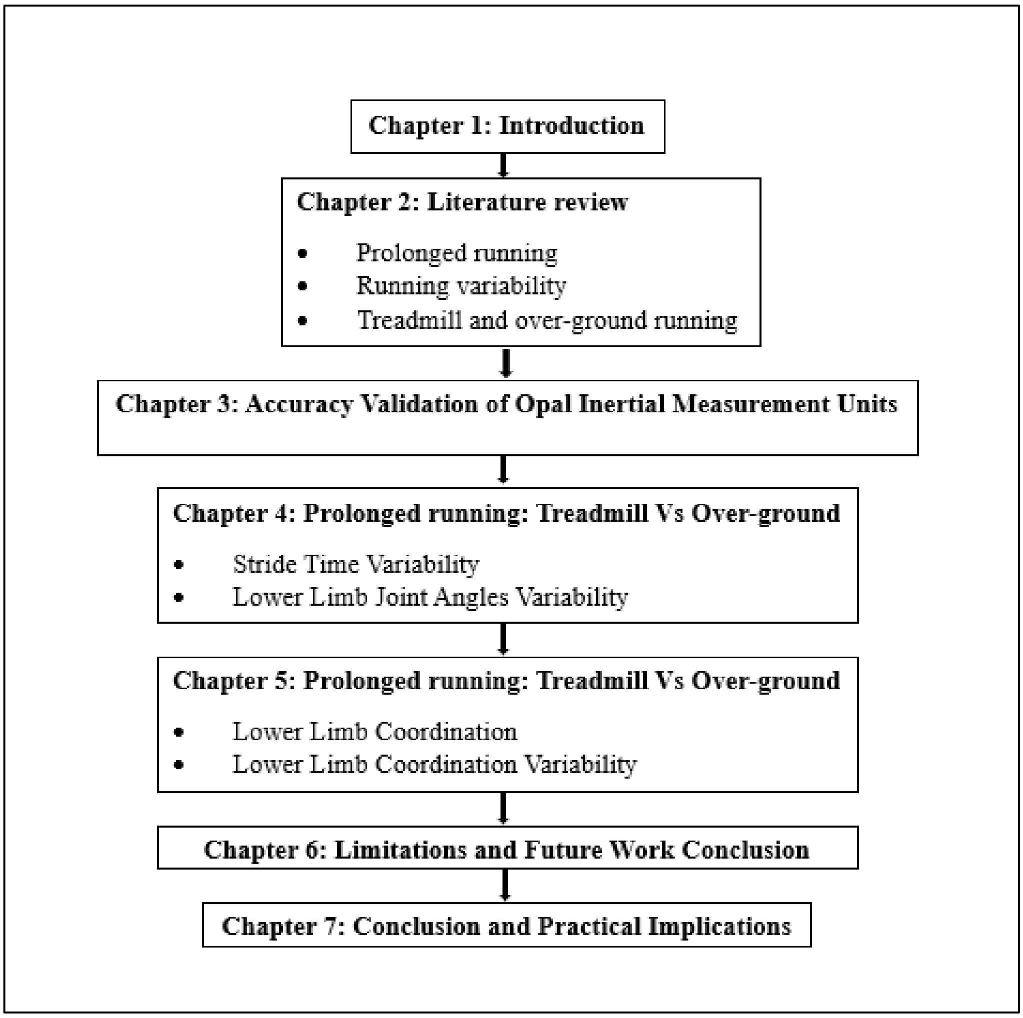

Figure 1.1: Outline of the thesis

Chapter 2: Literature Review

2.1 Prolonged Running

Prolonged running refers to the act of running continuously for an extended period of time, typically beyond the duration of a normal or regular running session. This can vary depending on individual fitness levels and goals, but generally involves running for an extended period of time without stopping or taking breaks. In general, prolonged running is typically considered to involve continuous running for an extended period of time beyond what is considered a standard or moderate running session. Prolonged running challenges the endurance, stamina, and mental toughness of the runner, requiring them to sustain a consistent pace over a prolonged distance. This form of running is often used for endurance training, ultra-marathon preparation, or as a personal challenge to push one's limits and improve overall fitness (Pescatello, 2014). The minimum duration for prolonged running can vary depending on individual fitness levels, training goals, and the specific context in which the term is being used. Ultimately, the minimum duration for prolonged running is subjective and can vary based on individual capabilities and training objectives.

Prolonged running is an effective way to improve cardiovascular endurance, stamina, and mental toughness. By running continuously for an extended period of time, runners can build aerobic capacity, strengthen muscles, and improve overall fitness levels (Cooper, 1998). It is commonly used as a form of endurance training for runners preparing for longer distance races, such as half marathons, marathons, or ultra-marathons. Gradually increasing the duration of running builds the endurance needed to sustain pace over longer distances (Daniels, 2013). Prolonged running is typically done at a moderate pace that allows runner to maintain a steady effort level throughout the run. It is important to pace appropriately to avoid fatigue and complete the run without stopping. Proper hydration and nutrition are essential for prolonged running to maintain energy levels and prevent dehydration. Runners should make sure to drink water before, during, and after the run, and consider fueling with carbohydrates and electrolytes to support their performance. After a prolonged running session, it is important to allow the body time to recover and repair. Proper cool down and refueling with a balanced meal or snack, and adequate amount of rest is essential to support recovery and prevent injury (Millet & Lepers, 2004). It is important for novice runners to gradually increase the duration of running to avoid overtraining and reduce the risk of injury. They need to start with shorter runs and gradually build up your endurance over time.

Research on prolonged running has been conducted to understand the physiological adaptations, performance benefits, and potential risks associated with this form of exercise. Studies have shown that prolonged running can lead to various physiological adaptations in the body, such as increased aerobic capacity, improved cardiovascular function, enhanced muscle endurance, and changes in metabolic efficiency (Hawley & Noakes, 1992). These adaptations help improve performance and endurance during prolonged running activities. Research has demonstrated that endurance training, including prolonged running, can improve running economy, increase lactate threshold, and enhance overall endurance performance (Helgerud et al., 2007). By training at longer durations and higher intensities, runners can improve their ability to sustain a faster pace for extended periods of time. Studies have investigated different training strategies for prolonged running, such as long slow distance training, tempo runs, interval training, and hill repeats. Research has shown that a combination of these training methods can help improve endurance, speed, and overall performance in prolonged running events (Stöggl & Sperlich, 2015). Research has also focused on the importance of nutrition and hydration for prolonged running. Studies have examined the role of carbohydrate loading, hydration strategies, electrolyte replacement, and fueling during long runs to optimize performance and prevent fatigue (Burke & Hawley, 2002). Some research has also explored the potential risks and injury prevention strategies associated with prolonged running. Studies have looked at the impact of training volume, intensity, running mechanics, footwear, and recovery strategies on reducing the risk of overuse injuries in endurance runners (Laursen & Jenkins, 2002). Research has also delved into the psychological aspects of prolonged running, such as mental toughness, motivation, goal setting, and coping strategies during challenging races or training sessions (Callen, 1983). Understanding the psychological factors that influence performance can help athletes improve their mental resilience and performance in endurance events.

2.2 Prolonged Running and Running-Related Injuries

Running over long distances has become a popular physical activity among people of all ages, abilities and ethnicities worldwide with higher number of people participating in marathon events each year (Mo, 2018; Scheerder et al., 2015). Prolonged running has been shown to be associated with improvement in health, fitness as well as performance (Fields et al., 2010; Lee et al., 2014). Studies have shown decreased joint pain among runners with higher number of marathon participation and lower prevalence of arthritis among active long-distance runners who have completed at least 5 marathons (Ponzio et al., 2018). The increase in participation in prolonged running events over the recent years has also been associated with an increase in RRIs (Winter et al., 2016).The nature of long-distance running places repetitive strain on the hip and knee joints which leads to fatigue and biomechanical changes and potential injuries (Girard et al., 2013). These joints are subjected to a load of approximately 5 and 8 times the body weight while running (Miller et al., 2014; van den Bogert et al., 1999). Runners need to organize their body's physiological and neuromuscular response effectively to the demands of prolonged running (Riccio, 1993). The body undergoes various adaptations to cope with the stress placed upon it. These adaptations include changes in cardiovascular function, muscle activation patterns, and joint mechanics among others.

Novice and injured runners lack the functional variability needed to adapt to these demands (Heiderscheit et al., 2002; Miller et al., 2008). Functional variability refers to the ability of the body to adjust its movement patterns and responses to different conditions. In other words, it is the body's ability to adapt and change its movements in response to various demands. Similarly, injured runners may also lack functional variability due to the restrictions or limitations caused by their injuries. These limitations can affect their ability to effectively respond to the demands of running, increasing the risk of further injury or prolonging the recovery process. Further research is needed to understand the relation between environmental demands, body’s response and development of injuries among long-distance runners. There is still much to learn about how the body adapts and responds to the specific demands of prolonged running. By studying these relationships, researchers can gain insights into the causes of damage to the body during running and develop strategies to prevent or mitigate these injuries.

Fatigue while prolonged running involves neuromuscular exhaustion that can result in biomechanical changes in the lower limbs (Luo et al., 2019). Some of the common changes that may occur include decreased stride length, increased step width, altered foot strike pattern, decreased knee flexion, and increased vertical oscillation (Clermont et al., 2019; Diss & Parmar, 2021). It's important to note that these biomechanical changes vary among individuals and can also depend on factors such as running experience, fitness level, and running technique. Nonetheless, fatigue-induced alterations in lower limb biomechanics can impact running efficiency and potentially increase the risk of injury if not properly managed (Hamill et al., 2012). Studies have shown prolonged walking induces increased ankle joint asymmetry and variability among elderly (Wong et al., 2020). The adverse effect of asymmetry is obvious but not for variability. Increased variability in ankle joint movement during prolonged walking among the elderly can have both positive and negative effects. The potential effects of increased variability can be adaptability. Increased variability can reflect a greater range of movement patterns and adaptability in the ankle joint. This adaptability can be beneficial for the elderly, as it allows them to adjust their gait and movement strategies to accommodate various environmental challenges or changes in walking surfaces. Greater variability can help reduce the risk of falls and improve overall stability. It can distribute the load and stress on the ankle joint more evenly. This can help prevent excessive strain on specific structures, such as tendons or ligaments, reducing the risk of overuse injuries or joint degeneration. Variability in ankle joint movement during walking can also contribute to energy efficiency. It allows the body to find optimal movement patterns that minimize energy expenditure while maintaining stability and balance. This can be particularly important for elderly individuals who may have decreased energy reserves. Increased variability can serve as a compensatory mechanism for age-related decline in muscle strength or joint flexibility. It allows the body to utilize alternative movement strategies to overcome physical limitations and maintain functional mobility. However, it's important to note that excessive or uncontrolled variability can have negative effects, especially if it leads to inefficient or unstable movement patterns. Excessive variability may indicate underlying neuromuscular or balance impairments, which can increase the risk of falls and injuries. One study has suggested that fatigue induced by prolonged running can alter the symmetry of lower limb movements while running that can increase the risk of injury (Gao et al., 2020).

Researchers have also explored the effects of prolonged running on gait parameters using nonlinear fluctuations like long-range correlations (Jordan et al., 2006, 2007a, 2007b). Trained runners have been shown to have lower correlations compared to novice runners (Nakayama et al., 2010). A lower correlation suggests that the fluctuations in gait parameters among trained runners exhibit less persistence or self-similarity compared to novice runners. In other words, the gait patterns of trained runners may exhibit more irregularity or randomness, and the fluctuations in their gait parameters may not exhibit as strong or long-lasting dependencies as those of novice runners. This finding implies that with training and increased running experience, the gait patterns of runners become more variable and less predictable in terms of long-range correlations. Trained runners may have developed more refined and efficient movement patterns, leading to a wider range of gait parameter fluctuations that do not conform to persistent patterns observed in novice runners. The lower correlation observed in trained runners could reflect the adaptability and individuality of their gait patterns. As runners gain experience and refine their technique, they may develop personalized movement strategies that result in increased variability and reduced long-range correlations. It is important to note that lower correlation does not necessarily indicate a negative or detrimental effect. Instead, it suggests a different organization of gait patterns in trained runners, highlighting the complexity and individual characteristics of their movement. Further research is needed to fully understand the implications of lower correlation in gait parameters and its relationship to running performance and injury risk.

Differences in body response among novice and injured runners as compared to healthy and experienced counterparts makes it further important to understand how they are exposed to higher risk of injury (Winter et al., 2016). The effects of fatigue and running surface on gait motor control while prolonged running in the natural environments and their role in development of injuries have not been extensively studied (Brahms et al., 2022; Meardon et al., 2011). There are several potential mechanisms that may lead to adverse effects including neuromuscular fatigue, altered biomechanics (changes in joint angles, muscle activation patterns, and ground reaction forces), decreased proprioception, and impaired shock absorption. It is important to note that the specific mechanisms leading to adverse effects of fatigue and running surface on gait motor control and injury development in natural environments may vary among individuals and depend on various factors such as running experience, fitness level, and individual biomechanics. Further research is needed to comprehensively understand these mechanisms and their role in injury prevention and performance optimization in natural running settings.

Although long-distance running has various psychological and physiological benefits, including improved cardiovascular fitness, muscular endurance, bone health, metabolism, weight management, and mental well-being, (Tonoli et al., 2010), it carries a higher risk of developing RRIs. Multicausal pain in legs associated with bone, muscles, tendons or vascular structures has been reported among long-distance runners, (Gallo et al., 2012) but biomechanical or anatomical cause has not been identified yet. The presence of such multicausal pain in the legs among long-distance runners suggests the possibility of repetitive loading injuries. Repetitive loading injuries occur when tissues are subjected to repetitive stress or loading without adequate time for recovery and adaptation. Risk factors identified for development of these injuries include a history of previous injury, irregular menstruation or amenorrhea in female runners, and a cumulative weekly mileage of over 40 miles (Hulme et al., 2017; Lopes et al., 2011; Myburgh et al., 1990). However, limitations of the existing studies affect their credibility and ability to provide definitive conclusions. More high-quality research about repetitive loading injuries among long-distance runners with rigorous study designs and larger sample sizes is needed to provide more conclusive evidence on the association between risk factors and injury development. This can be done by using the preexisting epidemiological concepts to understand and design future studies investigating development of injuries among long-distance runners (Finch & Cook, 2014). Understanding the risk factors and developing interventions for injury prevention is crucial for the long-term health and safety of long-distance runners. This shall not only help to identify and manage these injuries at an early stage but also potentially prevent their development.

2.3 Running Variability

Running variability refers to the differences or fluctuations in various aspects of running gait, such as speed, stride length, cadence, heart rate, and biomechanical factors (Mason et al., 2023). Stride, which is the interval between two successive heel strikes of the same foot, is the fundamental unit of running gait, which consists of swing and stance phases (Dugan & Bhat, 2005). Differences between strides, known as SSV, have been observed in walking and running. Initially, these differences were considered as noise and were believed to be detrimental to performance and injury prevention. Researchers now understand that SSV is not solely a sign of dysfunction or impairment but is a functional characteristic of the sensorimotor system (Bartlett et al., 2007; Davids et al., 2003a).The human body and the sensorimotor system are inherently adaptable and constantly adjust to varying environmental and internal conditions during running. These differences occur due to the many degrees of freedom of the biological system (Davids et al., 2003b; Newell & Vaillancourt, 2001). SSV reflects this adaptability, allowing the body to respond to changes in terrain, speed, fatigue, and other factors. It enables the system to optimize movement patterns for efficiency and stability. It contributes to maintaining dynamic stability during running. Small variations in stride length, cadence, and foot strike allow the body to distribute forces more evenly, reducing the risk of overloading specific structures. This adaptability helps in adapting to uneven surfaces, unexpected obstacles, or sudden changes in running conditions. It plays a role in movement optimization. It allows the sensorimotor system to explore different movement strategies and find the most efficient and comfortable pattern for an individual. This adaptability can lead to improved running economy and performance. Recognizing SSV as functional has implications for rehabilitation and training. Monitoring running variability can provide insights into the efficiency, performance, and potential risk of injury for a runner (Möhler et al., 2022). This new perspective has led to a shift from considering variability as purely detrimental to recognizing its functional significance in human locomotion.

SSV while running can have both beneficial and adverse effects. It allows for efficient adaptation to changes in terrain, promotes joint flexibility, and helps distribute impact forces throughout the body, potentially reducing the risk of overuse injuries. Some studies suggest that variability in stride length and cadence is associated with improved running economy and performance (Schubert et al., 2014). However, excessive variability can have adverse effects. It may indicate poor running form, such as inefficient mechanics, asymmetries, or muscle imbalances. This can increase the injury risk, particularly in lower limbs. The "variability-RRIs hypothesis" that suggests that decrease in SSV may increase the risk of RRIs has been proposed (Wheat, 2005). Excessive variability can also lead to inefficient energy expenditure, as the body is constantly adapting to different movement patterns.

The clinical significance of SSV lies in its potential as a diagnostic and monitoring tool in certain populations like older adults, individuals with neurological conditions or athletes where it can provide valuable information regarding gait abnormalities, fall risk, rehabilitation progress, and athletic performance. Among runners, monitoring SSV following an injury or surgery, during gait rehabilitation can provide valuable insights. It can help track progress, assess functional improvement, and guide the effectiveness of interventions. Understanding SSV can help optimize training programs in sports and athletic performance. Analyzing variability metrics can provide insights into the efficiency and economy of movement, identifying areas for improvement and guiding training strategies. SSV can be categorized based on its occurrence during task execution or outcome. Studying variability at both levels helps in understanding the development of RRIs.

2.3.1 Types of Variability: Task Execution and Outcome Variability

a. Task Execution Variability: Variability at the task execution level refers to the variability in stride parameters, such as cadence, stride length, ground contact time, vertical oscillation, and other kinetic or kinematic factors, during the act of running. It provides insights into the efficiency, consistency, and mechanics of a runner's performance. Studies have shown that coordination variability, which is a measure of lower limb coordination, corresponds to running variability at the movement execution level (Jordan et al., 2006; Mo & Chow, 2019). Coordination variability provides flexibility to allow adaptation to environmental constraints and impact absorption while running. It has been shown to be linked with the development of RRIs (Bartlett et al., 2007; Hamill et al., 2012). Such linkage is a topic of ongoing research. While the relationship is complex and not fully understood, there are several proposed mechanisms that may explain this association.

Coordination variability reflects the ability of the sensorimotor system to adapt and adjust movement patterns to changing conditions. However, excessive variability or instability in movement control may lead to suboptimal mechanics, increasing the risk of tissue overloading and injury. For example, if a runner exhibits high variability in foot strike patterns, it may result in inconsistent loading patterns on the lower limbs, potentially increasing the risk of stress fractures or other repetitive strain injuries. Efficient load distribution is crucial for injury prevention during running. Coordination variability may impact how forces are distributed throughout the body during each stride. Excessive variability in joint angles or muscle activation patterns may lead to uneven loading, placing excessive stress on specific structures and increasing injury risk. Running with excessive coordination variability may result in inefficient energy expenditure. When movement patterns are highly variable, the body may be expending more energy to control and stabilize movements. Increased energy expenditure may contribute to fatigue and decrease the body's ability to absorb impact effectively, potentially increasing injury risk.

While some studies have found associations between higher variability and injury occurrence, others have found conflicting results or no clear relationship. Studies have reported lower coordination variability among injured runners as compared to healthy runners (Hamill et al., 1999; Heiderscheit, 2000; Heiderscheit et al., 2002; Seay et al., 2011). Decreased coordination variability shows reduction in the degrees of freedom in the sensorimotor system, leading to a "loss of complexity" and increased cumulative load on specific structures, potentially resulting in injury (Bartlett et al., 2007; Bertelsen et al., 2017; Lipsitz, 2002). However, there are conflicting findings as some studies reported either no differences or higher coordination variability in different couplings among healthy runners and those with specific injuries like iliotibial band syndrome and low back pain (Hafer et al., 2017; Miller et al., 2008; Seay et al., 2014). These discrepancies may be due to methodological differences in quantifying coordination variability and the influence of various factors such as gender, age, structural differences (such as quadriceps angle), running experience, fatigue, and cadence on lower limb coordination during running (Boyer et al., 2017; Brown et al., 2016; Dierks et al., 2010; Floría et al., 2018; Hafer et al., 2017; Hafer et al., 2016; Heiderscheit et al., 1999; Miller et al., 2008). Additionally, individual factors, such as training history, biomechanics, and conditioning, may interact with coordination variability to influence injury risk. Further research is needed to better understand the specific mechanisms and determine how coordination variability can be effectively managed to reduce injury risk in runners.

Coordination variability has been commonly quantified using continuous relative phase (CRP) and modified vector coding techniques. For the CRP method, a phase angle is obtained by plotting angular displacement and angular velocity of the coupling joints or segments while in modified vector coding technique, a coupling angle is obtained only through the angular displacement of the coupling joints or segments (Van Emmerik et al., 2004).

b. Task Outcome Variability: Variability at the task outcome level focuses on variations in performance indicators related to stride outcomes. It assesses the reliability and consistency of performance outcomes across different running conditions and sessions. Specifically, stride time variability is often used as a measure of outcome variability (Jordan et al., 2006; Mo & Chow, 2019). Research has shown that stride time variability exhibits predictable patterns with fractal-like long-range correlations, quantified by the scaling exponent alpha (Jordan et al., 2006, 2007a). However, there are conflicting reports regarding how these correlations are affected among injured runners. Some studies have found larger alpha values among injured runners, while others have reported smaller alpha values compared to healthy runners (Mann, Malisoux, Nührenbörger, et al., 2015; Meardon et al., 2011). These discrepancies may be attributed to differences in experimental conditions like sample characteristics, injury types, measurement techniques, running conditions, study design and statistical analysis. Despite extensive research, the relationship between task outcome variability and development of RRIs remains unclear and requires further investigation.

2.3.2 Optimal Gait Variability

Runners with lower limb injuries have been found to exhibit lower stride variability, while higher stride variability has been shown to negatively impact performance and energy efficiency while running (Belli et al., 1995; Hamill et al., 1999; Lilley et al., 2018; Miller et al., 2008; Morgan et al., 1989). Both higher and lower variability has been sown to be detrimental to motor tasks like running (Davids et al., 2003a; Hamill et al., 2012; Srinivasan et al., 2015). It's important to strike a balance. Moderate SSV can be beneficial for performance and injury prevention. However, extreme variability or consistent deviations from an individual's natural stride pattern should be addressed through proper running form analysis, strength training, and potential gait corrections to mitigate potential adverse effects. Therefore, an "optimal window" of variability should exist where individuals demonstrate greater flexibility while executing movements for desired outcomes (Hamill et al., 2012; Hanley & Tucker, 2018). This optimal window allows for efficient movement control and adaptation to changing conditions, reducing the risk of injury. However, the specific range and characteristics of this optimal window are still under investigation and may vary across individuals. The understanding of the influence of factors like experience, surface, duration and progressive fatigue while running on stride variability is still limited, it is challenging to identify this “optimal window”. Most studies examining stride changes during running do not account for individual performance levels and running experience. More research is needed, particularly in natural running environments, to comprehensively identify the factors contributing to running variability among skilled runners. Understanding the clinical meaning is important for quantifying the variability.

2.3.3 Factors Affecting Gait Variability

Gait variability is influenced by various factors. Understanding these factors is important for assessing gait stability and identifying potential risks, such as falling or developing neuromotor conditions and sports injuries. It is important to consider that these factors are interconnected and can influence each other. For example, fatigue can result from high training load, which may also contribute to an increased risk of injury. Furthermore, factors affecting variability can vary among individuals, and what affects one person's gait variability may not have the same impact on another person. The following section explores the factors that affect running gait variability, including fatigue, injury, age, and pathological neuromotor conditions.

a. Fatigue: Fatigue is a common factor that can affect gait variability during running. It can lead to changes in muscle activation patterns and coordination. As muscles become tired, they may not fire with the same timing or intensity, which can result in alterations in running gait. One common alteration in running gait due to muscle fatigue is a decrease in stride length. As the muscles responsible for propelling the body forward, such as the hip extensors, become tired, they may not contract as forcefully, resulting in a shorter stride. This can lead to an increased number of steps taken to cover the same distance, potentially increasing joint loading. Another alteration is a decrease in joint stability. Fatigued muscles may not provide adequate support and stabilization to the joints, particularly the ankles and knees. This can result in a reduced ability to absorb shock and control the movement of the joints during running, leading to increased stress on the joints and potentially increasing joint loading. Additionally, muscle fatigue can also lead to compensatory movements or muscle imbalances. When certain muscles are fatigued, other muscles may have to work harder to compensate, which can disrupt the normal movement patterns. This can put additional stress on certain joints and increase the risk of joint loading. Overall, alterations in running gait due to muscle fatigue can result in increased joint loading. The increased stress on the joints can potentially lead to overuse injuries, such as patellofemoral pain syndrome, iliotibial band syndrome, or stress fractures. It is important for runners to be aware of these effects and manage their training and recovery appropriately to minimize the risk of joint loading and related injuries. In turn, these changes can increase variability as the body tries to compensate for the fatigue and maintain a steady stride. Some studies have reported that fatigue decreases long-range correlations between strides, making the SSV less predictable (Meardon et al., 2011; Mo & Chow, 2018b). This suggests that fatigue can influence the neuromuscular output during running. However, conflicting results have been found, with some studies indicating no differences in some gait spatiotemporal parameters and strike index between fatigued and non-fatigued conditions (Fuller et al., 2017; Mann, Malisoux, Urhausen, et al., 2015). As speed is a combination of stride length and stride interval, these inconsistencies may be due to speed fluctuations (Jordan et al., 2006). Additionally, the fatigue effect on gait variability differs based on running experience, with novice runners showing an increase in vertical center-of-mass displacement after fatigue as compared to experienced runners (Maas et al., 2018).

The bodily responses to fatigue can vary among individuals, which should be considered when studying running kinematics after fatigue. However, these individual responses are often overlooked in group analysis (Zandbergen et al., 2022). Although acute and chronic fatigue resulting from intense training have been extensively studied (Fuller et al., 2017; Mann, Malisoux, Urhausen, et al., 2015), their impact on gait variability remains unclear. In these studies, fatigue is mostly induced by having participants run a relatively short distance on a treadmill in a laboratory setting, and fewer studies have examined the effects of fatigue during long-distance running in a natural environment. There is a prediction that the interaction between fatigue, experience, speed, health, training of runners, as well as running surface could affect complexity of stride time (Jordan et al., 2007a; Lindsay et al., 2014; Mann, Malisoux, Nührenbörger, et al., 2015; Meardon et al., 2011; Nakayama et al., 2010). However, more research is needed to confirm such relationship. Future studies should also consider large sample size to address the challenges of controlling for multiple factors simultaneously. By combining different research approaches, including cross-sectional studies, prospective cohort studies, and meta-analyses, researchers can gather a more comprehensive understanding of the complex interactions between fatigue and various factors on stride interval complexity. These studies can help confirm the relationships and provide valuable insights into optimizing running performance and reducing the risk of injuries.

b. Injury: Injury is another factor that can affect gait variability. When an injury occurs, it can disrupt the normal biomechanics of running. Studies have shown that individuals with RRIs, such as patellofemoral pain syndrome, iliotibial tibial band syndrome, and chronic ankle instability, have lower SSV compared to healthy individuals (Hamill et al., 1999; Lilley et al., 2018; Miller et al., 2008). Injured runners have been shown to have smaller scaling exponent alpha depicting altered structure of their stride time dynamics (Meardon et al., 2011). On the contrary, Mann et al. found no differences in stride time variation while comparing healthy and injured runners (Mann, Malisoux, Nührenbörger, et al., 2015). The "variability-RRIs hypothesis" suggests that risk of injury may increase with a decrease in stride variability while running (Wheat, 2005). Furthermore, literature suggests that injured runners have lesser coordination variability that is required to adapt to change in environments (Heiderscheit et al., 2002; Miller et al., 2008). While some studies have suggested a connection between stride variability and development of RRIs (Hamill et al., 1999; Lilley et al., 2018; Miller et al., 2008), there is still limited understanding of how stride variability varies with duration and surface of running and its impact on injury risk.

c. Age: Age is a significant factor that affects gait variability. Age-related differences in muscle strength, flexibility, and coordination can impact running gait variability. Gait dynamics vary among different age groups, with young children and older adults showing distinct patterns. In young children, stride-to-stride fluctuations have been shown to be less correlated as gait control and posture are not fully developed, leading to unsteady motion (Hausdorff et al., 1999). Although the neuromuscular system matures by the age of 3 years, changes in walking dynamics after this age are unknown (Hausdorff, 2007). Neurophysiological investigations among healthy children suggest that different aspects of gait dynamics mature at different ages. Gait of a child looks somewhat similar to that of adults by the age of 7 years (Hausdorff et al., 1999; Hausdorff, 2007). As we age, muscle strength and flexibility may decline, leading to changes in running biomechanics and increased variability. Understanding SSV, which has been shown to fluctuate among healthy older adults, is essential, as they have a heightened risk of falling and greater dependence for mobility (Hausdorff, 2007; Rubenstein & Josephson, 2002; Verghese et al., 2006). Studies have shown that complexity of gait dynamics changes with healthy aging. Older adults who fall frequently exhibited higher variability in swing and stride times but the magnitude of fluctuations in step timings and the stride length remained unchanged (Hausdorff, Edelberg, et al., 1997; Hausdorff, Mitchell, et al., 1997; Owings & Grabiner, 2004a). This indicates the potential need to identify the risk of fall using quantitative assessments of gait instability that are based on gait dynamics (Hausdorff, 2007; Hausdorff, Edelberg, et al., 1997). SSV assessments are considered crucial for studying instability of gait (Bartlett et al., 2007; Hausdorff, Rios, et al., 2001). Higher variability in stride time and length has been linked to an increased risk of falling, while lower variability is associated with a safer gait (Beauchet et al., 2005; Dubost et al., 2006; Hausdorff, Rios, et al., 2001; Maki, 1997). However, both low and high SSV has been observed in both fallers and non-fallers, making its interpretation challenging (Bilney et al., 2003; Brach et al., 2005). Research indicates that high SSV is an indicator of gait instability (Beauchet et al., 2007). It is important for researchers to exercise caution when evaluating gait variability and stability since both high and low variability can indicate gait stability among healthy adults (Beauchet, Allali, et al., 2009). When interpreting age-related gait variability, it is essential to consider a natural decrease in walking or running speed. Although some studies have examined gait variability among older populations (Helbostad & Moe-Nilssen, 2003; Kang & Dingwell, 2008; Owings & Grabiner, 2004b; Woledge et al., 2005), further research and literature reviews are necessary to deepen our understanding in this area.

d. Cognitive factors and pathological neuromotor conditions: Attentional focus, distraction, or cognitive load can influence running gait variability. When a runner is distracted or has a high cognitive load, their attention to maintaining a consistent gait may decrease, leading to increased variability. Pathological neuromotor conditions can significantly affect gait variability. In fact, gait variability has been used to quantify the magnitude as well as dynamics of altered walking features in neurological syndromes like Alzheimer's and Parkinson's diseases (Hausdorff, 2005; Herman et al., 2005; Schaafsma et al., 2003; Sheridan et al., 2003). Gait variability measures, such as symmetry and ratio indices, have been used to quantify asymmetric behavior in patients with motor diseases (Błażkiewicz et al., 2014; Iosa et al., 2012; Mileti et al., 2016; Prosser et al., 2010; Rossi et al., 2013). The relationship between neurodegenerative diseases such as multiple sclerosis, Alzheimer's, and Parkinson's disease and higher stride time variability has also been documented. However, the relationship between stride time variability and slower walking speeds is controversial and has been less examined (Beauchet, Annweiler, et al., 2009; Jordan et al., 2007b; Owings & Grabiner, 2004b). Additionally, mild cognitive impairment, a transitional state between normal aging and Alzheimer's disease, has been linked to variable gait patterns (Byun et al., 2018). Old adults with mild cognitive impairment have been shown to maintain their cognitive performance at the cost of gait variability and dynamics under dual-task conditions (Hawkins, 2019; Hawkins et al., 2019), and this can be used to identify neuromotor diseases at early stages.

e. Biomechanics and anatomical factors: Individual variations in musculoskeletal anatomy can impact running gait. Leg length discrepancies or muscle imbalances can lead to asymmetrical movement patterns and increase gait variability. These factors can affect the alignment and coordination of the body during running. The type and condition of footwear can also impact gait variability. Different types of shoes can alter the foot strike pattern and distribution of forces, affecting the overall running gait. Ill-fitting or worn-out shoes can also lead to instability, as they may not provide adequate support or cushioning. Running on uneven or slippery surfaces requires the body to constantly adapt to maintain balance.

This adaptation can increase gait variability as the body adjusts to the changing terrain. In contrast, running on a stable and even surface typically reduces gait variability. Studies have also shown differences in SSV while short duration running over treadmill and track surfaces inside laboratory (Mo & Chow, 2018b; Mo & Chow, 2019). Sudden increases in training volume or intensity can lead to fatigue and changes in running gait. The body may not have enough time to adapt to the increased demands, resulting in increased variability. Gradually increasing training load allows the body to adjust and reduce gait variability over time. Similarly, poor running technique or form can also contribute to gait variability. Over striding, where the foot lands too far in front of the body, can increase braking forces and instability. Excessive lateral movement or asymmetrical arm swing can also disrupt the natural rhythm of running and increase gait variability.

f. Speed: Running speed is a critical factor and running variability may not be constant at different running speeds. As running speed increases or decreases, there are corresponding changes in gait mechanics and movement patterns, which can impact gait variability. At faster running speeds, individuals tend to take longer strides, have shorter ground contact times, and increase their cadence. These changes in gait mechanics can lead to lower SSV as the movements become more synchronized and consistent. On the other hand, at slower running speeds, individuals may take shorter strides, have longer ground contact times, and lower cadence. These alterations in gait mechanics may result in higher SSV as the movements become more variable and less consistent. Therefore, running speed plays a critical role in gait variability. It is important to consider speed as a factor when studying or assessing running gait variability, as the relationship between speed and variability can provide valuable insights into an individual's gait mechanics and performance.

2.3.4 Methods Used to Measure Gait Variability

Gait variability is studied using data acquisition using various methods that can provide valuable insights into gait patterns and can be used in the assessment of various neurological, musculoskeletal, and aging-related conditions. These methods can be categorized into qualitative measures, instrumented methods, wearable sensors, and others. Their details are provided in the following section.

a. Qualitative Measures: The Modified Gait Abnormality Rating Scale, a qualitative measure, provides an easy and cheaper alternative for measuring gait variability. It demonstrated good to excellent interrater and intrarater reliability (0.878 and 0.989, respectively) against instrumented walkways (Huang et al., 2008; Vandenberg et al., 2015). It has been used to identify gait abnormalities associated with the risk of fall among older adults as well as to assess and quantify abnormal gait patterns in individuals with neurological conditions like stroke, parkinson's disease, and multiple sclerosis (VanSwearingen et al., 1996). The scale was developed by Morris and colleagues in 2002 as an adaptation of the Gait Abnormality Rating Scale. It comprises of seven items that assess different aspects of gait abnormalities, including posture, arm swing, step length, step symmetry, trunk movement, and walking speed to depict variability, arrhythmicity and inconsistency in stepping (VanSwearingen et al., 1998). Each item is rated on a four-point scale, with higher scores indicating more severe gait abnormalities. The total score ranges from 0 to 18, with higher scores indicating more severe gait abnormalities. It is often used in research studies and clinical practice to monitor changes in gait patterns over time and evaluate the effectiveness of interventions. It is important to note that the Modified Gait Abnormality Rating Scale is just one of many tools available for assessing gait abnormalities, and needs to be complemented with a comprehensive clinical evaluation by a healthcare professional.

b. Instrumented Methods: Most of the previous studies on gait variability have quantified it using advanced techniques and technology (Brach et al., 2005; Hausdorff et al., 2003). Foot switches and instrumented walkways are commonly used but are relatively expensive and labor-intensive, thus impractical for daily clinical practice (Brach et al., 2001; Hausdorff, Rios, et al., 2001). Instrumented walkways involve the use of pressure-sensitive walkways embedded with force sensors or electronic mats. They can measure various gait parameters, such as step length, step time, and stride time, which can be used to assess gait variability. Video-based motion capture systems are considered the gold standard but require expensive equipment and large indoor laboratories (Dahl et al., 2020; Gholami et al., 2020; Robert-Lachaine et al., 2020). However, it has a limited coverage of few strides and cannot provide optimal SSV measurements in external environment in sports settings or activities of daily living (Hausdorff, 2005).

c. Wearable Sensors: Wearable sensors, particularly IMUs, have emerged as a practical and cost-effective alternative to traditional laboratory-based motion capture systems for capturing real-time data on running patterns outdoors (Benson et al., 2018). IMUs consist of gyroscopes, accelerometers, magnetometers, and barometers, enabling the measurement of joint angles, acceleration, and angular velocity (Chow et al., 2021; Horenstein et al., 2020). These sensors are widely used during sports events and can be attached to various body segments for comprehensive motion analysis (Chow et al., 2021; Gholami et al., 2020). Studies have shown that IMUs offer higher reliability and validity compared to marker-based motion analysis systems when estimating joint kinematics during walking and running (Brice et al., 2020; Clemente et al., 2021; Hafer et al., 2020; Senanayake et al., 2021). They have demonstrated effectiveness in predicting the kinematics of the same as well as opposite limb, i.e. cross-leg prediction, using a single sensor unit (Chow et al., 2022; Chow et al., 2021). Wearable sensors have been extensively utilized in research and clinical settings to capture kinematic data across various activities. The validity and reliability of these sensors in determining typical kinematic data may vary based on factors such as sensor type, placement, and the specific activity being measured. While several studies have investigated the validity and reliability of wearable sensors in capturing kinematic data, it is essential to consider the limitations and specific conditions under which the assessments were conducted.

The accuracy of IMUs relative to gold standard video-based motion analysis systems can vary depending on factors such as sensor quality, calibration procedures, and data processing algorithms (Favre et al., 2008). Generally, IMUs can provide accurate measurements of motion parameters like acceleration, angular velocity, and orientation, but they may not always match the precision of video-based systems. Studies have shown that IMUs can achieve accuracy levels within a few degrees or millimeters of the gold standard, making them a reliable alternative for motion analysis in various applications (Washabaugh et al., 2017). In terms of reliability, IMUs are considered highly reliable for capturing motion data, with sensors designed to provide consistent and repeatable measurements. However, factors such as sensor drift, noise, and calibration errors can impact the reliability of data collected by an IMU. Regular calibration and sensor fusion techniques can be employed to minimize errors and ensure accurate measurements. Intra-device reliability of IMUs is generally high, with consistent measurements across repeated trials, although inter-device reliability can vary based on sensor quality and calibration procedures. Validity studies have demonstrated good agreement between IMUs and video-based systems for capturing joint angles and gait parameters, with reported correlations ranging from 0.80 to 0.95 (Guignard et al., 2021). Overall, IMUs are considered a valid tool for motion analysis in various applications, offering accurate and reliable motion data collection.

d. Others: Other tools, such as goniometers, force plate mounted treadmills, etc., are available for measuring SSV under various ambulatory conditions having their own advantages and disadvantages (Bonato, 2005; Hausdorff, 2007; Malatesta et al., 2003; Menz et al., 2003; Moe-Nilssen & Helbostad, 2005; Owings & Grabiner, 2003). Global positioning systems have shown promise in monitoring spatial and temporal gait measurements but are not yet available for clinical use (Terrier & Schutz, 2003; Terrier et al., 2005). Overall, advancements in measurement techniques continue to improve our understanding of gait variability in outdoor settings.

2.3.5 Parameters Used to Quantify Gait Variability

There are several parameters used to quantify gait variability in human movement analysis, which capture different aspects of stride-to-stride fluctuations and irregularities in gait patterns. These parameters can be calculated from various gait parameters, such as step length, step time, stride time, or spatial and temporal characteristics of gait. The choice of parameters used to measure gait variability depends on the research objectives, available instruments, and methods employed. These parameters can be used individually or in combination to assess various aspects of gait variability and provide insights into the underlying patterns and dynamics of walking. These parameters can be calculated for various gait parameters and can be used to quantify gait variability in different populations, such as healthy individuals, older adults, or individuals with gait disorders. Different parameters have been reported to have various effects on gait variability, leading to a lack of consensus on the optimal methods and parameters for analyzing gait variability (Lord et al., 2011). The measurement of gait variability in running has traditionally focused on analyzing the properties of a typical stride (Hausdorff, 2007).

There are several traditional methods and non-linear analytical methods that are used to extract meaningful information and quantify various aspects of gait variability-

a. Standard deviation (SD): SD represents the absolute dispersion or spread of a gait parameter, such as step width or swing phase duration, around its mean value. It provides a measure of the average magnitude of deviations from the mean. It is one of most common measures used to examine the overall variation in gait during ambulation.

b. Coefficient of variation (CV): CV that measures the relative variability of a gait parameter and is calculated as the ratio of the SD to the mean value of the parameter and expressed as a percentage. It quantifies the relative variability of a gait parameter, such as step length or stride time, by considering the spread of the data relative to its mean. CV is often preferred due to its higher sensitivity (Böhm & Döderlein, 2012; Taborri et al., 2014). A greater value of CV indicates a higher level of dispersion around the mean. Gait cycle variability can be determined by calculating the stride time variability, which is the CV of the stride time series of all participants (Hausdorff, 2007; Hausdorff, Rios, et al., 2001; Maki, 1997). Similarly, variability of other gait measures, such as swing time, etc. can be calculated. It is one of most common measures used to examine the overall variation in gait during ambulation.

c. Range: Range is the difference between the maximum and minimum values of a gait parameter that provides an estimate of the overall variability in the parameter.

d. Interquartile range: Interquartile range measures the spread of a gait parameter. It is calculated as the difference between the 75th percentile and the 25th percentiles of a gait parameter distribution. It measures the spread of the middle 50% of the data and is less sensitive to outliers compared to the range.

Linear measures, such as CV, can assess the magnitude of variability but cannot detect deviations from regular and stable patterns. Non-linear analytical approaches, drawing from chaos theory, are utilized to examine the temporal aspects of variability (Harbourne & Stergiou, 2009; Stergiou & Decker, 2011). These methods offer insights into complexity, non-linear relationships between variables, and capture more intricate and time-varying dynamics (Strongman & Morrison, 2020). They provide a more accurate representation of system interactions. Non-linear methods analyze systems as dynamical systems, focusing on attractors, stability, and transitions within the system. They help identify patterns, attractor states, and potential bifurcations or phase transitions in system dynamics. These methods are closely associated with chaos theory and complexity theory. They can reveal deterministic chaos, which refers to seemingly random and unpredictable behavior emerging from deterministic systems. Non-linear methods aid in identifying patterns within chaotic systems, understanding underlying mechanisms, and potentially predicting or controlling system behavior. It is important to note that interpreting and comparing the coefficients produced by non-linear methods can be challenging due to their complexity. However, these methods offer valuable applications and insights in various fields, particularly in investigating injury rehabilitation and movement variability. Some of the commonly used non-linear methods have been described in the following section.

a. Fractal Measures: Fractal analysis, such as DFA, estimates the scaling properties or long-range correlations in gait patterns or the complexity or irregularity of a gait parameter. It provides an estimate of the space-filling capacity of a time series and assesses the persistence or self-similarity of fluctuations across different time scales. The scaling exponent α of stride time has been calculated using DFA (Jordan et al., 2006, 2007a, 2007b). Value of α between 0.5 and 1.0 indicates a predictable and persistent stride time pattern, while a value less than 0.5 and greater than 1.0 indicates suggests irregular and long-range correlations in strides respectively (Goldberger et al., 2000). Regarding the magnitude of the value, it can have clinical meaning depending on the specific parameter being assessed. However, it is important to consider that the clinical meaning of the magnitude of the value may vary depending on the specific context and the variability parameter being assessed.

It is necessary to consider other factors, such as the individual's functional limitations, symptoms, and overall gait characteristics, to fully interpret the clinical significance of the variability parameter. Stride length affects spatial variability, while stride frequency affects temporal variability, although studies often analyze these variabilities separately without establishing a specific relationship between them (Maruyama & Nagasaki, 1992; Sekiya et al., 1997).

b. Entropy Measures: Entropy-based measures, such as approximate entropy or sample entropy, quantify the complexity or regularity of gait variability. They assess the predictability or irregularity of the time series data. (Costa et al., 2005; Richman & Moorman, 2000). Frequency domain analysis involves transforming gait data into the frequency domain using techniques like Fourier transform or wavelet transform (Dingwell & Cusumano, 2000). It provides information about the distribution of gait variability across different frequency bands and involves transforming the time-domain gait signals into the frequency domain. This allows for the examination of gait variability distribution across different frequency bands (Bruijn et al., 2013). Approximate entropy quantifies the regularity or predictability of a time series by comparing patterns of a given length within the series. Lower values of approximate entropy and sample entropy indicate more regular and predictable patterns, while higher values indicate more irregular and unpredictable patterns. These can detect changes associated with different conditions or interventions. Analyzing gait data in the frequency domain also helps researchers identify rhythmic patterns and oscillatory components of gait, providing information about spectral content and potential abnormalities not apparent in time domain analysis (Delignieres et al., 2006). Entropy measures and DFA are both methods used to analyze gait variability, but they capture different aspects of variability and employ different mathematical approaches. Entropy measures focus on quantifying the complexity or irregularity of a time series by assessing the patterns and predictability, while DFA specifically examines the presence of long-range correlations and self-similarity in the data. The choice of method depends on the specific research or clinical objectives and the characteristics of the variability being analyzed.

c. Lyapunov exponent: The Lyapunov exponent is a measure used in non-linear dynamics to quantify the sensitivity to initial conditions in a dynamical system. It provides information about the rate of divergence or convergence of nearby trajectories in the system (Mehdizadeh, 2018). In the context of gait analysis, the Lyapunov exponent can be used to assess the stability and predictability of gait patterns. It is typically a positive or negative value that represents the average exponential growth or decay rate of nearby trajectories. A positive Lyapunov exponent indicates chaotic behavior, where nearby trajectories diverge exponentially over time. A negative Lyapunov exponent suggests stable behavior, where nearby trajectories converge or stay close to each other. A Lyapunov exponent of zero indicates neutral or marginal stability. The absolute magnitude of the Lyapunov exponent is meaningful, as it provides information about the degree of divergence or convergence in the system. Higher absolute values of the Lyapunov exponent indicate greater sensitivity to initial conditions and more chaotic behavior. Smaller absolute values suggest more stable and predictable dynamics. It is challenging to provide a specific typical value for the Lyapunov exponent, as it can vary depending on the specific system being analyzed. The Lyapunov exponent is influenced by various factors, including the characteristics of the system, the specific gait parameter being analyzed, and the data collection and analysis methods used. Therefore, the typical value of the Lyapunov exponent can vary across studies (Politi, 2013).

d. Poincaré analysis: Poincaré analysis is a method used to assess the variability and patterns in a time series data by examining the points of intersection between the data points and a line of identity (also known as the Poincaré plot or scatter plot) (Goldberger et al., 2002). This analysis reveals short-term and long-term variability in gait and helps identify abnormalities or changes in gait dynamics. Parameters such as the SD of the perpendicular distance to the line of identity (SD1) and the SD along the line of identity (SD2) quantify the shape, orientation, and dispersion of the scatterplot. SD1 reflects short-term variability, while SD2 reflects long-term variability. Poincaré analysis provides a visual and quantitative assessment of gait variability, enabling the characterization of gait dynamics and the detection of changes associated with different conditions or interventions (Dingwell & Cusumano, 2000). The typical range of Poincaré analysis varies depending on the specific characteristics of the time series being analyzed and the research context. The range can be influenced by factors such as the specific parameter being measured, the duration of the time series, and the physiological or pathological conditions being studied. In general, the Poincaré plot will have an elliptical or cloud-like shape, and its width and length can provide information about different aspects of the time series variability (Kahlon et al., 2023). The typical range of SD1 and SD2 can vary depending on the specific study population and the physiological or pathological conditions being investigated. In healthy individuals, the range for SD1 is generally around 3-10 ms and for SD2 is around 10-50 ms. However, these ranges can differ in various clinical conditions, such as cardiac disorders, neurological conditions, or aging-related changes (Satti et al., 2019).

e. Harmonic Ratio: The harmonic Ratio is a measure of stride regularity and symmetry. It calculates the ratio of the power at the fundamental frequency (stride frequency) to the total power in the frequency spectrum derived from accelerometry or kinematic data.

f. Principal Component Analysis: Principal Component Analysis is a multivariate analysis technique that can be applied to gait variability data to identify underlying patterns or factors contributing to variability. It reduces the dimensionality of the data and identifies the principal components representing the most significant sources of variability.

2.4 Differences Between Treadmill and Over-ground Running

Treadmills are commonly used by coaches and athletes for training and assessing running performance (Bishop et al., 2017; Cappa et al., 2014). Such assessment range from running-related physiological response to perceived effort which for long time have been thought simulate the demands and body response while running over-ground for tracking athletic performance and research purposes (Edwards et al., 2017; Taunton et al., 2003). They offer a controlled indoor environment and require limited space, making them popular for research purposes (Morin & Sève, 2011). However, comparing running on a treadmill to running over-ground has revealed various differences, raising questions about the treadmill's ability to accurately simulate the physiological demands of outdoor running (Miller et al., 2019). The main argument is that running on treadmill cannot provide a reliable surrogate for the physiological demands of the body while running over-ground to truly reflect the natural pattern of gait adopted on outdoor tracks (McCrory et al., 2022).

Running on a treadmill provides a stable and uniform condition but requires more metabolic energy and less impulse and excursion of the center of mass compared to over-ground running. Studies comparing treadmill running to over-ground running have primarily focused on physiological, perceptual, performance, and kinetic measures during short-duration and distance runs. However, these indices alone cannot fully predict musculoskeletal loading and potential injuries. One early study by Pugh found that oxygen uptake was higher while running on an outdoor track compared to a treadmill, possibly due to the lack of air resistance on the treadmill (Maksud et al., 1971; Pugh, 1970). Other physiological variables such as heart rate and blood lactate concentration have also been investigated, but with mixed results. Some studies suggest that running on a treadmill results in lower oxygen consumption and energy cost compared to running over-ground, while others show no significant differences or even higher consumption (Heck et al., 1985; Maksud et al., 1971). Similarly, there is conflicting evidence regarding running performance and endurance on treadmills versus over-ground (Morin & Sève, 2011; Peserico & Machado, 2014).

There have been several studies that have compared biomechanical differences between treadmill and over-ground running. Various differences in the kinetics of walking and running between treadmills and over-ground have been revealed including energy consumption, ground reaction force, and muscle activity (Berryman et al., 2012; Kluitenberg et al., 2012; Martin & Li, 2017; Miller et al., 2008; Riley et al., 2008; Yao et al., 2019). Similarly, studies have also shown differences in joint kinematics between treadmill and over-ground running. Treadmill running has been found to result in shorter stride lengths and higher stride frequencies compared to over-ground running. This is likely due to the limited space and shorter belt length on a treadmill. Over-ground running allows for more freedom of movement, resulting in longer stride lengths. Studies have shown that ground reaction forces differ between treadmill and over-ground running. Treadmill running typically results in lower vertical ground reaction forces compared to over-ground running (Bovalino, 2021). This could be attributed to the lack of wind resistance and the cushioning effect of the treadmill belt, which reduces the impact on the body. Sometimes the vertical ground reaction force may be similar but the anterior-posterior forces differ between the two conditions (Berryman et al., 2012; Firminger et al., 2018; Kluitenberg et al., 2012; Martin & Li, 2017; Riley et al., 2008). During treadmill running, the anterior-posterior ground reaction forces tends to be lower compared to over-ground running. This is primarily because the treadmill belt assists in propelling the runner forward, reducing the need for active push-off. The belt's movement also contributes to a more consistent and controlled forward motion, resulting in lower anterior-posterior forces. In contrast, over-ground running typically involves higher anterior-posterior ground reaction forces compared to treadmill running. This is because, during over-ground running, runners need to generate more force to propel themselves forward without the assistance of a moving belt. The variability in terrain and external factors like wind resistance can also contribute to higher anterior-posterior forces during over-ground running. Overall, these differences can be attributed to external loading conditions and neuromodulation strategies (Sousa et al., 2012, 2013; Sousa & Tavares, 2012). It is important to note that the exact magnitude of the differences in anterior-posterior forces between treadmill and over-ground running can vary depending on several factors, including running speed, incline, and individual characteristics. Additionally, the specific measurement techniques and equipment used in different studies can also lead to variations in reported results. Further research is needed to gain a more comprehensive understanding of the specific differences in anterior-posterior forces and their impact on running performance and injury risk.

Some studies have found differences in joint kinematics between treadmill and over-ground running (Miller et al., 2019). For example, ankle dorsiflexion and knee flexion angles have been reported to be greater during treadmill running compared to over-ground running. These differences may be related to differences in surface properties and the need to adapt to the moving treadmill belt. Muscle activation patterns can also differ between treadmill and over-ground running. Some studies have shown that certain lower limb muscles, such as the gastrocnemius and soleus muscles, exhibit higher activity during over-ground running compared to treadmill running. This could be due to the need for greater propulsive forces during over-ground running. It is important to note that biomechanical differences between treadmill and over-ground running can vary depending on various factors, including running speed, incline, and individual characteristics. Further research is needed to fully understand the biomechanical implications of prolonged running on treadmill versus over-ground running.

Conflicting evidence regarding physiological variables, running performance, and endurance could be due to the methodological variations among studies comparing treadmill and over-ground running. Some examples of methodological variations that could impact the results are different treadmill settings, such as speed, incline, or mode, protocol duration, sample characteristics, such as age, sex, fitness level, or running experience, measurement techniques, environmental conditions, such as temperature, humidity, or altitude, and statistical analyses. It is important to consider these variations when interpreting the results and to consider the specific details of each study when comparing findings. The direction and magnitude of differences in physiological parameters and perceived effort between treadmill and over-ground running can be influenced by speed differences (Miller et al., 2019). Generally, runners prefer over-ground running because it allows for greater control over speed and reduces the risk of falling compared to running on a treadmill (Taunton et al., 2003). Treadmill running does provide a more stable speed compared to over-ground running. On a treadmill, the speed is set by the user and remains consistent unless adjusted. In contrast, over-ground running can be influenced by external factors such as terrain, wind, and fatigue, which can lead to variations in speed. Over-ground running is generally considered to have a lower risk compared to treadmill running. On a treadmill, there is a risk of tripping or losing balance due to the moving belt. However, it is important to note that the risk of falling can also be influenced by individual factors, running technique, and environmental conditions.

Differences between treadmill and over-ground running can lead to poor rehabilitation outcomes, incorrect training workload prescriptions, and challenges for competitive runners (Miller et al., 2019). While these differences do not diminish the clinical utility of treadmill training, these findings suggest that running on a treadmill may result in a more consistent gait pattern compared to over-ground running. Rehabilitation programs often include running as part of the recovery process for various injuries. However, if the rehabilitation is primarily conducted on a treadmill, there may be limitations in replicating the specific demands of over-ground running. Treadmill running may not adequately address the specific biomechanical and neuromuscular requirements of the individual's sport or activity, leading to suboptimal rehabilitation outcomes. These differences can also impact the training workload and the stress placed on different body tissues. If training loads are prescribed based solely on treadmill running data, without considering over-ground running demands, it may lead to incorrect training prescriptions. This can potentially result in inadequate training adaptations, overuse injuries, or performance decrements. Competitive runners often train and compete in outdoor environments, which involve various terrains, weather conditions, and external factors. Treadmill running may not fully replicate these real-world conditions. As a result, competitive runners who primarily train on treadmills may face challenges when transitioning to over-ground running, such as adjusting to different surfaces, adapting to wind resistance, and managing changes in terrain. These challenges can impact performance, pacing strategies, and overall race outcomes. It is important to note that while treadmill running offers certain benefits, such as controlled environments and convenience, it may not fully replicate the demands and complexities of over-ground running. Understanding these differences and incorporating a combination of treadmill and over-ground running, when appropriate, can help mitigate the potential challenges and ensure more effective rehabilitation and training outcomes.

Although there has been extensive research on spatiotemporal and kinetic parameters of gait, there is limited research on the SSV of joint angles during treadmill running versus over-ground running. Despite the extensive research on physiological, performance, and perceptual differences between treadmill and over-ground running, there are fewer studies that have investigated SSV among runners, especially after prolonged running and as a function of duration and surface of running. Prolonged running can easily be done and monitored on treadmill inside laboratory but not while over-ground running. Thus, studies comparing treadmill and over-ground for distance running is sparse. Further research in this area could provide valuable insights into the differences between treadmill and over-ground running and help improve training, injury prevention and rehabilitation practices for athletes as well as patients.

2.5 Summary

Prolonged running has become a popular activity worldwide, with a growing number of people participating in marathon events. While prolonged running has been shown to have health and fitness benefits, it is also associated with an increased risk of RRIs. The repetitive strain placed on the hip and knee joints during long-distance running can lead to fatigue, biomechanical changes, and potential injuries. Understanding the relationship between environmental demands, the body's response, and the development of injuries among long-distance runners is crucial. While there are several risk factors identified for the development of RRIs, such as a history of previous injury and high weekly mileage, more research is needed to fully understand how these factors contribute to the development of injuries. Understanding the risk factors and developing interventions for injury prevention is crucial for the long-term health and safety of long-distance runners.

Running variability refers to the differences or fluctuations in various aspects of running gait, such as speed, stride length, cadence, and heart rate. Monitoring running variability can provide insights into the efficiency, performance, and potential risk of injury for a runner. SSV, which is the differences between strides, has been observed during ambulation. Initially considered as noise, researchers now recognize that SSV is functional and reflects the adaptability of the sensorimotor system. However, both high and low SSV can be detrimental to running, and there is an "optimal window" of variability that allows for greater flexibility while maintaining desired outcomes. Various factors, including fatigue, injury, age, and pathological neuromotor conditions, can affect running gait variability. Fatigue during running can result in biomechanical changes and alterations in lower limb movements. Studies have shown that fatigue induced by prolonged running can increase the risk of injury. Older adults and individuals with neurodegenerative diseases may exhibit higher variability in gait patterns. Understanding these factors is important for assessing gait stability and identifying potential risks. Different methods, like instrumented methods, qualitative measures, wearable sensors, and others, can be used to measure gait variability. Parameters including traditional linear methods, like CV, SD, range and frequency domain measures, and nonlinear analytical methods, like Lyapunov exponent, fractal dimension, and Poincaré analysis are used to quantify gait variability.

Comparing running on a treadmill to running over-ground has revealed various differences. Treadmills provide a controlled indoor environment and require limited space, making them popular for training and research purposes. However, treadmill running may not accurately simulate the physiological demands of outdoor running. Studies have shown differences in physiological, perceptual, performance, and biomechanical measures between treadmill and over-ground running. Runners generally prefer running over-ground as it allows for greater control over speed and reduces the risk of falling. The kinetics of walking and running also differ between treadmills and over-ground, including energy consumption, ground reaction force, and muscle activity. These differences can lead to poor rehabilitation outcomes and training workload prescriptions. While there has been extensive research on spatiotemporal and kinetic parameters of gait, there is limited research on the SSV of joint angles during treadmill running versus over-ground running. Further research is needed to explore the effect of duration and surface of running on variability, injury risk, and performance outcomes, particularly in natural running environments and among different populations. Future studies should aim to see that to what extent that treadmill assessment can reflect over-ground situation.

They may not need to be identical but help to identify the extent that treadmill kinematics can reflect over-ground kinematics in running.

Chapter 3: Accuracy Validation of Opal Inertial Measurement Units for Measuring Lower Limb Joint Angles

3.1 Introduction

Human movement is evaluated by analysis of gait that includes assessment of ability, pattern, posture, as well as balance while walking (He et al., 2024). This provides information about physiological status, capability, and any disorders about an individual (Klöpfer-Krämer et al., 2020; Sethi et al., 2022). Gait analysis has been widely used to assess neurological, musculoskeletal and other age-related disorders like dementia (Chakraborty et al., 2022; Cicirelli et al., 2021; Mc Ardle et al., 2020). It has also been used to develop and assess rehabilitation plan that can assist athletes to improve their walking postures and techniques to enhance their performance (DeJong et al., 2022; Mangone et al., 2023).

There are various methods available that have been widely used to analyze gait including motion capture systems, optical motion capture systems, pressure mats, and accelerometers (He et al., 2024). Motion capture systems have been regarded as gold standard for gait analysis. They track human movements using multiple cameras and sensors, converting them into raw data that can be converted into meaningful information (Jakob et al., 2021). Although these systems have been widely used in research over decades, they are highly expensive, involve complicated operation and are too large to be used outside laboratory (Abhayasinghe et al., 2019). These factors relatively limit their use in clinical applications and sports assessment in natural settings.

Another popular method for gait analysis is using wearable sensors based on IMUs (Zhang et al., 2023). Smaller size and lower cost as compared to motion capture systems has led to their increased usage for movement analysis outside the laboratory (Manupibul et al., 2023; Prasanth et al., 2021). Besides these, IMU-based systems have other advantages like comport, portable, user-friendly and simple calibration process (Park & Yoon, 2021). Various studies have reported feasibility, reliability and validity of using IMUs for different purposes, including measuring biomechanical gait outcomes during walking in different terrains, measuring lower limb gait kinematics and temporal-spatial parameters among healthy as well as patient population (Kobsar et al., 2020; Park & Yoon, 2021; Tao et al., 2012). As the previous studies have mainly focused on spatiotemporal parameters and patients with gait impairment and other disorders, it is more accurate to compare IMUs and motion capture systems for measuring joint angles in gait analysis among healthy adult subjects (He et al., 2024). In order to further promote the universal use of the IMU-based system, especially in sports settings outside the laboratory, it is important to analyze their accuracy and reliability against the gold standard motion capture system.

The APDM IMU sensors have been developed for gait and mobility assessment, tailored according to the needs of clinical assessment. They allow researchers to assess gait unobtrusively in a quick and simple way (Mancini & Horak, 2016). These sensors are wireless devices with a size of a wristwatch designed to capture as well as store three-dimensional data using onboard accelerometers (linear acceleration), gyroscopes (angular velocity), and magnetometers (magnetic field data to provide directional orientation) (Rebula et al., 2013). These data can provide information about various gait parameters like stride length, velocity, cadence, stance and swing time, etc. using in built Mobility Lab software (Salarian et al., 2004). Several researchers and clinicians use the APDM devices throughout the world, however, repeatability and validity data are limited, especially for joint angles and other gait metrics during movement. Therefore, the objective of this study was to evaluate the accuracy of opal sensors (APDM Inc., Portland, OR, USA) with gold standard motion capture system (Vicon Nexus, 2.12.1 Oxford, UK) for estimating lower limb joint angles during sagittal plane movements while walking. This information is critical as it could significantly affect the validity of research evaluating and analyzing the gait data in the natural settings outside the laboratory.

3.2 Methods

3.2.1 Study Design

This study was a validation study where gait analysis was done using both Opal IMUs and Vicon Motion capture system systems simultaneously.

3.2.2 Instrumentation

Seven Opal Movement Monitoring IMUs (APDM Inc., Portland, OR, USA) were tightly fixed on lower back at the level of L5S1 and two each on bilateral thighs, shank, and foot using elastic straps according to instructions provided by the manufacturers. In addition, a high-precision motion capture system (Vicon Nexus 2.12.1) with 8 cameras was also used with 28 reflective markers placed on bilateral pelvis, hip, thigh, knee, leg, ankle and foot according to Conventional Gait Model (CGM) instructions provided by the manufacturers CGM 2.3. Placement of sensors and markers are shown in figure 3.1. Body height, mass, leg length, knee and ankle joints width were input to scale the Vicon biomechanical model. Frequency of the data captured by the Vicon system was adjusted to match with the frequency of the Opal system, i.e. 128 Hz. Both systems were calibrated as per manufacturer guidelines before data collection.

3.2.3 Procedure

The sensors and markers were placed simultaneously on the subject. Data was collected while 10-sec walk test at a comfortable speed inside the laboratory. Subject was given time to familiarize with the set-up and procedure. Same single subject performed 10 repetitions of 10 sec walk test. The Opal system recorded data throughout the process while Vicon system focused at the midpoint of the walking path where three force plates were located. Gait cycles that met Vicon system standards and their corresponding data from Opal system were included in the final analysis. The maximum and minimum ranges and total range of motion of Hip, Knee and Ankle joints during each gait cycle were included in the analysis.

Figure 3.1: Sensor and marker positions for Opal and Vicon systems

3.2.4 Data Processing and Analysis

For Opal system, Moveoexplorer Mobility Lab software was used to analyze and process the raw data into desired joint angles. For Vicon system, the motion analysis data were analyzed using Visual 3D (C-Motion, Germantown, MD). A model was created that identified and defined the locations of all the reflective markers and bone segments. Each trial was separated into gait cycles and the local maximum and minimum peaks were identified within each gait cycle for both systems. The difference between these two values were identified as the joint range of motion.

To evaluate the accuracy of Opal system, the root mean squared error (RMSE) was calculated for the discrete variables maximum and minimum ranges of hip (flexion/extension), knee (flexion/extension), and ankle (dorsiflexion/plantarflexion) joints and their total range of motion over all the gait cycles. RMSE has been widely used in the literature to evaluate the performance of different IMU-based systems (Ferrari et al., 2010; Robert-Lachaine et al., 2017; Teufl et al., 2018).

3.3 Results

RMSE between Opal and Vicon systems for all parameters over all gait cycles are presented in Table 3.1. Range of RMSE for maximum (3.19 and 7.57), minimum (1.87 and 8.20) and total range of motion (0.73 and 2.29) in sagittal plane movements between the two systems was 4.38, 6.33 and 1.56 degrees respectively. The poorest outcome concerning the RMSE was evident in knee flexion/extension (5.83-8.20) while best outcome seen in ankle plantar-flexion/dorsi-flexion (1.87-6.10). RMSE for range of motion difference was least for all the joints (0.73-2.29).

3.4 Discussion

This part of study evaluated the accuracy of lower limb joint angle measurements using APDM's Opal IMU-based system in comparison to Vicon Nexus motion capture system to enable us to reliably interpret data obtained using this system in the later parts of our study. Motion capture data are considered as the gold standard for gait analysis. The findings of this study indicate that it is accurate and comparable to measure joint angles between the two systems.

Various studies have compared the validity of different IMU-based systems with motion capture systems and most of them have indicated that they are useful for determining spatiotemporal variables and calculating ranges of motion (ROM) (Al-Amri et al., 2018; Hafer et al., 2020; He et al., 2024; Park & Yoon, 2021). Similar to results of current study, it has been reported that the difference of RMSE and range of motion error of the joint angles in the sagittal plane were less than 1 degree between the IMU-based system and Mocap (Teufl et al., 2018).

Besides healthy adults, IMU-based systems shave also been shown to record stable gait data among patients who underwent total hip arthroplasty where RMSE in the joint kinematics ranged from 0.24° to 1.25° demonstrating them to measure the spatiotemporal gait parameters with high accuracy (Teufl et al., 2019).

Table 3.1: Root mean squared error (RMSE, degrees) between Opal and Vicon systems over all gait cycles.

Joint	RMSE, deg		
	Flexion	Extension	Range of motion
RIGHT HIP	6.00	4.45	2.29
LEFT HIP	5.32	5.13	0.73
RIGHT KNEE	7.57	8.20	0.89
LEFT KNEE	6.15	5.83	0.90
	Dorsiflexion	Plantarflexion	Range of motion
RIGHT ANKLE	6.10	4.92	1.37
LEFT ANKLE	3.19	1.87	1.42

Researchers have evaluated the validity and repeatability of measuring spatiotemporal gait parameters with APDM's IMUs and Mobility Lab system (Mancini & Horak, 2016). Opal IMUs have displayed moderate to high validity while measuring most of the tested parameters while using both treadmill and over-ground and widely considered as repeatable and accurate for measuring spatiotemporal gait parameters among healthy young adults (Washabaugh et al., 2017). However, current study is one of the fewer study to measure and compare lower limb joint angles between Opal and Vicon systems while walking.

There are some concerns over various errors that are associated with estimation using IMU-based systems, including drift error over time that could be due to the integration of a signals from gyroscope (Xing et al., 2017). Interference from local magnetic field could also distort the IMU-based data (Teufl et al., 2018). In addition, unlike motion capture systems, IMU-based kinematic position data are not directly measured (Berner et al., 2020). These concerns can be overcome by validating the performance of IMU-based systems with gold standard motion capture systems and controlling the fixation of the sensors. This shall further promote the widespread use of such system in gait analysis outside the laboratory.

As compared to other systems, the advantages of IMU-based systems is that they have higher rate of data extraction, the number of gait cycles captured in the same time is much higher and they can record gait data throughout the process (He et al., 2024). As the IMU sensors are a combination of magnetometer, gyroscopes and accelerometers, they can be used to detect angular velocity and acceleration to indicate motion and its intensity.

3.5 Summary

The results of this part of the study show that Opal IMU-based system are accurate for measuring lower limb joint angles and comparable to the Vicon system that is considered to be a gold standard for gait analysis. These findings further suggest that the wearable IMU-based systems are capable of accurately evaluating and analyzing the gait data in the natural sports settings outside the laboratory.

Chapter 4: Stride Time, Lower Limb Joint Angles and Their Variability While Prolonged Running on a Treadmill and Over-ground

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4.1 Introduction

Research has widely explored physiological, performance, and perceptual differences, however, fewer studies have examined differences in SSV among runners, particularly after prolonged running. Prolonged running, such as marathons, offers numerous health benefits (Fields et al., 2010; Lee et al., 2014). Previous research has shown that increased marathon participation is associated with decreased joint pain and lower prevalence of arthritis (Ponzio et al., 2018). Various factors, such as running speed, fatigue, injury, and experience, have been shown to affect SSV (Maas et al., 2018; Mo & Chow, 2018b). It has also been shown to be affected by running surface inside the laboratory (Schütte et al., 2016). Most of the previous studies have studied fatigue induced by running a relatively shorter distance inside the laboratory and the effects of fatigue as a result of distance running over-ground has not been considered. Traditional motion capture systems used in laboratories have limited the long-term monitoring of gait patterns outside controlled environments. There is a research gap in long-term gait monitoring in natural settings to understand how gait patterns changes with duration and surface of running and runner's body response different demands of running environments and surfaces during prolonged running. With the advancements in portable technology and wearable sensors, like IMUs, have enabled gait analysis in daily life and real sports settings (Bussmann et al., 2009; Weiss et al., 2013). It has helped to overcome the limitations of laboratory research by providing more extensive and real-world data (Weiss et al., 2013). Such quantitative analysis of gait provides valuable insights into dynamic stability, fall risk, and task performance under various conditions to enhance injury prevention, rehabilitation, and training protocols (Dobkin & Dorsch, 2011). IMUs can also be utilized to differentiate motor behavior associated with running on different surfaces, such as indoor treadmills and outdoor over-ground settings, particularly during long-distance runs.

Stride time, the duration between consecutive foot landings during walking or running, is a crucial factor in running biomechanics, impacting efficiency and performance (Gindre et al., 2016; Hollman et al., 2011). Stride time depicts gait performance which can be indirectly measured by gait variability (Bruijn et al., 2013; Hamacher et al., 2011). Traditionally, stride time variability has been seen as random fluctuations or errors in gait performance (Baida et al., 2018). Linear methods like SD and CV have been used to measure this variability around the average value to understand its implications (Hollman et al., 2007; Riva et al., 2014; Stergiou & Decker, 2011). Stride time variability is significant parameter to consider while studying human locomotion and running biomechanics for several reasons. It is closely related to running efficiency, injury risk, running performance optimization, and fatigue assessment (Brindle et al., 2020; Folland et al., 2017). Monitoring and analyzing stride time variability can help to identify individuals prone to injury and adjust their training strategies, prevent overtraining, and optimize performance accordingly (Ducharme et al., 2018). A healthy and adaptable locomotor system can modulate stride time to adjust according to external factors. Understanding the patterns and structure of stride time variability can provide insights into how the body self-organizes and adapts to different running conditions. Previous studies have explored stride time variability in trained runners (Fuller et al., 2016; Fuller et al., 2017; Lindsay et al., 2014; Mann, Malisoux, Nührenbörger, et al., 2015; Mann, Malisoux, Urhausen, et al., 2015; Meardon et al., 2011; Nakayama et al., 2010) and established the relevance of SSV to performance and injury. However, little is known about its characteristics during prolonged running on a treadmill and over-ground.

While extensive research has focused on spatiotemporal and kinetic gait parameters (Agresta et al., 2018), there is limited research on SSV of joint angles while running on a treadmill and over-ground. Researchers have highlighted the need for exploration of kinematic differences in SSV of specific joint angles to understand the origins of variability within the stride as a whole (Mok et al., 2009; Van Hooren et al., 2020; Wight et al., 2022). Lower limb joint angle variability refers to the fluctuations and variations in joint angles that occur while running. It is unclear how SSV of joint angles can alter while running on a treadmill or over-ground. Monitoring joint angle variability on a treadmill and over-ground can provide valuable insights into adjustments that can be made to running technique and form to reduce injury risk, designing rehabilitation programs and optimize running efficiency. Furthermore, it can contribute to the understanding of biomechanical adaptations during different running surfaces, benefiting sports science, rehabilitation, and ergonomics. Given the importance of these factors, this study examines the lower limb joint angle variability during prolonged running on a treadmill and over-ground.

This part of the thesis study aimed to compare mean and variability of stride time, lower limb joint range of motion, and angles in different phases of gait cycle in the sagittal plane movements while prolonged running as a function of duration and surface (treadmill and over-ground) of running among healthy runners. It was hypothesized that SSV in stride time and lower limb joint angles would depend on duration and running surface, with an expected increase in variability from indoor treadmill running to outdoor over-ground running. The findings of this study would enhance the understanding of treadmill and over-ground-based assessments and training of running to real-life conditions and further device strategies to prevent RRIs, improve rehabilitation and performance outcomes.

4.2 Methods

4.2.1 Study Design

This study was conducted in a crossover study design where same group of subjects participated in both phases of the study.

4.2.2 Participants

Eleven healthy adult regular runners (9 males, age 40.8 ± 8.9 years, weight 61.4 ± 8.5 kg and height 170.2 ± 6.3 cm) participated in this study. Demographic data of individual participants has been included as appendix A. They were recruited from the university and local community through advertisements and invitations. They were included if there was no history of musculoskeletal injury. Individuals with any pain, gait or neurological abnormality that could affect balance were excluded from the study. All participants were required to fill revised Physical Activity Readiness Questionnaire prior to any data collection. They were informed about all the procedures adopted in the study to obtain their informed consent before data collection. The ethical approval for this study was granted by The Human Research Ethics Committee of The Education University of Hong Kong (Ref. No. 2022-2023-0044, Appendix A).

4.2.3 Procedure

This study was conducted in 2 phases in random order with enough washout period to separate each running session. Subjects were asked to run indoors on a treadmill (GE Marquette 2000, U.S.A.) in phase 1 while in phase 2 they had to run outdoors on an all-weather track over-ground for 31 minutes at their preferred speed (Furlong & Egginton, 2018). Both the phases were conducted on separate days in random order. Subjects had to wear 7 Opal Movement

Monitoring IMUs (APDM Inc., Portland, OR, USA) while running (He et al., 2021; Lanovaz et al., 2017; Washabaugh et al., 2017). One sensor was tightly fixed on lower back at the level of L5S1 and two each on bilateral thighs, shank, and foot using elastic straps according to instructions provided by the manufacturers (figure 4.1). Data was wirelessly streamed to a computer at the sampling rate was 128 Hz. Subjects were asked to wear comfortable shoes and clothes and perform sufficient warm-up before each phase.

Figure 4.1: Placement of inertial measurement units on lower back at the level of L5S1 and two each on bilateral thighs, shank, and foot using elastic straps

4.2.4 Data processing

MATLAB (MathWorks BV, USA) was used to process raw unfiltered data obtained from the Mobility Lab software which is embedded within MoveoExplorer package provided with the Opal IMUs. Middle 30 min data were extracted to avoid any effects of initializing and ending of the recording. The initial (T1) and final (T2) 5 min were compared between the treadmill and over-ground running. Empirical Model Decomposition, which is a well-known time-frequency technique, was utilized to decompose the data into intrinsic mode functions. Each intrinsic mode function has a meaningful interpretation, such as trend, oscillation, and noises. Those intrinsic mode functions which corresponded to trend or non-stationary artifacts were removed, while remaining intrinsic mode functions were preserved to reconstruct the pure signal. Using built-in MATLAB functions, various features such as local maxima and local minima were extracted from the reconstructed pure signal.

4.2.5 Parameters

Stride time is defined as the time elapsed between the heel strike or initial contact of two consecutive footsteps of the same foot and is expressed in milliseconds (figure 4.2). It was obtained by locating the heel strike or initial contact denoted by the peak hip extension (local minimum of the sensors paced on the left hip) to mark the beginning of gait cycle (De Asha et al., 2012). Total range of motion of hip, knee and ankle joints in the sagittal plane movements of both the sides were obtained according to procedure described in data processing section. Similarly, maximum hip extension and flexion for hip joint; knee angle at heel strike and terminal stance, maximum knee flexion in stance and swing phase for knee joint; and maximum ankle dorsiflexion and plantarflexion for ankle joint were also obtained. All the angles were measured for lower limb joints on both sides (figure 4.3). Mean, SD and Variability (denoted by CV) of stride time and lower limb joint angles and were used to compare main effect of duration and surface of running (Hausdorff, Nelson, et al., 2001; Hollman et al., 2007; Mo & Chow, 2018b; Riva et al., 2014). Hip (right side, 1 subject while treadmill running) and ankle (left side, 2 subjects while treadmill running and right side, 2 subjects while over-ground running) joints data were unstable with substantial offset errors and couldn't be included in analysis as abnormal waveforms were observed.

4.2.6 Data analysis

SPSS version 21 (IBM Corp., Chicago, IL, U.S.A) was used to conduct two-way 2 x 2 repeated-measures mixed-design analysis of variance to compare the means and CV of all the parameters between 2-time intervals (T1 and T2) and the 2 running surfaces (treadmill and over-ground) during prolonged running with statistical significance set at p less than 0.05. Post-hoc analysis was conducted using the Bonferroni correction.

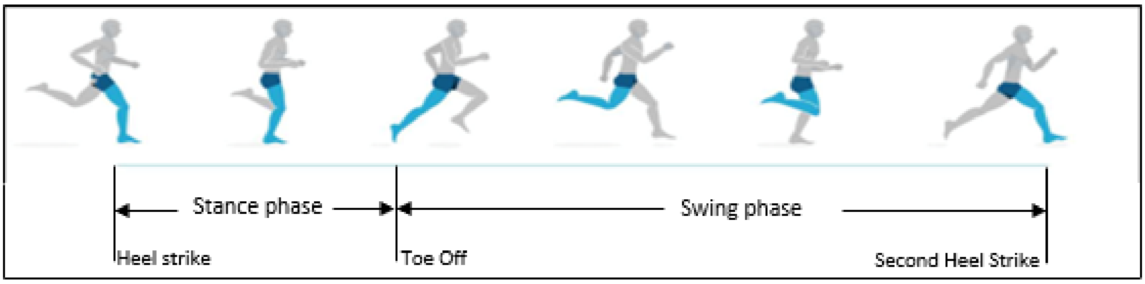


Figure 4.2: Stride time, also known as stride interval or gait cycle duration, is defined as the time elapsed between the heel strike or initial contact of two consecutive footsteps of the same foot and is expressed in milliseconds

4.3 Results

Mean, SD and CV of stride time, lower limb joints total range of motion and angles in the sagittal plane movements while prolonged running on a treadmill and over-ground are presented in tables 3.1 and 3.2.

4.3.1 Comparison of within and between subject's stride time while prolonged running on a treadmill and over-ground

There was no main or interaction effect of duration and surface of running on mean and variability of stride time while prolonged running on a treadmill and over-ground (all $p>0.05$) (figure 4.4).

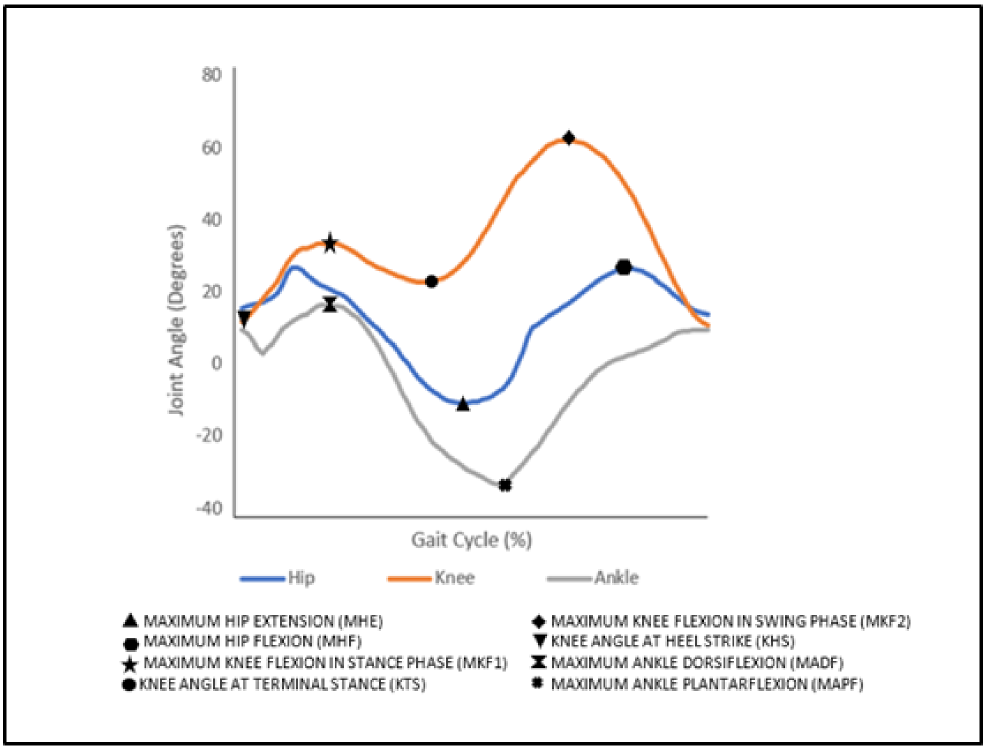


Figure 4.3: Joint angles in different phases of gait used in this study over sample participant's gait cycle

4.3.2 Comparison of within and between subject's lower limb joints total range of motion while prolonged running on a treadmill and over-ground

a. Total range of motion of hip joint: Although there was no main effect of duration of running but as compared to running on a treadmill, mean total range of motion of hip joint was significantly higher while running over-ground ($p<0.05$) in both right and left hip joints. In addition, there was significant running duration by running surface interaction effect ($p<0.05$) on variability of total range of motion of right hip joint. On left side, although there was significant interaction effect ($p<0.05$) but there was no simple effect (figure 4.5).

b. Total range of motion of knee joint: Although there was no significant main effect of running surface or interaction effect, but as compared to T2, mean total range of motion of left knee joint was significantly higher in T1 ($p<0.05$). There was no main or interaction effect of duration or surface of running in mean and variability of total range of motion of right knee joint (figure 4.6).

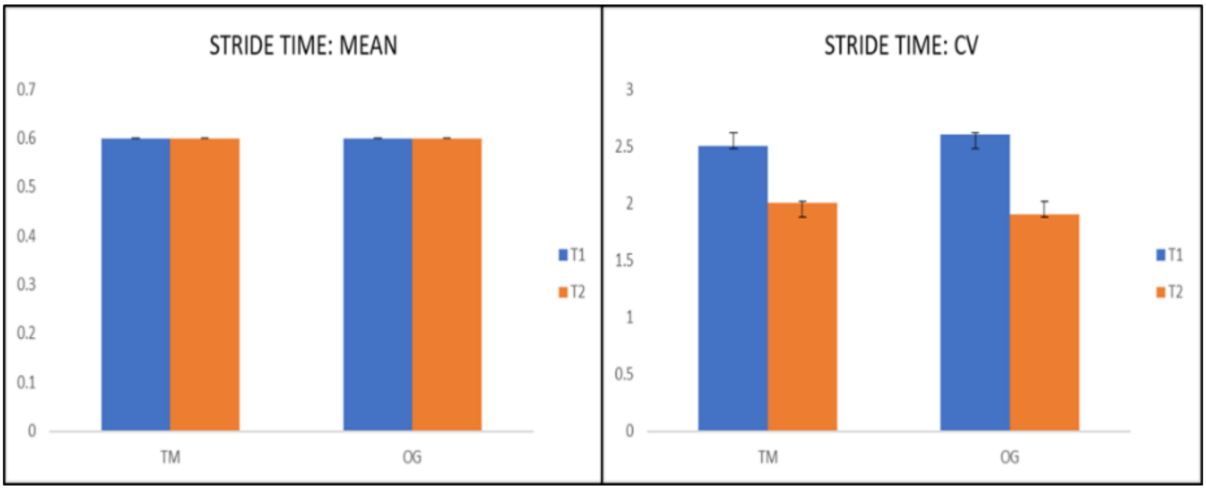


Figure 4.4: Comparison of mean and coefficient of variation (CV, %) of stride time between both durations (T1: 0-5 and T2: 25-30 minutes) and surfaces (treadmill, TM and over-ground, OG) of running. Note that there are no significant differences on the basis of duration or surface of running

c. Total range of motion of ankle joint: Although there was no significant main or interaction effect of duration or surface of running on mean or variability of total range of motion of ankle joint on both sides, but as compared to T2, its variability of was significantly higher in T1 ($p<0.05$) in the right side (figure 4.7).

4.3.3 Comparison of within and between subject's lower limb joint angles while prolonged running on a treadmill and over-ground

a. Maximum hip extension: There was no significant main effect or interaction effect of duration or surface of running on mean maximum hip extension of both sides. As compared to running on a treadmill, there was significantly higher variability of maximum hip extension ($p<0.05$) while running over-ground as well as running duration by running surface interaction effect ($p<0.05$) on left side. There was no significant difference in variability of maximum right hip extension (figure 4.5).

Table 4.1: Description of mean stride time and lower limb joint angles for both durations (T1: 0-5 and T2: 25-30 minutes) and surfaces (treadmill, TM and over-ground, OG) of running. Values are reported as Mean (standard deviation, SD). Significantly higher parameters are denoted by bold text.

ITEMS	RUNNING SURFACE	RUNNING DURATION	
		T1	T2
STRIDE TIME, Sec	TM	0.6 (0.0)	0.6 (0.0)
	OG	0.6 (0.0)	0.6 (0.0)
HIP JOINT: LEFT			
MAXIMUM HIP EXTENSION, Deg	TM	17.2 (7.5)	17.8 (7.4)
	OG	14.0 (6.1)	13.6 (6.5)
MAXIMUM HIP FLEXION***, ****, Deg	TM	31.8 (8.4)	**33.1 (8.9)**
	OG	45.7 (12.8)	**46.1 (13.7)**
HIP TOTAL RANGE OF MOTION**, Deg	TM	49.1 (5.8)	51.0 (7.3)
	OG	56.6 (7.7)	56.9 (7.8)
HIP JOINT: RIGHT			
MAXIMUM HIP EXTENSION, Deg	TM	15.7 (8.3)	15.8 (7.7)
	OG	15.2 (10.0)	15.1 (9.9)
MAXIMUM HIP FLEXION**, Deg	TM	34.4 (8.5)	35.0 (8.3)
	OG	45.4 (13.5)	46.6 (14.4)
HIP TOTAL RANGE OF MOTION**, Deg	TM	49.9 (6.8)	50.7 (7.5)
	OG	57.8 (5.3)	58.9 (6.6)
KNEE JOINT: LEFT			
ANGLE AT HEEL STRIKE, Deg	TM	16.1 (6.7)	17.2 (6.6)
	OG	18.1 (7.5)	18.8 (7.4)
MAXIMUM KNEE FLEXION IN STANCE PHASE, Deg	TM	39.9 (5.9)	40.4 (8.5)
	OG	42.8 (5.2)	42.4 (5.7)
ANGLE AT TERMINAL STANCE, Deg	TM	21.9 (5.9)	25.2 (11.4)
	OG	25.0 (5.6)	26.2 (6.6)
MAXIMUM KNEE FLEXION IN SWING PHASE, Deg	TM	78.0 (9.7)	77.8 (12.3)
	OG	82.2 (8.2)	81.6 (8.2)
KNEE TOTAL RANGE OF MOTION*****, Deg	TM	**56.0 (7.1)**	52.6 (7.1)
	OG	**57.2 (9.5)**	55.4 (10.8)
KNEE JOINT: RIGHT			
ANGLE AT HEEL STRIKE, Deg	TM	7.2 (5.0)	4.1 (10.9)
	OG	9.9 (7.4)	12.0 (10.7)
MAXIMUM KNEE FLEXION IN STANCE PHASE, Deg	TM	33.7 (5.4)	35.35 (10.4)
	OG	36.4 (6.8)	35.39 (6.6)
ANGLE AT TERMINAL STANCE, Deg	TM	14.8 (6.3)	19.5 (15.1)
	OG	17.0 (7.2)	16.0 (7.2)
MAXIMUM KNEE FLEXION IN SWING PHASE, Deg	TM	83.9 (10.1)	86.5 (11.3)
	OG	88.8 (12.0)	90.7 (12.5)
KNEE TOTAL RANGE OF MOTION, Deg	TM	69.1 (13.8)	67.0 (20.1)
	OG	71.7 (13.6)	74.7 (15.1)

ANKLE JOINT: LEFT			
MAXIMUM ANKLE DORSIFLEXION, Deg	TM	22.5 (4.6)	23.7 (5.0)
	OG	20.1 (5.2)	20.4 (4.3)
MAXIMUM ANKLE PLANTARFLEXION**, Deg	TM	25.5 (5.8)	25.6 (4.9)
	OG	33.9 (6.8)	34.2 (8.4)
ANKLE TOTAL RANGE OF MOTION, Deg	TM	48.0 (3.3)	49.3 (4.2)
	OG	54.0 (6.8)	54.6 (8.5)
ANKLE JOINT: RIGHT			
MAXIMUM ANKLE DORSIFLEXION*, Deg	TM	19.6 (4.5)	**20.7 (3.2)**
	OG	20.0 (4.8)	**20.7 (4.0)**
MAXIMUM ANKLE PLANTARFLEXION, Deg	TM	34.3 (7.1)	34.7 (6.8)
	OG	35.7 (5.6)	35.6 (5.6)
ANKLE TOTAL RANGE OF MOTION, Deg	TM	54.0 (7.2)	55.4 (5.3)
	OG	55.8 (7.1)	56.3 (7.0)

*main effect of time, p<0.05 **main effect of group, p<0.05 #interaction effect phase by time, p<0.05 ## interaction effect time by phase, p<0.05

b. Maximum hip flexion: As compared to while running on a treadmill, mean maximum hip flexion was significantly higher while running over-ground (p<0.05) on both sides. Additionally, as compared to T1 there was significantly higher mean maximum left hip flexion in T2 (p<0.05). There was no significant main effect of running duration in right side as well as there was no significant main or interaction effect of duration or surface of running on variability of maximum hip flexion in both sides (figure 4.4).

c. Knee angle at heel strike: There was no significant main effect or interaction effect of duration or surface of running on mean and variability of knee angle at heel strike on both sides (figure 4.6).

d. Knee angle at terminal stance: There was no significant main or interaction effect of duration or surface of running on mean and variability of knee angle at terminal stance on both sides but as compared to T2, its variability was significantly higher in T1 (p<0.05) in the left side (figure 4.6).

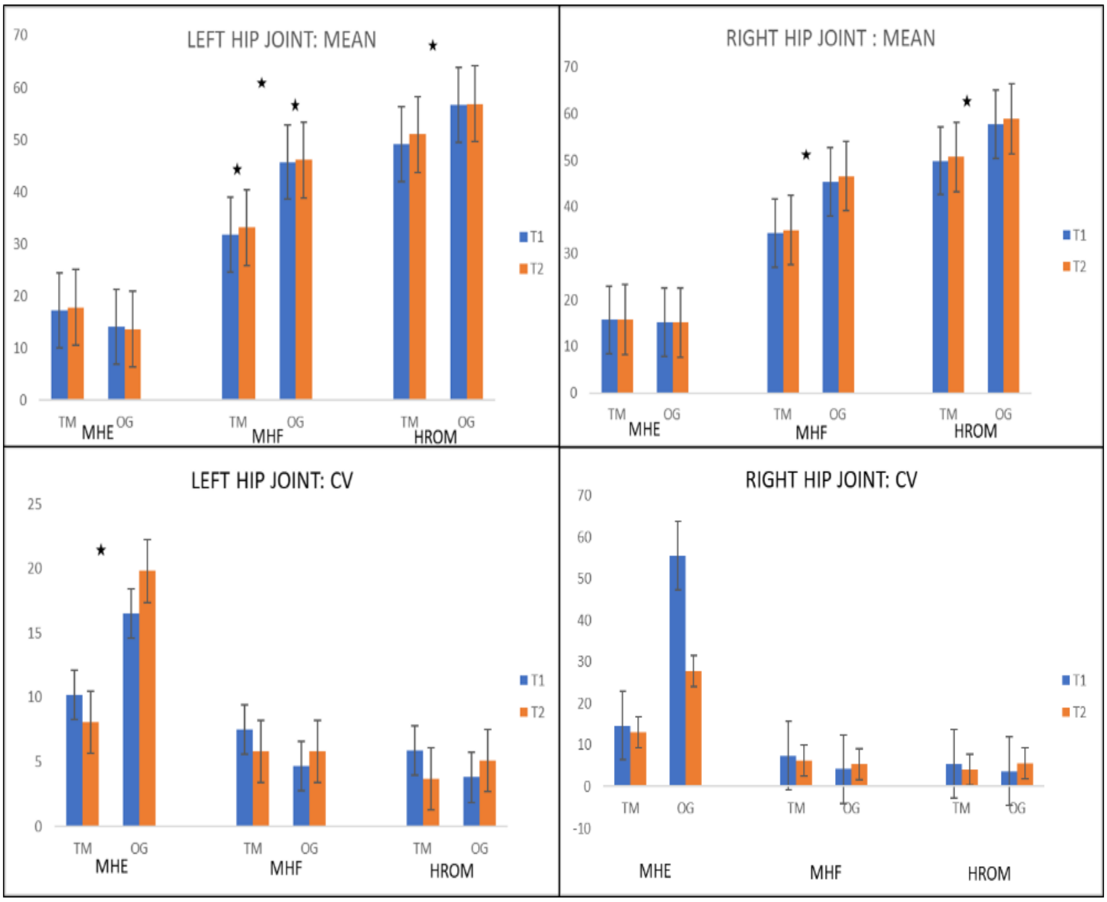


Figure 4.5: Comparison of mean and coefficient of variation (CV, %) of hip joint angles between both durations (T1: 0-5 and T2: 25-30 minutes) and surfaces (treadmill, TM and over-ground, OG) of running. Note the significant differences denoted by *. MHE: Maximum hip extension, MHF: Maximum hip flexion, and HROM: Total range of motion of hip joint

e. Maximum knee flexion in stance phase: There was no significant main or interaction effect of duration or surface of running on mean and variability of maximum knee flexion in stance phase on both sides but as compared to running on a treadmill, there was its variability was significantly higher while running over-ground ($p<0.05$) in the left side. Additionally, there was significant running duration by running surface and running surface by running duration interaction effects ($p<0.05$) in the left side (figure 4.6).

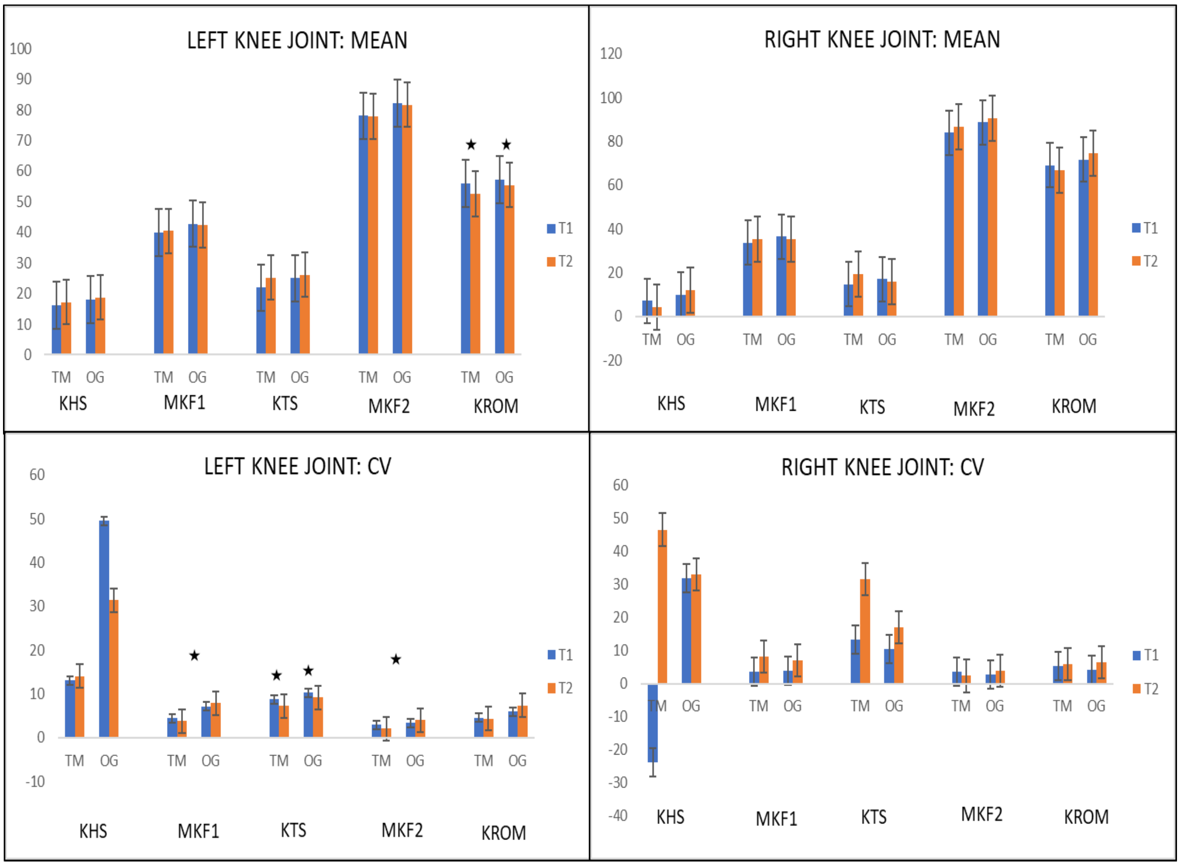


Figure 4.6: Comparison of mean and coefficient of variation (CV, %) of knee joint angles between both durations (T1: 0-5 and T2: 25-30 minutes) and surfaces (treadmill, TM and over-ground, OG) of running. Note the significant differences denoted by *. KHS: Knee angle at heel strike, MKF1: Maximum knee flexion in stance phase, KTS: Knee angle at terminal stance, MKF2: Maximum knee flexion in swing phase and KROM: Total range of motion of knee joint

f. Maximum knee flexion in swing phase: There was no significant main or interaction effect of duration or surface of running on mean and variability of maximum knee flexion in swing phase on both sides but as compared to running on a treadmill, its variability was significantly higher while running over-ground ($p<0.05$) in the left side (figure 4.6).

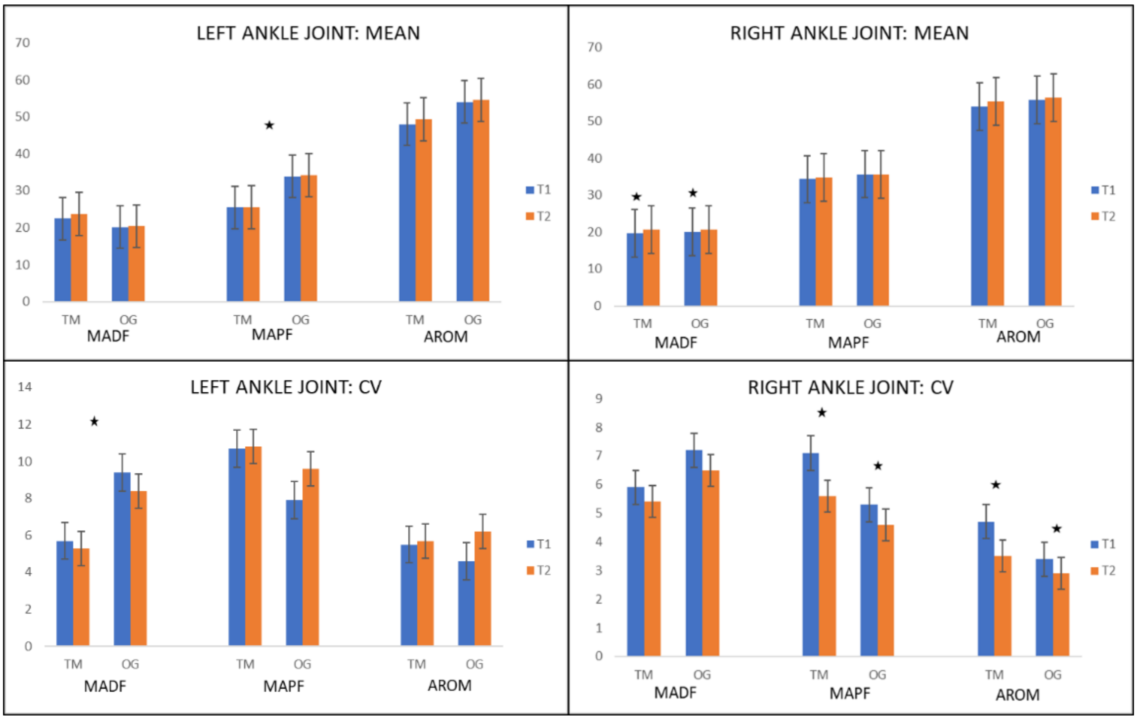


Figure 4.7: Comparison of mean and coefficient of variation (CV, %) of ankle joint angles between both durations (T1: 0-5 and T2: 25-30 minutes) and surfaces (treadmill, TM and over-ground, OG) of running. Note the significant differences denoted by *. MADF: Maximum ankle dorsiflexion, MAPF: Maximum ankle plantarflexion, and AROM: Total range of motion of ankle joint

g. Maximum ankle dorsiflexion: There was no significant main or interaction effect of duration or surface of running on mean maximum left ankle dorsiflexion but as compared to running on a treadmill, its variability was significantly higher while running over-ground ($p<0.05$). In right side, as compared to T1, mean maximum ankle dorsiflexion was significantly higher in T2 ($p<0.05$) while there was no significant difference in the variability of maximum ankle dorsiflexion (figure 4.7).

h. Maximum ankle plantarflexion: As compared to running on a treadmill, mean maximum ankle plantarflexion was significantly higher while running over-ground ($p<0.05$) but there was no main effect of running duration or interaction effect in the left side. There was no main or interaction effect of duration or surface of running on variability of left maximum ankle plantarflexion. On right side, although there was no significant main or interaction effect of duration or surface of running on mean and variability of maximum ankle plantarflexion but as compared to T1, its variability was significantly higher in T1 ($p<0.05$) (figure 4.7).

Table 4.2: Description of coefficient of variation of stride time and lower limb joint angles for both durations (T1: 0-5 and T2: 25-30 minutes) and surfaces (treadmill, TM and over-ground, OG) of running. Values are reported as mean percentage (standard deviation, SD). Significantly higher parameters are denoted by bold text.

ITEMS	RUNNING SURFACE	RUNNING DURATION	
		T1	T2
STRIDE TIME, Sec	TM	2.5 (1.0)	2.6 (1.8)
	OG	2.0 (0.8)	1.9 (0.6)
HIP JOINT: LEFT			
MAXIMUM HIP EXTENSION**, ##, Deg	TM	10.2 (4.2)	8.1 (3.6)
	OG	16.5 (11.0)	19.8 (12.8)
MAXIMUM HIP FLEXION MHF, Deg	TM	7.5 (4.1)	5.8 (2.5)
	OG	4.7 (1.5)	5.8 (3.2)
HIP TOTAL RANGE OF MOTION@, Deg	TM	5.9 (2.9)	3.7 (1.2)
	OG	3.8 (0.9)	5.1 (3.5)
HIP JOINT: RIGHT			
MAXIMUM HIP EXTENSION, Deg	TM	14.7 (14.2)	13.1 (10.4)
	OG	55.5 (151.7)	27.8 (45.7)
MAXIMUM HIP FLEXION, Deg	TM	7.4 (3.6)	6.3 (4.2)
	OG	4.2 (2.0)	5.3 (2.9)
HIP TOTAL RANGE OF MOTION##, Deg	TM	5.4 (2.6)	4.0 (1.9)
	OG	3.7 (0.6)	5.6 (3.7)
KNEE JOINT: LEFT			
KNEE ANGLE AT HEEL STRIKE, Deg	TM	13.0 (5.6)	14.0 (12.6)
	OG	49.5 (104.9)	31.4 (46.6)
MAXIMUM KNEE FLEXION IN STANCE PHASE**, #, ##, Deg	TM	4.4 (1.5)	3.8 (1.5)
	OG	7.1 (4.6)	7.9 (5.2)
KNEE ANGLE AT TERMINAL STANCE*, Deg	TM	**8.7 (3.9)**	7.2 (3.6)
	OG	**10.2 (6.3)**	9.2 (5.7)
MAXIMUM KNEE FLEXION IN SWING PHASE**, Deg	TM	2.9 (1.3)	2.0 (0.5)
	OG	3.3 (0.7)	4.0 (2.7)
KNEE TOTAL RANGE OF MOTION, Deg	TM	4.5 (1.0)	4.3 (1.6)
	OG	5.9 (3.5)	7.3 (5.8)
KNEE JOINT: RIGHT			
ANGLE AT HEEL STRIKE, Deg	TM	-23.8 (197.8)	46.6 (55.6)
	OG	31.8 (26.1)	33.1 (23.8)

MAXIMUM KNEE FLEXION IN STANCE PHASE, Deg	TM	3.7 (0.7)	8.2 (17.2)
	OG	3.8 (1.3)	7.1 (11.6)
ANGLE AT TERMINAL STANCE KTS, Deg	TM	13.2 (12.2)	31.7 (51.9)
	OG	10.5 (7.1)	17.0 (26.7)
MAXIMUM KNEE FLEXION IN SWING PHASE, Deg	TM	3.6 (1.6)	2.3 (0.7)
	OG	2.8 (1.0)	3.9 (3.7)
KNEE TOTAL RANGE OF MOTION, Deg	TM	5.2 (1.9)	5.9 (7.3)
	OG	4.2 (1.4)	6.5 (8.1)
ANKLE JOINT: LEFT			
MAXIMUM ANKLE DORSIFLEXION**, Deg	TM	5.7 (1.4)	5.3 (2.2)
	OG	9.4 (4.5)	8.4 (2.9)
MAXIMUM ANKLE PLANTARFLEXION, Deg	TM	10.7 (5.9)	10.8 (8.3)
	OG	7.9 (2.8)	9.6 (8.6)
ANKLE TOTAL RANGE OF MOTION, Deg	TM	5.5 (1.9)	5.7 (3.7)
	OG	4.6 (1.9)	6.2 (7.0)
ANKLE JOINT: RIGHT			
MAXIMUM ANKLE DORSIFLEXION, Deg	TM	5.9 (2.1)	5.4 (1.7)
	OG	7.2 (3.0)	6.5 (1.9)
MAXIMUM ANKLE PLANTARFLEXION*, Deg	TM	**7.1 (3.5)**	5.6 (1.9)
	OG	**5.3 (1.2)**	4.6 (1.1)
ANKLE TOTAL RANGE OF MOTION*, Deg	TM	**4.7 (1.8)**	3.5 (0.9)
	OG	**3.4 (0.6)**	2.9 (0.5)

*main effect of time, $p<0.05$ **main effect of group, $p<0.05$ #interaction effect phase by time, $p<0.05$ ## interaction effect time by phase, $p<0.05$, @interaction effect but no simple effect

4.4 Discussion

This is one of the first study to explore SSV differences during prolonged running that has been enabled by the use of IMUs allowing researchers to assess runners while running in the natural environment outside the laboratory. The mean and variability of stride time, lower limb joint's total range of motion and angles in different phases of gait cycle in the sagittal plane movements were compared between initial and final 5-minute duration of 30-minute run over a treadmill and over-ground. Results show that there was no significant difference in the stride time or its variability on basis of duration or surface of running. However, there was significantly higher variability in lower limb joint angles in the initial duration, while certain mean angles were higher toward in the final duration of running. As compared to running on a treadmill, both mean and variability were significantly higher while running over-ground. These findings support the hypothesis of the study that SSV in lower limb joint angles is dependent on duration and surface of running and variability of the tested parameters was higher outdoor over-ground as compared to indoor on a treadmill.

4.4.1 Prolonged running on a treadmill and over-ground

It is important assess dynamic and mechanical differences while prolonged running on a treadmill and over-ground as SSV and stability of specific characteristics of gait has been shown to vary with the environment and testing conditions (Tamburini et al., 2018). Treadmill provides a constrained environment where runners run on a fixed speed creating a more stable and uniform conditions but require more energy to maintain it. It has been proposed that treadmill requires less propulsion force as it is the supporting leg that moves on the belt and not the body moving over it (Frishberg, 1983). Theoretically, according to the coordinate system that moves with the belt, running on a treadmill and over-ground are similar when speed of belt is constant and air resistance is negligible (van Ingen Schenau, 1980). However, belt speed has been shown to decelerates at heel strike and accelerates during at toe off that causes the difference in biomechanics (Frishberg, 1983; Nigg et al., 1995; Schache et al., 2001). Runners may also feel less comfortable while running on a treadmill due to higher risk of falling, lack of evaporation of sweat, lower thermoregulation and absence of any extra attentional requirements compared to over-ground running which provides room to focus more on physiological sensations like increased heart rate and fatigue which can further affect strategies to control running technique, speed and economy (Heesch & Slivka, 2015; Schücker & Parrington, 2019). Altogether, these findings suggest that such differences may lead to higher energy cost and reduced running performance while running on a treadmill as compared to over-ground. Running over-ground offers more degrees of freedom that differs in loading and neuromuscular modulation that are clinically more important.

Although there are various studies that have compared kinetic, kinematic, spatiotemporal and muscle activity related parameters while running on a treadmill and over-ground, it has been recommended that kinematic differences in sagittal plane movements on different surfaces needs to be explored further (Mok et al., 2009; Van Hooren et al., 2020). This study focused on the sagittal plane motions since running is broadly a movement in sagittal plane and kinematics in sagittal plane have been shown to be more relevant to injury and performance (Wille et al., 2014). Results of current study show that all the parameters had significantly higher mean and variability while running over-ground that adds fuel to the argument that treadmill running is not able to simulate over-ground running. These findings are also in line with the previous studies that show that walking and short-term running on a treadmill exhibits less variant characteristics than over-ground and that laboratory-based assessment of running biomechanics does not represent the correct picture of natural outdoor running over-ground (Hollman et al., 2016; McCrory et al., 2022).

Running posture is an important factor to consider while comparing running on a treadmill and over-ground. Although body posture has not been considered in studies involving gait analysis, it has been observed that as the treadmill offers running at fixed speed, runners usually run in upright position due to reduced air resistance (Van Hooren et al., 2020). On the other hand, runners have to lean forward while running over-ground in order to achieve faster speeds and avoid air resistance. Various studies that have reported body kinematics while running on a treadmill and over-ground provide contradictory reports on truck lean angle (Nigg et al., 1995; Riley et al., 2008; Schache et al., 2001). Although the mechanism is still unclear, the change in the truck lean angle while running on a treadmill could also be due to the external drag force from the belt (Wank et al., 1998). As the IMUs provide joint angles as the relative position of the proximal and distal sensors, running posture may be a potential cause of the differences in joint angles while running on a treadmill and over-ground. Running Posture is an important factor that should be considered while conceptualizing future research studies. Further investigation considering trunk motion and experience of runners as factors that can affect running pattern would be more valuable to draw further conclusions.

Another important factor to consider while comparing the effect of duration and surface of running is speed of running. Previous studies have reported inconsistent findings regarding impact of speed during prolonged running on stride time variability and should be interpreted with caution (Fuller et al., 2017; Paquette et al., 2017). Different perception of speed among runners can affect running biomechanics while running on a treadmill (Sloot et al., 2014). Participants in this study were asked to run at their preferred speed throughout running duration on both surfaces. Previous studies comparing walking or running on a treadmill and over-ground provided mixed instructions to subjects regarding speed. While some permitted them to choose their own preferred speeds others constrained them to a predetermined speed (Dingwell et al., 2001; Terrier & Dériaz, 2011). Imposing speed lower than the preferred speed may influence stride time variability (Dingwell & Marin, 2006). As humans tend to adopt an energy efficient gait and prefer to walk or run at their optimal speed, it is difficult to determine the effect of the speed of running on the running style and economy (Selinger et al., 2022). In order to improve their running economy, long distance runners have been shown to modulate their segments and joints as a part of pacing strategy to maintain self-preferred speed and intensity (Agresta et al., 2018). The advantage of running at the self-preferred speed is that it is familiar and comfortable for the runner. Running at a different speed from this acts as a biological stressor that can lead to a more predictable and less adaptable stride variability (Jordan et al., 2006; Meardon et al., 2011), hence it is important to consider that variable running speeds can affect the cross-sectional nature while such comparisons. It is still not very clear how running speed can influence SSV measures (Wight et al., 2022). Biomechanical changes while running on a treadmill could also be affected by the runner's comfort, familiarity, focus, difference in perception, surface hardness, and mechanical features of the model (Asmussen et al., 2019; Fu et al., 2015; Lucas-Cuevas et al., 2018; Nigg et al., 1995; Schieb, 1986; Sloot et al., 2014; van Ingen Schenau, 1980). While degree of runner's familiarization and comfort while running on a treadmill has been shown to affect running biomechanics, their comfort and experience while running have rarely been considered in previous studies (Miller et al., 2019; Schieb, 1986).

Variability in the training modalities can influence motor performance by allowing users to accommodate to contextually different demands by moving at different speeds (Reisman et al., 2007). The change in gait variability with duration and surface of running among healthy adults shows that it is not a negative point as it is usually considered for older people (Tamburini et al., 2018; Verghese et al., 2009). Variability is required to maintain the stability of gait in response to the perturbations in the environment. Humans are designed ambulate such that they are able to adequately adapt and accommodate to dynamic environment that demands equally symmetric, stable yet rhythmic gait to prevent fall. Such a state of balance cannot be achieved without "optimal variability" in the pattern of gait. Inability to achieve this healthy variability in locomotor system has been considered as a unnatural or pathological (Lipsitz & Goldberger, 1992). In fact gait variability results from multiple degrees of freedom of the locomotor system that may bring more stability for achieving better movement control (Bruijn et al., 2013; Hamacher et al., 2011). Testing and environmental conditions are not expected to affect gait of a healthy young adult with a mature and structured pattern of motor control but to change the variability of the motor pattern as an adaptation to maintain stability (Gallahue & Ozmun, 2006). This crossover study testing the young healthy subjects on a treadmill and over-ground didn't expect gait stability to be affected much but modifications in the variability to adjust the gait pattern to the dynamic environment in order to maintain stability. Results further highlights the caution needed while applying laboratory-based findings to over-ground conditions in the context of rehabilitation, enhancement of running performance and importance of considering variability while conducting research and applying it clinically. When a runner would train in a condition that limits variability it would otherwise restrict optimal sensorimotor performance.

4.4.2 Stride time and its variability

While comparing the characteristics of stride time variability while long-distance running on a treadmill versus over-ground, several key differences can be observed, including magnitude of variability, influence of external factors such as terrain, weather conditions, surface irregularities, adaptability and response to external stimuli, feedback, proprioception, fatigue and performance. Understanding these differences can provide valuable insights into the characteristics and implications of stride time variability in different running contexts. Results of this study showing that stride times do not differ while running on treadmill and over-ground are consistent with previous studies that show similar findings while comparing walking on a treadmill and over-ground among healthy individuals (Dingwell et al., 2001; Terrier & Dériaz, 2011). Previous studies have also reported no difference in stride time, length and frequency while running on a treadmill and over-ground indicating that spatiotemporal parameters do not vary with running surface (Adams et al., 2016; Mann et al., 2014). Results of the current study also show that there was no difference in mean stride time or its variability among runners between the initial and final duration of 30 minutes run. Stride time variability increased from 2.5% (T1) to 2.6% (T2) while running on a treadmill while it decreased from 2% (T1) to 1.9% (T2) over-ground. These figures are in line with the range (1 to 3%) reported in previous studies (Fuller et al., 2017; Mann, Malisoux, Nührenbörger, et al., 2015; Mo & Chow, 2018b). A more consistent stride time allows for better pacing, efficient energy transfer, and improved overall performance. Stride time variability reflects the ability of the locomotor system to adapt and respond to varying environmental conditions and constraints (Ducharme et al., 2018). A healthy and adaptable locomotor system can modulate stride time to navigate uneven terrain, adjust to changes in running speed, or accommodate external factors.

4.4.3 Lower limb joint angles and their variability

Results of this study show that as running progressed, mean and variability of joint angles, (specifically maximum hip flexion and maximum ankle dorsiflexion) increased in both treadmill and over-ground conditions. Motor control has been shown to be affected by accumulation of fatigue (Corbeil et al., 2003; Mann, Malisoux, Urhausen, et al., 2015). With the precipitation of fatigue, as muscles tire, there may be increased variability in joint angles as the body compensates for reduced muscle strength and control. Fatigue can impact coordination, muscle activation patterns, and overall stability, leading to greater variability in lower limb joint angles. This increased variability was more apparent while over-ground running due to the additional demands posed by the outdoor environment. On the other hand, knee total range of motion was significantly decreased in final duration of running. Knee joint, majorly controlled by large quadriceps muscle, has higher strength reserve that delays fatigue accumulation thereby increasing the chances to achieve higher variability and more range of motion during activities of prolonged duration. Combined together, with decreased variability and increased joint angles, it may also be a compensatory adjustment by runners in order to achieve higher stride length while prolonged running.

Studies that have compared lower limb joint angles in different phases of running gait on a treadmill and indoor tracks using motion capture systems have reported some consistent results with the current study. These studies have reported that as compared over-ground running, lower limb joint angles in sagittal plane motions decreased while running on a treadmill in varying extents (Riley et al., 2008; Sinclair et al., 2013; Van Hooren et al., 2020; Yao et al., 2019). There was smaller maximum hip flexion during stance and ankle dorsiflexion while running on a treadmill as compared to indoor laboratory run way. The knee was reported to be more flexed at heel strike while running on outdoor track. Increase in the lower limb joint angles while running have been considered to be part of strategies to decrease lower limb stiffness that can compensate higher surface stiffness (Dixon et al., 2000). The smaller angle formed by foot during heel strike while running on a treadmill shows a compensatory mechanism to decrease lower limb stiffness on a stiffer treadmill surface as compared to compliant ground surface (Hardin et al., 2004; Kerdok et al., 2002). Another possible reason for such differences in joint angles is the lower loading on knee extensors while higher loading on plantar flexors (Gastrocnemius and Soleus) while running on a treadmill (Yao et al., 2019). In response to adjustment to the constant speed on the treadmill, neuromuscular system adopts smaller steps to achieve dynamic adjustment of muscular force and as gastrocnemius provides the main force to maintain the speed of running, its loading increases remarkably (Yao et al., 2019). Biomechanical variation while running on a treadmill and over-ground could be due to difference in the surface stiffness (Running, 2017; Taunton et al., 2003). If deemed necessary, one way in which researchers, athletes and coaches could try to minimize the difference between treadmill and over-ground running is by using treadmills that claim to replicate the surface stiffness of ground surface (Van Hooren et al., 2020). This would also allow generalizability of the results to improve clinical gait analysis and transfer of training.

Lower limb joint angle variability during prolonged running can vary with individual differences. Factors such as running experience, fitness level, and biomechanical characteristics can influence the extent of variability. Experienced runners with better running economy and neuromuscular control may exhibit lower joint angle variability compared to novice runners (Meardon et al., 2011; Mo & Chow, 2018b). Running speed can also influence lower limb joint angle variability during prolonged running due to the need for more precise coordination and control at higher speeds placing higher demands on the musculoskeletal system. This effect may be more pronounced in over-ground running, where speed variations are more frequent, less controlled and influenced by external factors while treadmills enable runners to maintain a consistent pace and cadence. SSV in lower limb joint angles is also as an indicator of fatigue (Wight et al., 2022). The findings of this study show that lower limb joint angles and their variability while running on a treadmill are different from that on over-ground, with some outcomes being significant to potentially impact research, training and clinical practice practically, while others missing the $p<0.05$ mark with a trivial magnitude. Although these differences may not appear to be relevant but when considered in isolation, their combined effect may be important that may secondarily be a compensation strategy for differences between the two running surfaces.

4.5 Summary

This study compared the differences in stride time and lower limb joint angles in different phases of gait while prolonged running on a treadmill and over-ground as a function of duration and surface of running. It is important to study dynamic and mechanical differences between running on a treadmill and over-ground as SSV varies with environment. Although it is beyond the objective of this study to nullify the validity of running simulation on a treadmill, it compares the runner’s body response based on duration, environment and surface of running. Findings of this study show no significant difference in stride time or its variability between duration and surface of running. However, SSV in joint angles was higher while running over-ground and in the initial duration of running while certain mean angles were higher in the final duration of running, which support the hypothesis of this study. Such differences could be due to lack of control over running speed and different fatigue and muscle activation patterns over the two running durations and surfaces. These findings suggest that interpretations of studies using treadmill in laboratory may not fully reflect the dynamics of distance running outdoors in natural environment and highlight the importance of considering the external environment and its impact on biomechanics. This is one of the first study to be consider long-distance running outside the laboratory in natural environment conducted using IMUs. It is important to consider variability in gait analysis and research, as well as the potential impact on training and clinical practice. While the sample size in this study may be relatively small, the findings provide a proof-of-concept for using IMUs in outdoor settings by validation of IMU data in outdoor conditions to provide insights into long-term changes in running mechanics and highlight the potential for future research to collect long-term data on running biomechanics.

Chapter 5: Lower Limb Coordination and Coordination Variability While Prolonged Running on Treadmill and Over-ground

5.1 Introduction

Running is a continuous task that involves multi-joint cyclical motion that is critical to the performance in various sports activities. It is usually performed under different conditions and it is normal for runners to modify their skill through practice on different running surfaces, running techniques and protocols. There are minor changes in loading with every step while running. The accumulation of this repetitive load can lead to overload which has been associated with overuse injuries (Xiang et al., 2022; Xu et al., 2022; Yamane et al., 2023). The rate of developing RRIs has been reported to be as high as 85% with about 30% of runners reporting lower-limb injuries with long-term effects (Dempster et al., 2021; Junior et al., 2013; Kluitenberg et al., 2015). Variability is the measure of ability to adapt to constraints of a given task (Preatoni et al., 2013). Coordination and its variability have been suggested as indicators of runner's adaptability to the changes or disturbances in the environmental conditions (Bernstein, 1967). Coordination is a goal-directed behavior and its modifications are required to change the pattern of movement in response to the task demands (Davids et al., 2003b). As changes in coordination variability are required to adapt or respond to constraints of the new task, greater variability suggests greater adaptability. Both high and lower coordination variability have been shown to be associated with higher risk of injury and poor performance among athletes (Hamill et al., 2012; Seifert et al., 2013). Therefore, it has been proposed that runners should maintain optimal level of coordination variability for reducing the risk of RRIs without compromising their performance (Hamill et al., 2012; Mo & Chow, 2019). However, optimal level of coordination variability is difficult to be defined due to unclear effect of various factors, like duration and surface of running, among healthy runners.

Lower limb coordination has been studied among healthy individuals (Boyer et al., 2014; Floría et al., 2018). There is a growing interest in coordination variability over the recent years due to its link with different factors like gender, injury status, level of skill, running form and aging (Boyer et al., 2017; DeLeo et al., 2004; Hafer et al., 2016; Hamill et al., 1999; Miller et al., 2008; Nakayama et al., 2010; Silvernail et al., 2015). Some studies have suggested that coordination variability could diminish while running on treadmill due to the imposition of a constant speed and decreased need to adapt to the environmental variations (Cazzola et al., 2016; Lindsay et al., 2014; Wheat et al., 2005). Controlled belt speed on the treadmill has the capacity to alter intrinsic dynamics of human locomotion that makes comparisons between treadmill and over-ground running important (Toro et al., 2022). Studies have reported that coordination variability decreases with increase in running cadence, suggesting that change in coordination variability may be related to individual response to task constraints (Hafer et al., 2016). Effect of speed on gait coordination and its variability while running has also been studied showing that with increase in speed there were moderate alterations in the frequency of movement pattern without altering the coordination classification (Floría et al., 2019). However, contradictory results have been reported regarding the effect of speed on coordination variability (Bailey, Silvernail, et al., 2018; Mehdizadeh et al., 2015; Pohl et al., 2010; Seay et al., 2011). Running at a constant speed on treadmill could reduce the range of variability and decreasing the tendency to adapt to changes in the environment (Cazzola et al., 2016; Lindsay et al., 2014; Wheat et al., 2005). It is common to examine running coordination using a treadmill (Hafer et al., 2017; Rodrigues et al., 2015), however these factors could lead to differences in coordination from that while running over-ground running. It is relatively less known how coordination variability can vary with duration and surface of running particularly differences in lower limb coordination while long-distance running indoors and outdoors. Most of the aforementioned studies are retrospective involving running under one condition that have been conducted inside the laboratory using motion capture systems to capture kinematic data while treadmill running. These studies cannot reflect the real-world situations indicative of sports conditions. Similarly, it remains unknown about how coordination variability would change with changes in practice conditions. As the popularity of running and associated injury risks has increased, it is important analyze differences in coordination and its variability while treadmill and over-ground running.

Due to associated health benefits, distance running has become more popular among runners in the recent times (Fields et al., 2010; Lee et al., 2014; Williams, 2009). As estimated, the cadence while running a marathon can reach up to 1.42 strides per second and as the stride frequency is high some variability in coordination is inevitable in the involved movements (Chan-Roper et al., 2012; Miller et al., 2010). Running for long duration exposes the body to varying levels of fatigue that can put stress, shear, strain, as well as impact forces potentially altering the running mechanics (Dierks et al., 2010; Milgrom et al., 2007). Some studies have suggested that fatigue can induce alterations in running biomechanics after long-distance running especially in recreational runners (Ferber et al., 2009; Luo et al., 2019; Willson et al., 2015; Winter et al., 2016; Yahya et al., 2022). However, fatigue in these studies was induced by running a relatively shorter distance or other activity protocols not including running (Bellenger et al., 2019). Progressive fatigue has been regarded as a common risk factor for development of RRIs especially among novice runners. No studies have yet investigated the effect of fatigue while long-distance running over-ground on SSV. This is due to the limitations of the traditionally used motion capture systems that cannot be used outside the laboratory. With the availability of IMUs that are core elements in wearable technology it is now possible to investigate the characteristics of stride variability while a long-distance running in a natural setting (Meardon et al., 2011; Mo & Chow, 2018a; Norris et al., 2014). Researchers have suggested that running on different surfaces can cause biomechanical differences that may affect accuracy and consistency of how other factors affect running mechanics (Patra et al., 2022; Van Hooren et al., 2020). It is important to study the biomechanics of runners before and after long-distance running to understand the relationship between faulty running mechanics and development of injury especially for novice runners (Schmitz et al., 2014).

CRP between segmental movements has been used to quantify the human dynamic system by to establishing the relative motion of two adjacent joints or segments. The continuous, multi-joint nature of the running makes it best to use CRP (Clark & Phillips, 1993). CRP provides effective measure to assess movement performance and has been used to identify coordination disorders among children (Volman & Geuze, 1998). It has also been used to compare prognosis in medicated and non-medicated patients of Parkinson's disease (Winogrodzka et al., 2005). Other studies have used relative phase adaptation while walking and running across post-surgical rehabilitation among patients with anterior cruciate ligament reconstruction (Kurz et al., 2005; Wyatt et al., 2021). Coordination while walking and high-intensity sports like running, swimming and jumping have also been investigated using CRP (Hu et al., 2021; Seifert et al., 2010). CRP analysis has been deemed to be more sensitive to changes in coordination (Abbasi et al., 2020; Lukšys et al., 2021; Whittle et al., 2022). CRP variability (CRPv) represents the functional role in the development of healthy movement patterns and the flexibility of the body for adaptation to perturbations during movement (Hu et al., 2021). It has also been used in differentiation of healthy and injured runners (Abbasi et al., 2020; Heiderscheit et al., 2002). CRP analysis is most applicable to cyclic movements, such as walking, running, or cycling.

The objective of this part of the study was to investigate the differences in coordination and its variability in lower limb joint-couplings in sagittal plane motions while long-distance running on a treadmill and over-ground track among healthy runners. The CRP was used to compute coordination, denoted by CRP angle and its variability (CRPv) for knee-hip, ankle-knee, and ankle-hip joints couplings of the dominant side of the participants. It was hypothesized that there would be differences in coordination and its variability between duration and surface of running. Specifically, increase in variability from indoor treadmill running to outdoor over-ground running. Furthermore, it was expected that the coordination variability would increase with the duration of running in response to the higher demand of task. Findings of this study would add to the understanding and knowledge of how the lower limb coordination and its variability varies with duration and surface of running among healthy individuals.

5.2 Methods

5.2.1 Study Design

This study used a crossover study design where same group of subjects participated in both phases of the study. A two-way repeated-measures mixed-design analysis of variance was conducted to investigate the effect of duration and surface of running on coordination and its variability of knee-hip, ankle-knee and ankle-hip joints couplings while long-distance running.

5.2.2 Participants

Eleven healthy adult regular runners who participated in the study mentioned in earlier chapter also participated in this study. The details regarding recruitment, inclusion and exclusion criteria, ethical approval, etc. can be found under methods section in chapter 3.

5.2.3 Procedure

This study was conducted in 2 phases in random order with enough washout period to separate each running session. Subjects were asked to run indoors on a treadmill (GE Marquette 2000, U.S.A.) in phase 1 while in phase 2 they had to run outdoors on an all-weather track over-ground for 31 minutes at their preferred speed (Furlong & Egginton, 2018). Both the phases were conducted on separate days in random order. Subjects had to wear 7 Opal Movement Monitoring IMUs (APDM Inc., Portland, OR, USA) while running (He et al., 2021; Lanovaz et al., 2017; Washabaugh et al., 2017). One sensor was tightly fixed on lower back at the level of L5S1 and two each on bilateral thighs, shank, and foot using elastic straps according to instructions provided by the manufacturers (figure 4.1). Data was wirelessly streamed to a computer at the sampling rate was 128 Hz. Subjects were asked to wear comfortable shoes and clothes and perform sufficient warm-up before each phase.

5.2.4 Data Processing

MATLAB (MathWorks BV, USA) was used to process raw unfiltered data obtained from the Mobility Lab software which is embedded within MoveoExplorer package provided with the Opal IMUs. Middle 30 min data were extracted to avoid any effects of initializing and ending of the recording. The initial (T1) and final (T2) 5 min were compared between the treadmill and over-ground running. Second order Butterworth low pass filter with cutoff frequency 10 Hz was used to remove high-frequency noise or unwanted signal components from data. Heel strike or initial contact was identified as the start of the gait cycle using the peak hip extension on the opposite hip joint (De Asha et al., 2012). Toe-off was identified using peak knee extension to determine stance and swing phases (Fellin et al., 2010). Each running cycle time series was interpolated to 101 data points for data compression and to maintain standardization and consistency.

The kinematic data was then used for CRP analysis throughout the gait cycle, stance and swing phases according to the method described earlier (Glazier, 2006; Hu et al., 2021; Miller et al., 2008; Mutchler et al., 2020; Robertson et al., 2013). A phase plane for each of the studied joints throughout a running cycle was constructed by plotting normalized angular positions (θ: x-axis) versus normalized angular velocities (ω: y-axis). Normalized angular position (θ_{inorm}) and angular velocity (ω_{inorm}) were calculated by following equations:

θ_{inorm}= $[2(\theta_i - min\ (\theta_i))/max\ (\theta_i) - min\ (\theta_i)]\ -1$

ω_{inorm}= $\omega_i/max\ (|\omega_i|)$

where i denotes each data point of the running cycle

The minimum and maximum angle/velocity for the series of all running cycles in each time interval while running on both surfaces were used to normalize the angular position θ and velocity ω for each of the 100 data points. Phase angle (Φ) was calculated as Φ = tan-1(ω/θ) along each data point of the running cycle. For each of the couplings, the phase angle Φ was obtained by calculating the four-quadrant arctangent angle relative to the right horizontal axis at each instant in the running cycle (figure 5.1). The CRP for data point i was calculated by subtracting the phase angle of the distal segment from that of the proximal segment and used as the coordination of two segments:

$CRP_i = \Phi_{idist} - \Phi_{iprox}$

where Φ_{idist} and Φ_{iprox} are respectively the phase angles of the distal and proximal joints at data point i in the running cycle

CRPv was calculated as the between-running cycle standard deviation of the CRP data points within all trials for all participants.

5.2.5 Parameters

a. CRP angle: CRP describes the coordination during a movement phase. The value of 0° indicates that the respective movements of the coupled joints or segments are perfectly in-phase i.e. both rotate simultaneously in same direction. CRP angle of 180° indicates that they are perfectly antiphase i.e. both rotate simultaneously in opposite direction. Any value between these indicate that they are out of phase but relatively in-phase (close to 0°) or antiphase (close to 180°).

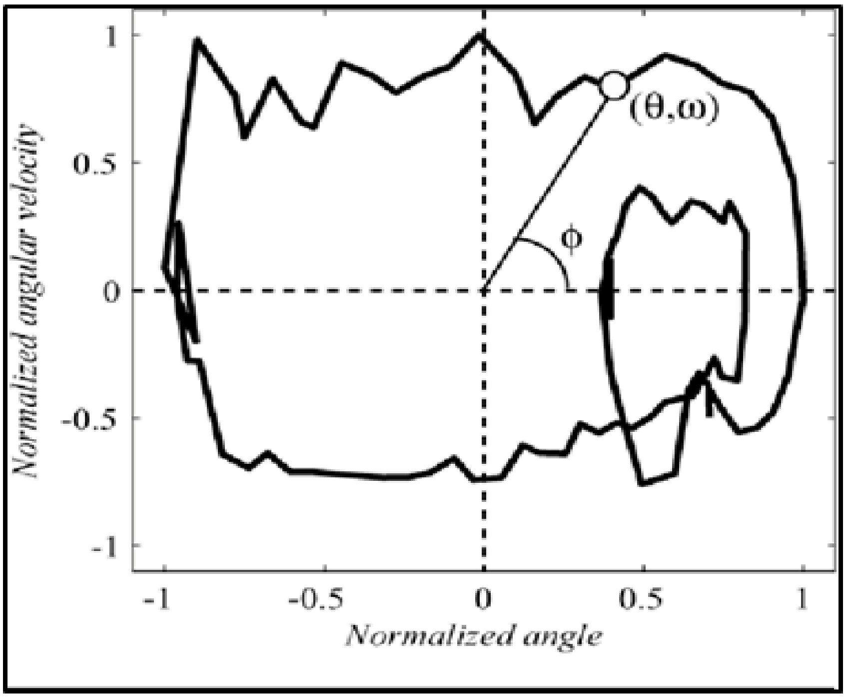


Figure 5.1: Miller, Ross H., et al. depiction of phase plot of two joints or segments with phase angle (Φ) calculated from each phase plane of normalized angular displacement (θ: x-axis) versus normalized angular velocities (ω: y-axis)

b. CRPv: CRPv represents the variability of the CRP angle. It is calculated by the mean SD of the CRP angles at each data point. CRPv represents the capacity of the neuromuscular system to provide stable movement. Lower CRPv indicates a more stable movement pattern or coordination. Three methods have been used to analyze and present data in previous studies that have been used in this study (Hein et al., 2012; Robertson et al., 2013). First is by calculating the average individual CRPv across the complete gait cycle for each participant in two-time intervals and compared between two surfaces of running. Second is by averaging the CRPv of each participant across stance and swing phases. Third is by displaying the CRPv continuously over the whole gait cycle by averaging the individual continuous standard deviations for the two surfaces of running at each point. For the last representation the calculation of CRPv was based on the CRP of five initial and last gait cycles from the T1 and T2 respectively for which standard deviations were calculated on a cycle to cycle basis.

5.2.6 Data Analysis

SPSS version 21 (IBM Corp., Chicago, IL, U.S.A) was used to conduct two-way 2 x 2 repeated-measures mixed-design analysis of variance to compare the CRP and CRPv of all the parameters between 2-time intervals (T1 and T2) and the 2 running surfaces (treadmill and over-ground) during long-distance running with statistical significance set at p less than 0.05. Post-hoc analysis was conducted using the Bonferroni correction.

5.3 Results

Mean and SD of CRP and CRPv of knee-hip, ankle-knee and ankle-hip joint couplings in sagittal plane motions while long-distance running on a treadmill and over-ground in both time intervals are presented in tables 4.1 and 4.2. There were few differences in CRP but almost equal amounts of CRPv for all tested joint couplings in both durations and surfaces of running. Averaged CRPv and standard error for both surfaces of running are graphically presented in figure 5.2. Similar overall patterns for all tested couplings were detected and no statistically significant differences between the two surfaces of running were found.

5.3.1 Comparison of within and between subject's lower limb coordination while prolonged running on a treadmill and over-ground

a. Overall gait cycle: There were no significant difference in CRP of tested lower limb joint couplings between initial and final 5 minutes or between treadmill and over-ground while long-distance running ($p>0.05$).

b. Stance phase: There were no significant difference in CRP of tested joint couplings between initial and final 5 minutes or between treadmill and over-ground while long-distance running ($p>0.05$) except ankle-knee coupling where coordination was significantly higher in final 5 minutes as compared to initial 5 minutes of running ($p<0.05$). There was simple

Table 5.1: Description of Mean and standard deviation (SD) of coordination (CRP) of Knee-Hip, Ankle-knee and Ankle-Hip couplings while long-distance running for both durations (T1: 0-5 and T2: 25-30 minutes) and surfaces (treadmill, TM and over-ground, OG) of running. Significantly higher parameters are denoted by bold text.

RUNNING SURFACE	**RUNNING DURATION**	
	T1	**T2**
OVERALL GAIT CYCLE		
Knee-Hip coupling		
TM	-6.05 (14.25)	-2.83 (14.55)
OG	-1.39 (14.91)	-4.66 (17.92)
Ankle-knee coupling		
TM	31.99 (25.03)	30.58 (25.90)
OG	34.58 (23.17)	38.85 (18.35)
Ankle-Hip coupling		
TM	25.94 (20.07)	27.74 (21.02)
OG	33.19 (20.72)	34.19 (17.37)
STANCE PHASE		
Knee-Hip coupling#		
TM	44.04 (26.46)	42.83 (33.32)
OG	58.12 (20.94)	33.39 (40.94)
Ankle-knee coupling*#		
TM	-9.21 (23.62)	**-8.02 (35.66)**
OG	-6.44 (46.14)	**17.57 (50.70)**
Ankle-Hip coupling		
TM	34.83 (13.51)	34.80 (11.65)
OG	51.67 (43.03)	50.97 (44.36)
SWING PHASE		
Knee-Hip coupling		
TM	-58.35 (9.00)	-51.97 (27.55)
OG	-60.86 (14.08)	-45.80 (41.65)
Ankle-knee coupling		
TM	37.29 (23.08)	45.69 (27.69)
OG	41.14 (33.14)	29.46 (40.36)
Ankle-Hip coupling*#		
TM	-21.06 (19.90)	**-6.27 (28.94)**
OG	-19.71 (34.76)	**-16.34 (34.44)**

*main effect of time, $p<0.05$ #interaction effect phase by time, $p<0.05$

main effect in knee-hip and ankle-knee couplings ($p<0.05$). In knee-hip coupling there was surface by duration interaction while over-ground running and mean coordination was significantly higher in final 5 minutes compared to initial 5 minutes ($p<0.05$) of running. In ankle-knee coupling there was also surface by duration interaction while over-ground running but the mean CRP was significantly higher in initial 5 minutes compared to final 5 minutes ($p<0.05$) of running.

c. Swing phase: There were no significant difference in CRP of tested joint couplings between initial and final 5 minutes or between treadmill and over-ground while long-distance running ($p>0.05$) except in ankle-hip coupling where CRP was significantly higher in final 5 minutes as compared to initial 5 minutes ($p<0.05$) of running. There was simple main effect in ankle-hip coupling with surface by duration interaction while treadmill running with mean CRP was significantly higher in final 5 minutes compared to initial 5 minutes ($p<0.05$) of running.

5.3.2 Comparison of within and between subject's lower limb coordination variability while prolonged running on a treadmill and over-ground

a. Overall gait cycle: There were no significant difference in CRPv of tested lower limb joint couplings between initial and final 5 minutes or between treadmill and over-ground while long-distance running ($p>0.05$).

b. Stance phase: There were no significant difference in CRPv of tested joint couplings between initial and final 5 minutes or between treadmill and over-ground while long-distance running ($p>0.05$).

c. Swing phase: There were no significant difference in CRPv of tested joint couplings between initial and final 5 minutes or between treadmill and over-ground while long-distance running ($p>0.05$).

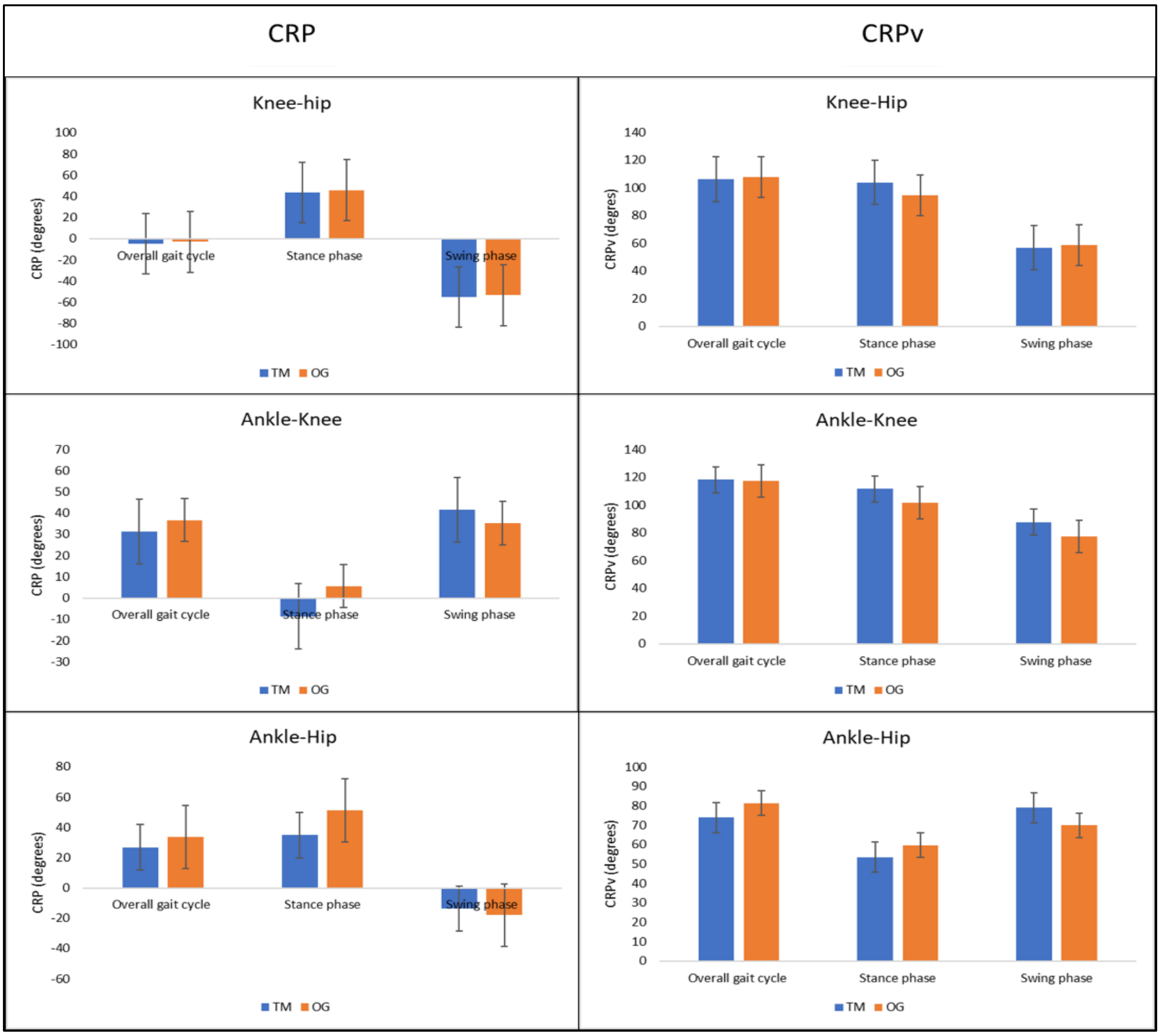


Figure 5.2: Comparison of coordination (CRP) and coordination variability (CRPv) of Knee-Hip, Ankle-knee and Ankle-Hip couplings while long-distance running on both surfaces (treadmill, TM and over-ground, OG) of running. Values are reported degrees as Mean and standard error (SEM). Note there are no significant differences between the two surfaces of running during overall gait cycle, stance and swing phases

5.3.3 Continuous coordination and coordination variability of tested joint couplings while prolonged running on a treadmill and over-ground

Continuous CRP and CRPv of knee-hip, ankle-knee and ankle-hip joint couplings in the sagittal plane movements are displayed in figures 5.3, 5.4 and 5.5 respectively. While treadmill running, knee-hip CRPv remained consistent until mid-cycle following which it increased until its maximum was reached by the end of the cycle. Similar pattern was seen while over-ground running but it was more irregular towards the end of running.

Table 5.2: Description of Mean and standard deviation (SD) of coordination variability (CRPv) of Knee-Hip, Ankle-knee and Ankle-Hip couplings while long-distance running for both durations (T1: 0-5 and T2: 25-30 minutes) and surfaces (treadmill, TM and over-ground, OG) of running.

RUNNING SURFACE	**RUNNING DURATION**	
	T1	**T2**
OVERALL GAIT CYCLE		
Knee-Hip coupling		
TM	107.84 (14.00)	104.45 (18.92)
OG	109.41 (18.96)	105.61 (15.39)
Ankle-knee coupling		
TM	116.06 (18.74)	120.29 (17.32)
OG	116.67 (21.60)	118.70 (12.89)
Ankle-Hip coupling		
TM	69.60 (17.48)	78.38 (27.68)
OG	82.55 (31.02)	80.17 (29.57)
STANCE PHASE		
Knee-Hip coupling		
TM	107.70 (26.50)	99.74 (24.28)
OG	92.45 (29.99)	96.24 (29.81)
Ankle-knee coupling		
TM	112.62 (29.92)	110.78 (28.57)
OG	97.12 (40.94)	105.82 (35.39)
Ankle-Hip coupling		
TM	47.94 (17.63))	59.25 (34.06)
OG	59.77 (25.12)	59.60 (23.39)
SWING PHASE		
Knee-Hip coupling		
TM	52.56 (7.63)	60.90 (16.18)
OG	60.31 (15.05)	56.47 (11.59)
Ankle-knee coupling		
TM	80.27 (41.15)	94.71 (36.89)
OG	76.25 (30.67)	78.18 (28.38)
Ankle-Hip coupling		
TM	70.95 (40.89)	87.01 (40.86)
OG	72.15 (34.72)	67.76 (28.78)

For ankle-knee coupling CRPv decreased after heel-strike which lasted until 10% of gait cycle following which it remained consistent to rise again at mid cycle to reach maximum. For ankle-hip coupling, CRPv decreased after heel strike in the initial gait cycle to remain consistently lower throughout the cycle to rise again at the end of the cycle just before toe-off. For ankle-knee and ankle-hip couplings, similar pattern was observed for both treadmill and over-ground running. All the three couplings exhibited more or less similar overall pattern in CRPv. However, the pattern was more irregular while over-ground running throughout the gait cycle. The differences in the graphs of continuous CRPv were noticeable but not significant statistically which may be due to the larger width of the confidence intervals.

5.4 Discussion

This part of study investigated the differences in coordination and coordination variability lower limb joint couplings in sagittal plane motions while long-distance running on a treadmill and over-ground among healthy runners. This is one of the first study to explore SSV in coordination during long-distance running in the natural environment outside the laboratory. The CRP and CRPv for knee-hip, ankle-knee, and ankle-hip joint couplings were compared between initial and final 5-minute duration of 30-minute run throughout the gait cycle and in stance and swing phases. On comparing overall gait cycle, there were no significant differences in coordination or its variability between initial and final 5 minutes or between treadmill and over-ground. While comparing stance phase, the coordination was significantly higher in the final 5 minutes as compared to initial 5 minutes in ankle-knee coupling. Additionally, there was simple main effect of surface by duration interaction in knee-hip coupling while over-ground running with mean coordination significantly higher in final 5 minutes as compared to initial 5 minutes duration of running. There was surface by duration interaction in ankle-knee coupling while over-ground running, with mean coordination significantly higher in the initial 5 minutes as compared to final 5 minutes duration of running. While comparing swing phase, the coordination was significantly higher in the final 5 minutes as compared to initial 5 minutes in ankle-hip coupling. There was simple main effect of surface by duration interaction in ankle-knee coupling while treadmill running with mean coordination significantly higher in final 5 minutes as compared to initial 5 minutes duration of running. There were no significant differences in the coordination variability between both durations and surface of running in overall gait cycle or stance and swing phases.

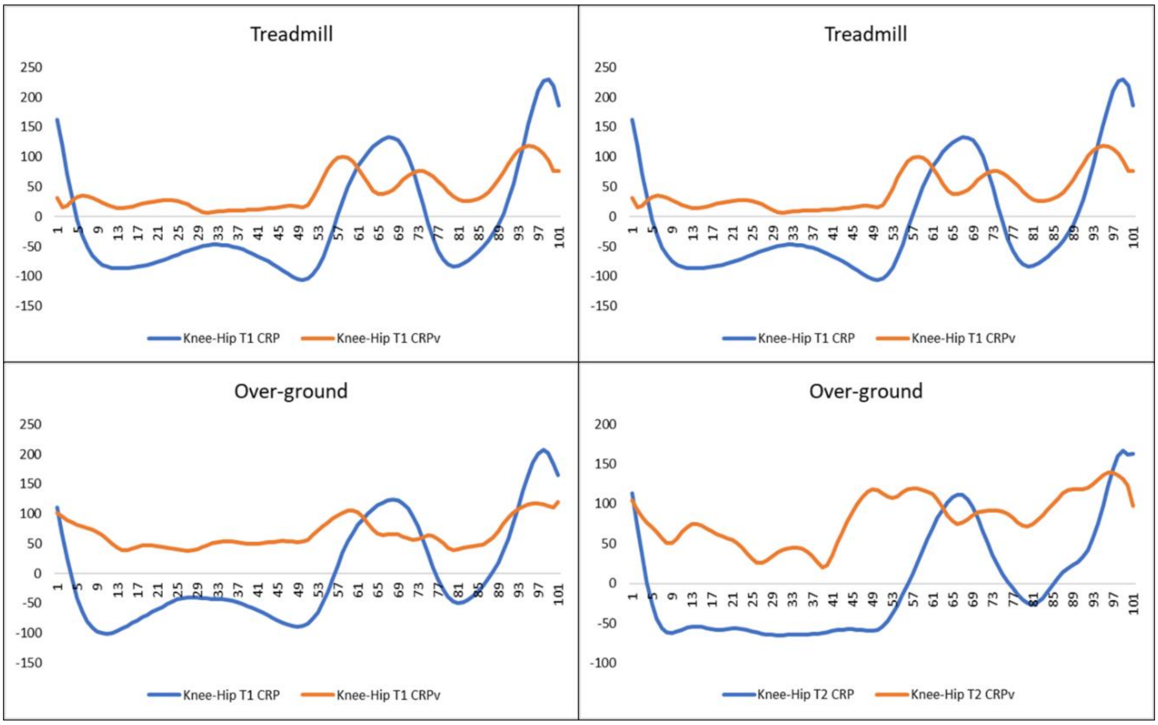


Figure 5.3: Continuous relative phase (CRP) and coordination variability (CRPv) of the Knee-Hip coupling in the sagittal plane movements averaged over the first (initial, T1) and the last (final, T2) five gait cycles while treadmill and over-ground running. Note that whole gait cycle is normalized to 101 data points. CRPv represents between subject variability at each point of stride for entire gait cycle.

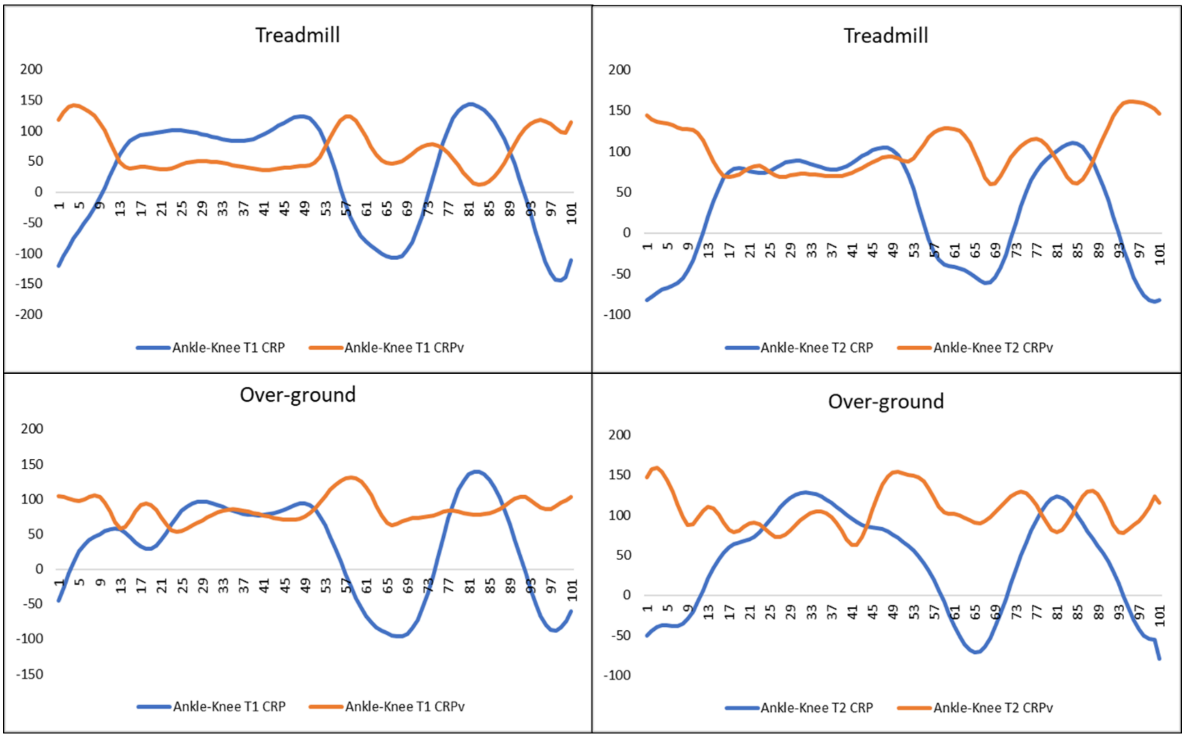


Figure 5.4: Continuous relative phase (CRP) and coordination variability (CRPv) of the Ankle-Knee coupling in the sagittal plane movements averaged over the first (initial, T1) and the last (final, T2) five gait cycles while treadmill and over-ground running. Note that whole gait cycle is normalized to 101 data points. CRPv represents between subject variability at each point of stride for entire gait cycle

It is big challenge to study biomechanical differences in the body under different conditions as it involves various complex systems working together in a precise and coordinated way (Stergiou, 2004). Dynamic system theory has been proposed as a better alternative and popularly used over the years to understand how different movements involving various joints and segments coordinate (Kelso, 1995). The dynamic human system has been quantified using CRP between joints or segment movements. Movement is accompanied by changes in displacement and velocity of joints or segments. The normalized segment or joint's angular velocity and displacement plot in a polar coordinate system provides phase portrait that can be used to detect changes in movement intuitively (Clark & Phillips, 1993). The phase angle, measured in degrees or radians, indicates the position of the joint or segment phase space or the relative position or timing difference between them. It represents the angular difference between the corresponding joints or segments at a given instant in time (Hein et al., 2012).

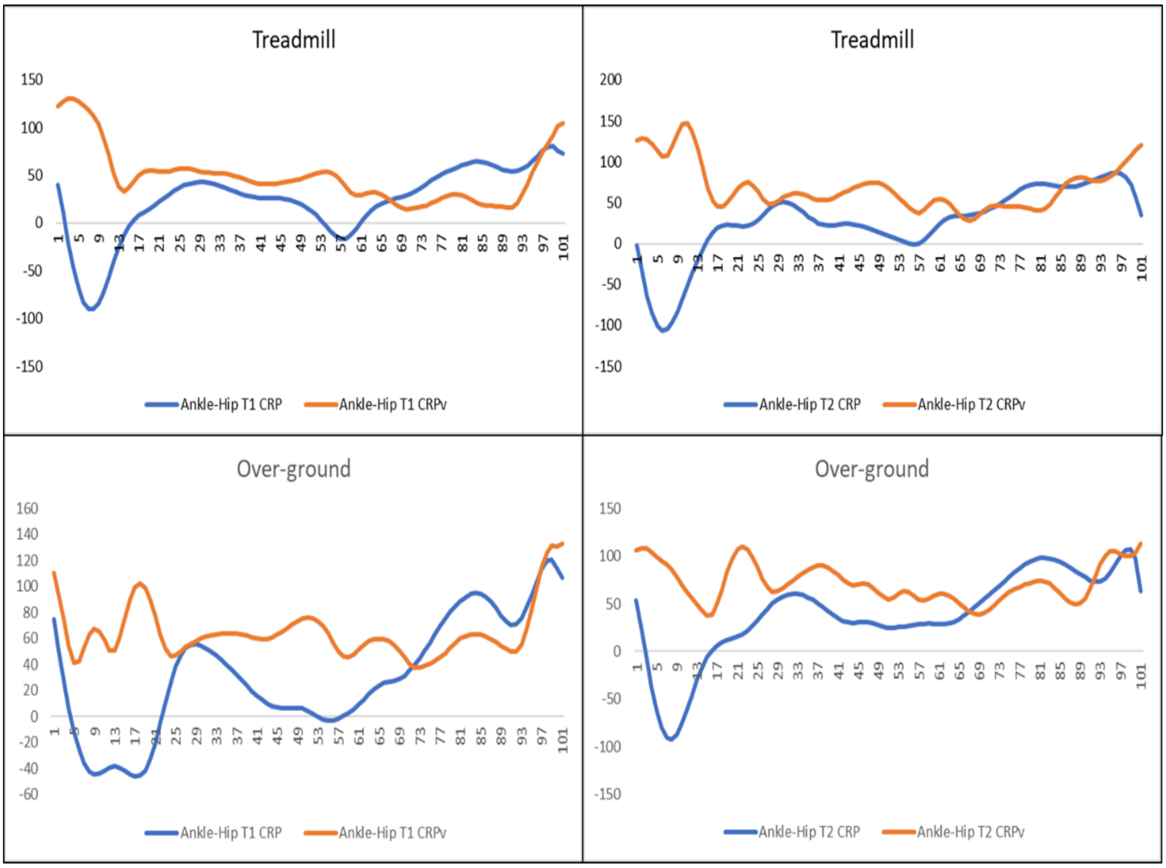


Figure 5.5: Continuous relative phase (CRP) and coordination variability (CRPv) of the Ankle-Hip coupling in the sagittal plane movements averaged over the first (initial, T1) and the last (final, T2) five gait cycles while treadmill and over-ground running. Note that whole gait cycle is normalized to 101 data points. CRPv represents between subject variability at each point of stride for entire gait cycle

CRP has been commonly used as an effective measure to assess performance of movement and identify coordination disorders (Winogrodzka et al., 2005). Available studies reporting CRP models have used variety of joint and segment relationships in different movement planes and data processing methods, e.g. filtering, etc., making interpretation difficult and inconsistent across different tasks (Hu et al., 2021). Some studies have investigated coordination patterns while walking but fewer researchers have studied coordination or its variability while running. Studying high intensity movements like running are important as they involve different coordination patterns due to different relationship between joint angles and velocities. CRP parameters can monitor differences in control mechanisms and responses to global, like environment or task-related, and local, like self-biological factors. CRP has often been used to quantify coordination between two couplings mostly in sagittal plane motions during whole gait cycle or stance phase only using digital motion capture systems or infrared cameras that limits it capacity to be used outside laboratories (Eslami et al., 2007; Floría et al., 2019; Gidley et al., 2020; Hu et al., 2021; Ko et al., 2017). Studies applying CRP variables in running have been conducted inside that laboratories and broadly focused on injuries, footwear, speed and level of running, Q-angle, age and gender differences and other test conditions (Frank et al., 2013; Hu et al., 2021; Prejean & Ricard, 2019). Current study is one of the first study to use IMUs to study the effect of duration and surface of running on SSV in coordination and its variability that has enabled to study long-distance running in natural environment.

Participants in the current study were asked to run at their preferred speed on both the surfaces. The results show that there were no significant differences in CRPv between the two durations and surfaces of running. Running speed is a key parameter that can alter kinetics and kinematics while running. In one of the previous studies, participants showed no significant differences in the lower limb CRP values while running on a treadmill at five different speeds, but the CRPv significantly decreased at a faster speed (Bailey, Silvernail, et al., 2018). These results indicate that CRPv is sensitive to changes in running speed. No differences in CRP and CRPv at the preferred speed shows that coordination is generally stable at the range of speeds that are comfortable to the runners. CRP has been inherently regarded sensitive to local or biological factors, like speed of running, and other global factors, like environment, task-related factors (Stergiou, 2004). Changes in any of these factors would lead to adaptation of sensorimotor system in order to achieve a state of new equilibrium (Kelso & Ding, 1993; Kelso, 1995). This helps to easily understand that different biological, task-related or environmental factors can influence coordination among individuals.

Progressive fatigue has been regarded as a common risk factor for development of RRIs especially among novice runners, however previous studies have studied fatigue induced by running a relatively shorter distance or protocols involving activities other than running (Bellenger et al., 2019; Panday et al., 2022). This is one of the first study to see the effect of fatigue induced by long-distance running on SSV of lower limb coordination pattern. The results show that CRP of ankle-knee, knee-hip and ankle-hip couplings were significantly higher in the final duration of running in both stance and swing phases. Experience and level of runners have been reported to alter their body's reaction to fatigue and in turn running performance (Bailey, Silvernail, et al., 2018; Floría et al., 2018; Pugh, 1970). As compared to novice runners, experienced runners have been shown to display significantly higher CRP and CRPv across different running speeds (Floría et al., 2018). They also showed no significant effect of fatigue protocol (not involving running) on CRP of thigh-shank and shank-foot couplings CRP while sprinting activities (Dal Pupo et al., 2017). CRP and CRPv have also been shown to change significantly during long-distance running with different fatigue levels (Bailey, Dufek, et al., 2018). So, fatigue plays an important role in maintaining coordination while continuous running.

Treadmills are commonly used to examine running coordination. As various biomechanical differences have been reported between treadmills with rigid and compliant surfaces, coordination has also been investigated to see the effect of running on a treadmill with different surfaces (Abbasi et al., 2020; Gidley et al., 2020). Running on treadmills with rigid surface resulted in higher lower limb CRP and lower limb CRP during the push-off phase. It also showed decreased lower limb CRPv during push-off and extension phases (Gidley et al., 2020).

Another study comparing over-ground running on a 10m track inside the laboratory with treadmill running revealed altered lower limb CRP but no change in CRPv (Abbasi et al., 2020). These results are in line with the results of current study showing no significant differences in lower limb CRPv between initial and final duration and between treadmill and over-ground running in overall gait cycle as well as stance and swing phases separately. Current study results also showed surface by time interaction in knee-hip and ankle-knee couplings with mean CRP significantly higher in final duration while over-ground running in stance phase. Modifications in the joint coordination pattern may be the response to the demands imposed by change in running conditions.

Among the studies considering coordination differences while running, most have considered complete gait cycle or stance phase (Floría et al., 2018; Foch & Milner, 2014; Hein et al., 2012; Hu et al., 2021). As there are biomechanical differences between stance and swing phases it is important to consider both while studying the effect of duration and surface of running on SSV. In stance phase, the lower limb acts as a closed kinetic chain to adjust the movement pattern while in the swing phase it is more like an open kinetic chain. As compared to open kinetic chain, activities involving closed kinetic chain could result in lower variability due to the decreased need for adaptability to changes in environment. When CRPv was analyzed between healthy and injured runners during complete gait cycle, higher differences were observed during the swing phase as compared to stance phase (Hamill et al., 1999). Current study also compared CRP and CRPv while swing phase and results were different from those obtained in stance phase. There was significant difference in coordination of ankle-hip coupling with coordination significantly higher in final duration as compared to initial duration of running. There was simple main effect with surface by duration interaction in ankle-hip coupling with mean coordination significantly higher in final duration of over-ground while treadmill running. However, similar to overall gait cycle and stance phase there were no significant difference in coordination variability of tested joint couplings between initial and final duration or between treadmill and over-ground running. Higher or lower coordination towards the end of the run is not the indicator of better performance but shows the ability of healthy participants to adapt various coordinative strategies for recovery during long-distance run (Bailey et al., 2020; Cazzola et al., 2016). Sensitivity and interpretation of CRP parameters can be increased by modeling the target based on joints or segments in a single movement plane, using sub-phases of the gait cycle, and investigating each discrete point of the full gait cycle (Abbasi et al., 2020; Floría et al., 2018).

Changes in the coordination variability is considered as the extent of adaptability to the new constraints of a motor task, therefore higher variability suggests higher adaptability (Preatoni et al., 2013). While continuous recording of the CRPv throughout the gait cycle peaks were observed in middle of the gait cycle that may coincide with transition from stance to swing phase. Another previous study has also shown similar peak of variability in the transition period from stance to swing phase (Floría et al., 2018), that may be relevant to changes in the running technique for propulsion. Maintenance of coordination at the early stance could explain the no change in loading rate (Chen et al., 2022). CRPv pattern was more irregular while over-ground running throughout the gait cycle. These peaks indicate the points of instability due to the higher demands while transition in comparison to the remaining gait cycle (Floría et al., 2019). These instances also point to important methodological considerations while analyzing coordination variability using average values obtained during longer periods that may not accurately describe the nature of the coordination variability while running. Throughout the gait cycle, the pattern of CRPv was changing continuously making it difficult to classify in-phase or antiphase and determine the degree of movement variability and see how a runner modify his variability when faced with a new task. These results also highlight that the coordination pattern and variability depend on the analyzed temporal phase of the gait. Further research is necessary analyzing continuous coordination variability across particular sub-phases in running. The analysis of continuous CRP and CRPv data shall help to further understand the organization and control of the gait cycle while running.

5.5 Summary

This part of study aimed to investigate the differences in coordination and coordination variability of lower limb joint movements in sagittal plane during long-distance running on a treadmill versus over-ground. This study is one of the first to explore coordination variability during long-distance running outside of a laboratory setting using IMUs. The study compared coordination and coordination variability for different joint couplings throughout the gait cycle and in stance and swing phases. Overall, there were no significant differences in coordination or its variability between the initial and final 5 minutes of running, or between treadmill and over-ground running. However, there were significant differences in coordination during specific phases and joint couplings. Ankle-knee coordination was higher in the final 5 minutes compared to the initial 5 minutes in the stance phase. Additionally, coordination was higher in the final 5 minutes compared to the initial 5 minutes in ankle-hip coupling during the swing phase. Fatigue induced by long-distance running was found to increase coordination in both stance and swing phases. The study also found that running surface (treadmill versus over-ground) influenced coordination in some joint couplings and phases. However, there were no significant differences in coordination variability based on duration or running surface. The study concluded that coordination patterns and their variability are influenced by factors such as fatigue and running surface, and further research is needed to understand the organization and control of the gait cycle during long-distance running in the natural environment outside the laboratory. The application of these findings can have significant implications for improving running performance, reducing injury risk, and enhancing overall biomechanical efficiency in athletes and recreational runners. By considering factors such as fatigue and running surface in training and performance settings, individuals can optimize their movement patterns and maximize their potential for success through injury prevention, better rehabilitation, equipment design and footwear development, and training monitoring and feedback.

Chapter 6: Limitations and Future Work

6.3 Limitations of the study

This research centered on examining the impact of running duration and surface type on SSV during long-distance running. The research underscored the necessity for longer-duration measurements outside of a laboratory setting to comprehend the impacts of prolonged running, yet current measurement systems may not be entirely reliable for such purposes. Running for more than 30 minutes would be needed to understand the effect of prolonged running, but we still lack the competent measurement systems that could record for longer duration outside the laboratory to provide reliable data. It is among the first studies to utilize IMUs for biomechanical analysis of long-distance running in real-world settings outside of a laboratory environment. While IMUs are valuable tools for studying biomechanics, there are concerns regarding issues like sensor drift, noise, and calibration errors that could potentially impact the accuracy of the data collected, indicating room for enhancement. Data drift was observed during the course of the run in this study, possibly due to sensor fixation and positioning problems. Fixation and positioning of sensors while prolonged running especially in the outdoor setting remains a concern. The study also encountered interference from running surfaces, particularly on treadmills, which affected ankle joint data for two participants. Interference caused by the running surface especially while running on a treadmill is a major concern while using IMUs (Meardon et al., 2011). Such interference mostly affects ankle joint sensors as they are closest to the ground (Kerdok et al., 2002).

This study was conducted on 11 subjects, but we were able to achieve a power of up to 0.99 on post hoc analysis using G-power, given alpha, sample size, and effect size. In addition, partial eta-squared effect sizes indicated medium (0.06) to large (0.14) effect size for different parameters. This study focused solely on sagittal plane motions in a single 30-minute running session rather than a fixed distance protocol among healthy participants. The participants were asked to run at the preferred running speed rather than a consistent speed to ensure a more natural gait pattern that could possibly affect overall biomechanical differences (Abbasi et al., 2020). Running at a constant speed also decreases the tendency to adapt to sudden environmental changes and the range of variability (Boyer et al., 2017; Hafer et al., 2016; Nakayama et al., 2010). However, although preferred speed can be fixed over treadmill it cannot be controlled while running over-ground. Uncontrolled running speed can influence data of stride time, lower limb joint ranges and angles. Different runners may react differently to long-distance running, particularly novice runners who may experience fatigue earlier. It is essential to recognize that individual characteristics such as body composition, muscle strength, and running technique can vary among participants, potentially influencing overall biomechanical differences. Various parameters such as stride time, lower limb joint angles, and coordination during different running phases were measured using linear metrics, with the CV preferred over SD as it depicts the changes in SD normalized to the mean value (Fadillioglu et al., 2022). However, some CV values were excessively high due to low mean values and SDs.

CRP analysis, a common method for quantifying coordination patterns in human movement, was utilized in the study but has limitations to consider. CRP analysis can be sensitive to noise and measurement errors, with small data variations or inaccuracies potentially leading to significant fluctuations in calculated CRP values, making it difficult to detect subtle changes in coordination patterns, particularly with noisy or low-quality data. While CRP analysis focuses on the temporal relationship between movement components, it may not capture other aspects of coordination dynamics such as changes in movement amplitude, speed, or coupling strength, potentially missing complex temporal coordination changes over time and not revealing causal relationships or underlying mechanisms driving observed coordination patterns. The explanations regarding differences in SSV should not be restricted to inherent biological mechanisms and should incorporate a holistic system interaction involving body, performance of task and the environment (Lindsay et al., 2014).

6.4 Future work

It is crucial to consider the above limitations while interpreting and applying the findings of this study and conceptualizing future studies. Future directions building on the findings, implications, and limitations of the current study, future research endeavors should explore a range of topics to further enhance our understanding and application of the insights gained regarding SSV during long-distance running. Potential areas for future investigation include:

a. Surface-Specific Training Interventions: Designing training programs tailored to address the differences in SSV between treadmill and over-ground running to facilitate smoother transitions for runners between indoor and outdoor settings. This would help in evaluating the impact of running duration and surface on variability, injury susceptibility, and performance, particularly in natural surroundings.

b. Population-Specific Studies: Investigating SSV in individuals with locomotor impairments to explore variations in SSV and coordination across diverse groups. Considering factors such as training level, experience, foot structure, footwear, speed, injury history, and other variables among runners is crucial when examining the impact of running duration and surface. Inclusion of a diverse participant pool can offer a comprehensive insight into how running duration and surface affect spatial variability in gait coordination. Additionally, considering gender, experience, and fatigue induced by long-distance running can provide further valuable insights.

c. Injury Prediction and Prevention: Refining injury prediction models using IMU data. Longitudinal studies could monitor athletes over time to validate and enhance injury prediction algorithms, potentially incorporating artificial intelligence and machine learning techniques.

d. Technological Advancements: Further refining and validating wearable technologies like IMUs for prolonged outdoor use. This could involve enhancing sensor accuracy, fixation and placement, data analysis approaches, and user-friendliness for athletes and coaches.

e. SSV across Motions in Different Planes: Exploring SSV in joint angles and coordination in motions beyond the sagittal plane to deepen our understanding of the effects of running duration and surface on spatial variability in gait coordination.

f. Running Speed and SSV: Exploring the impact of varying running speeds on SSV during long-distance over-ground running in natural settings to determine how speed influences spatial variability in gait coordination.

g. Running Kinetics and SSV: Incorporating additional biomechanical markers such as ground reaction forces, joint moments, and muscle activations alongside stride time, joint angles, and coordination to enhance the understanding of biomechanical changes during long-distance running.

h. Biomechanical Analysis in Varied Conditions: Considering external factors like weather conditions, terrain variations, and footwear types alongside running surface in future studies to comprehensively assess their impact on spatial variability in gait coordination and how they influence SSV and coordination compared to controlled laboratory environments.

i. Longitudinal and Cross-Sectional Studies: Conducting longitudinal studies tracking participants over an extended period to observe how spatial variability in gait coordination evolves with repeated long-distance running.

j. Fatigue Management: There are different reactions among runners to long-distance running especially among the novice runners who may fatigue earlier. Investigating the effects of diverse fatigue management strategies on SSV and joint coordination to understand how gender, running experience and level of runners, rest intervals, nutrition, hydration, and recovery methods influence gait variability and performance while long-distance running.

k. Optimal Variability Range: Determining the ideal range of SSV for individual runners based on their personal biomechanics, injury history, and performance objectives.

l. Cross-Disciplinary Approaches: Integrating insights from various fields like kinesiology, sports science, physical therapy, engineering, and computer science to develop comprehensive models of runner health and performance.

m. Non-linear analytical methods: Implementation of non-linear measures for data collection and analysis to measure SSV in gait analysis for enhanced insights, comprehensive and nuanced data interpretation.

Chapter 7: Conclusion and Practical Implications

Long-distance running, as seen in marathons, have both health benefits and an increased risk of RRIs due to the strain on joints. Understanding how the body responds to different running environments is crucial for preventing injuries in long-distance runners. This doctoral thesis focuses on SSV during prolonged running, specifically looking at how SSV changes with running duration and surface type (treadmill vs. over-ground) using IMUs in sagittal plane motions. The study hypothesizes that SSV varies with running duration and that there are differences in SSV between treadmill and over-ground running, with higher SSV expected in over-ground running. It also predicts an interaction between running duration and surface type. This section summarizes how this research addressed some of the research gaps highlighted in Chapter 1.

While extensive research has focused on gait parameters while treadmill running, limited work has addressed SSV particularly in joint angles and coordination comparing treadmill versus over-ground running. By examining SSV in stride time, lower limb joint angles, and joint coordination in natural running environments, this study seeks to provide insights into gait motor control under fatigue and tailor interventions for injury prevention and performance improvement for different groups of runners. This study is distinctive because it is one of the first to examine SSV in the natural environment outside the laboratory using IMUs considering the factors like running duration and surface. The outcomes of this research could be instrumental in understanding gait motor control regulations, biomechanics of running, and developing strategies to prevent RRIs, enhance performance, and customize interventions for various runner groups to aid in rehabilitation.

6.1 Key findings

The results of this study indicate that there were no significant differences in stride time or its variability between the two running durations or surfaces. However, there was higher SSV in joint angles observed during over-ground running and in the initial duration of run, which aligns with the hypothesis. This suggests that while treadmill studies provide a controlled setting, they may not fully capture the complexities of outdoor running, which could impact training and rehabilitation strategies. A sub-study focused on analyzing lower limb joint coordination and its variability during long-distance running found significant differences in joint coordination at certain gait phases. Ankle-knee coordination was higher in the final duration of running as compared to the initial duration in the stance phase. Coordination was higher in the final duration in ankle-hip coupling during the swing phase. Fatigue induced by long-distance running was found to increase coordination in both stance and swing phases. The study also found that running surface (treadmill versus over-ground) influenced coordination in some joint couplings and phases, but the coordination variability was not significantly impacted by duration or surface of running. These findings challenge the assumption that treadmill research accurately represents the dynamics of outdoor running and stress the importance of considering gait variability in research, training, and clinical applications. While the sample size in this study may be relatively small, the findings provide a proof-of-concept for using IMUs in outdoor settings and highlight the potential for future research to collect long-term data on running biomechanics. Larger-scale studies with diverse populations and running conditions can further validate the use of IMUs in outdoor settings and enhance our understanding of the complexities of human movement in natural environments. The norm for future studies involving long-term data recordings using IMUs will likely be to consider the insights and considerations provided by this study to ensure robust and reliable data collection and analysis. This study suggests a comprehensive approach that considers the interaction between the body and the environment when explaining variations in spatial-temporal gait parameters. Overall, the study underscores the significance of investigating gait organization and control during long-distance running in natural settings outside the laboratory. Despite the study's small sample size of 11 participants, post hoc analysis using G-power indicated a high statistical power of up to 0.99, suggesting that the findings are both generalizable and robust.

References

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

Abhayasinghe, N., Murray, I., & Sharif Bidabadi, S. (2019). Validation of thigh angle estimation using inertial measurement unit data against optical motion capture systems. *Sensors*, *19*(3), 596.

Adams, D., Pozzi, F., Carroll, A., Rombach, A., & Zeni Jr, J. (2016). Validity and reliability of a commercial fitness watch for measuring running dynamics. *journal of orthopaedic & sports physical therapy*, *46*(6), 471-476.

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

Al-Amri, M., Nicholas, K., Button, K., Sparkes, V., Sheeran, L., & Davies, J. L. (2018). Inertial measurement units for clinical movement analysis: Reliability and concurrent validity. *Sensors*, *18*(3), 719.

Alton, F., Baldey, L., Caplan, S., & Morrissey, M. (1998). A kinematic comparison of overground and treadmill walking. *Clinical Biomechanics*, *13*(6), 434-440.

Asmussen, M. J., Kaltenbach, C., Hashlamoun, K., Shen, H., Federico, S., & Nigg, B. M. (2019). Force measurements during running on different instrumented treadmills. *Journal of biomechanics*, *84*, 263-268.

Baida, S. R., Gore, S. J., Franklyn-Miller, A. D., & Moran, K. A. (2018). Does the amount of lower extremity movement variability differ between injured and uninjured populations?

A systematic review. *Scandinavian journal of medicine & science in sports*, *28*(4), 1320-1338.

Bailey, J. P., Dufek, J. S., Freedman Silvernail, J., Navalta, J., & Mercer, J. (2020). Understanding the influence of perceived fatigue on coordination during endurance running. *Sports Biomechanics*, *19*(5), 618-632.

Bailey, J. P., Dufek, J. S., Silvernail, J. F., Navalta, J., & Mercer, J. (2018). Understanding the influence of perceived fatigue on coordination during endurance running. *Sports Biomechanics*.

Bailey, J. P., Silvernail, J. F., Dufek, J. S., Navalta, J., & Mercer, J. A. (2018). Effects of treadmill running velocity on lower extremity coordination variability in healthy runners. *Human movement science*, *61*, 144-150.

Bartlett, R., Wheat, J., & Robins, M. (2007). Is movement variability important for sports biomechanists? *Sports Biomechanics, 6*(2), 224-243.

Beauchet, O., Allali, G., Annweiler, C., Bridenbaugh, S., Assal, F., Kressig, R. W., & Herrmann, F. R. (2009). Gait variability among healthy adults: low and high stride-to-stride variability are both a reflection of gait stability. *Gerontology*, *55*(6), 702-706.

Beauchet, O., Allali, G., Berrut, G., & Dubost, V. (2007). Is low lower-limb kinematic variability always an index of stability? *Gait & posture*, *2*(26), 327-328.

Beauchet, O., Annweiler, C., Lecordroch, Y., Allali, G., Dubost, V., Herrmann, F. R., & Kressig, R. W. (2009). Walking speed-related changes in stride time variability: effects of decreased speed. *Journal of neuroengineering and rehabilitation*, *6*(1), 1-6.

Beauchet, O., Dubost, V., Herrmann, F. R., & Kressig, R. W. (2005). Stride-to-stride variability while backward counting among healthy young adults. *Journal of neuroengineering and rehabilitation*, *2*(1), 1-8.

Bellenger, C. R., Arnold, J. B., Buckley, J. D., Thewlis, D., & Fuller, J. T. (2019). Detrended fluctuation analysis detects altered coordination of running gait in athletes following a heavy period of training. *Journal of science and medicine in sport*, *22*(3), 294-299.

Belli, A., Lacour, J., Komi, P., Candau, R., & Denis, C. (1995). Mechanical step variability during treadmill running. *European journal of applied physiology and occupational physiology*, *70*(6), 510-517.

Benson, L. C., Clermont, C. A., Bošnjak, E., & Ferber, R. (2018). The use of wearable devices for walking and running gait analysis outside of the lab: A systematic review. *Gait & posture*, *63*, 124-138.

Berner, K., Cockcroft, J., Morris, L. D., & Louw, Q. (2020). Concurrent validity and within-session reliability of gait kinematics measured using an inertial motion capture system with repeated calibration. *Journal of Bodywork and Movement Therapies*, *24*(4), 251-260.

Bernstein, N. (1967). The coordination and regulation of movements. *(No Title)*.

Berryman, N., Gayda, M., Nigam, A., Juneau, M., Bherer, L., & Bosquet, L. (2012). Comparison of the metabolic energy cost of overground and treadmill walking in older adults. *European journal of applied physiology*, *112*, 1613-1620.

Bertelsen, M., Hulme, A., Petersen, J., Brund, R. K., Sørensen, H., Finch, C., Parner, E. T., & Nielsen, R. (2017). A framework for the etiology of running-related injuries. *Scandinavian journal of medicine & science in sports*, *27*(11), 1170-1180.

Bilney, B., Morris, M., & Webster, K. (2003). Concurrent related validity of the GAITRite® walkway system for quantification of the spatial and temporal parameters of gait. *Gait & posture*, *17*(1), 68-74.

Bishop, C., Hillier, S., & Thewlis, D. (2017). The reliability of the Adelaide in-shoe foot model. *Gait & posture*, *56*, 1-7.

Błażkiewicz, M., Wiszomirska, I., & Wit, A. (2014). Comparison of four methods of calculating the symmetry of spatial-temporal parameters of gait. *Acta of bioengineering and biomechanics*, *16*(1).

Böhm, H., & Döderlein, L. (2012). Gait asymmetries in children with cerebral palsy: do they deteriorate with running? *Gait & posture*, *35*(2), 322-327.

Bonato, P. (2005). Advances in wearable technology and applications in physical medicine and rehabilitation. In (Vol. 2, pp. 1-4): BioMed Central.

Bovalino, S. (2021). *The Effect of Long-Distance Overground Running on Foot Strike Patterns and Performance* La Trobe].

Boyer, K. A., Freedman Silvernail, J., & Hamill, J. (2017). Age and sex influences on running mechanics and coordination variability. *Journal of Sports Sciences*, *35*(22), 2225-2231.

Boyer, K. A., Silvernail, J. F., & Hamill, J. (2014). The role of running mileage on coordination patterns in running. *Journal of Applied Biomechanics*, *30*(5), 649-654.

Brach, J. S., Berlin, J. E., VanSwearingen, J. M., Newman, A. B., & Studenski, S. A. (2005). Too much or too little step width variability is associated with a fall history in older persons who walk at or near normal gait speed. *Journal of neuroengineering and rehabilitation*, *2*(1), 1-8.

Brach, J. S., Berthold, R., Craik, R., VanSwearingen, J. M., & Newman, A. B. (2001). Gait variability in community-dwelling older adults. *Journal of the American Geriatrics Society*, *49*(12), 1646-1650.

Brahms, C. M., Zhao, Y., Gerhard, D., & Barden, J. M. (2022). Long-range correlations and stride pattern variability in recreational and elite distance runners during a prolonged run. *Gait & posture*, *92*, 487-492.

Breine, B., Malcolm, P., Van Caekenberghe, I., Fiers, P., Frederick, E. C., & De Clercq, D. (2017). Initial foot contact and related kinematics affect impact loading rate in running. *Journal of Sports Sciences*, *35*(15), 1556-1564.

Brice, S. M., Phillips, E. J., Millett, E. L., Hunter, A., & Philippa, B. (2020). Comparing inertial measurement units and marker-based biomechanical models during dynamic rotation of the torso. *European Journal of Sport Science*, *20*(6), 767-775.

Brindle, R. A., Taylor, J. B., Rajek, C., Weisbrod, A., & Ford, K. R. (2020). Association between temporal spatial parameters and overuse injury history in runners: a systematic review and meta-analysis. *Sports medicine*, *50*, 331-342.

Brown, A. M., Zifchock, R. A., Hillstrom, H. J., Song, J., & Tucker, C. A. (2016). The effects of fatigue on lower extremity kinematics, kinetics and joint coupling in symptomatic female runners with iliotibial band syndrome. *Clinical Biomechanics*, *39*, 84-90.

Bruijn, S. M., Meijer, O., Beek, P., & van Dieen, J. H. (2013). Assessing the stability of human locomotion: a review of current measures. *Journal of the Royal Society Interface*, *10*(83), 20120999.

Burke, L. M., & Hawley, J. A. (2002). Effects of short-term fat adaptation on metabolism and performance of prolonged exercise. *Medicine and science in sports and exercise*, *34*(9), 1492-1498.

Bussmann, J. B., Ebner-Priemer, U. W., & Fahrenberg, J. (2009). Ambulatory activity monitoring: Progress in measurement of activity, posture, and specific motion patterns in daily life. *European Psychologist*, *14*(2), 142-152.

Byun, S., Han, J. W., Kim, T. H., Kim, K., Kim, T. H., Park, J. Y., Suh, S. W., Seo, J. Y., So, Y., & Lee, K. H. (2018). Gait variability can predict the risk of cognitive decline in cognitively normal older people. *Dementia and geriatric cognitive disorders*, *45*, 251-261.

Callen, K. E. (1983). Mental and emotional aspects of long-distance running. *Psychosomatics*, *24*(2), 133-151.

Cappa, D., García, G., Secchi, J., & Maddigan, M. (2014). The relationship between an athlete's maximal aerobic speed determined in a laboratory and their final speed reached during a field test (UNCa Test). *J Sports Med Phys Fitness*, *54*(4), 424-431.

Cazzola, D., Pavei, G., & Preatoni, E. (2016). Can coordination variability identify performance factors and skill level in competitive sport? The case of race walking. *Journal of Sport and Health Science*, *5*(1), 35-43.

Chakraborty, J., Upadhyay, S., & Nandy, A. (2022). Musculoskeletal Injury Recovery Assessment using Gait analysis with ground reaction force sensor. *Medical engineering & physics*, *103*, 103788.

Chan-Roper, M., Hunter, I., Myrer, J. W., Eggett, D. L., & Seeley, M. K. (2012). Kinematic changes during a marathon for fast and slow runners. *Journal of sports science & medicine*, *11*(1), 77.

Chau, T. (2001). A review of analytical techniques for gait data. Part 1: fuzzy, statistical and fractal methods. *Gait & posture*, *13*(1), 49-66.

Chen, T. L.-W., Wong, D. W.-C., Wang, Y., Tan, Q., Lam, W.-K., & Zhang, M. (2022). Changes in segment coordination variability and the impacts of the lower limb across running mileages in half marathons: Implications for running injuries. *Journal of Sport and Health Science*, *11*(1), 67-74.

Chow, D. H.-K., Iqbal, Z. A., Tremblay, L., Lam, C.-Y., & Zhao, R.-B. (2022). Cross-Leg Prediction of Running Kinematics across Various Running Conditions and Drawing from a Minimal Data Set Using a Single Wearable Sensor. *Symmetry*, *14*(6), 1092.

Chow, D. H. K., Tremblay, L., Lam, C. Y., Yeung, A. W. Y., Cheng, W. H. W., & Tse, P. T. W. (2021). Comparison between accelerometer and gyroscope in predicting level-ground running kinematics by treadmill running kinematics using a single wearable sensor. *Sensors*, *21*(14), 4633.

Cicirelli, G., Impedovo, D., Dentamaro, V., Marani, R., Pirlo, G., & D'Orazio, T. R. (2021). Human gait analysis in neurodegenerative diseases: a review. *IEEE journal of biomedical and health informatics*, *26*(1), 229-242.

Clark, J. E., & Phillips, S. J. (1993). A longitudinal study of intralimb coordination in the first year of independent walking: a dynamical systems analysis. *Child development*, *64*(4), 1143-1157.

Clemente, F. M., Akyildiz, Z., Pino-Ortega, J., & Rico-González, M. (2021). Validity and reliability of the inertial measurement unit for barbell velocity assessments: A systematic review. *Sensors*, *21*(7), 2511.

Clermont, C. A., Pohl, A. J., & Ferber, R. (2019). Fatigue-related changes in running gait patterns persist in the days following a marathon race. *Journal of Sport Rehabilitation*, *29*(7), 934-941.

Combs-Miller, S. A., Kalpathi Parameswaran, A., Colburn, D., Ertel, T., Harmeyer, A., Tucker, L., & Schmid, A. A. (2014). Body weight-supported treadmill training vs. overground walking training for persons with chronic stroke: a pilot randomized controlled trial. *Clinical rehabilitation*, *28*(9), 873-884.

Cooper, P. (1998). *The American Marathon*. Syracuse University Press.

Corbeil, P., Blouin, J.-S., Bégin, F., Nougier, V., & Teasdale, N. (2003). Perturbation of the postural control system induced by muscular fatigue. *Gait & posture*, *18*(2), 92-100.

Costa, M., Goldberger, A. L., & Peng, C.-K. (2005). Multiscale entropy analysis of biological signals. *Physical review E, 71*(2), 021906.

Dahl, K. D., Dunford, K. M., Wilson, S. A., Turnbull, T. L., & Tashman, S. (2020). Wearable sensor validation of sports-related movements for the lower extremity and trunk. *Medical Engineering & Physics*, *84*, 144-150.

Dal Pupo, J., Detanico, D., Ache-Dias, J., & Santos, S. G. d. (2017). The fatigue effect of a simulated futsal match protocol on sprint performance and kinematics of the lower limbs. *Journal of Sports Sciences, 35*(1), 81-88.

Daniels, J. (2013). *Daniels' running formula*. Human Kinetics.

Davids, K., Glazier, P., Araújo, D., & Bartlett, R. (2003a). Movement systems as dynamical systems. *Sports medicine, 33*(4), 245-260.

Davids, K., Glazier, P., Araújo, D., & Bartlett, R. (2003b). Movement systems as dynamical systems: the functional role of variability and its implications for sports medicine. *Sports medicine*, *33*, 245-260.

De Asha, A., Robinson, M., & Barton, G. (2012). A marker based kinematic method of identifying initial contact during gait suitable for use in real-time visual feedback applications. *Gait & posture*, *36*(3), 650-652.

DeJong, P., Hatamiya, N. S., & Barkley, L. C. (2022). Running gait analysis and biomechanics. *Current sports medicine reports*, *21*(4), 107-108.

DeLeo, A. T., Dierks, T. A., Ferber, R., & Davis, I. S. (2004). Lower extremity joint coupling during running: a current update. *Clinical Biomechanics*, *19*(10), 983-991.

Delignieres, D., Ramdani, S., Lemoine, L., Torre, K., Fortes, M., & Ninot, G. (2006). Fractal analyses for 'short'time series: a re-assessment of classical methods. *Journal of mathematical psychology*, *50*(6), 525-544.

Dempster, J., Dutheil, F., & Ugbolue, U. C. (2021). The prevalence of lower extremity injuries in running and associated risk factors: a systematic review. *Physical Activity and Health*, *5*(1).

Dierks, T. A., Davis, I. S., & Hamill, J. (2010). The effects of running in an exerted state on lower extremity kinematics and joint timing. *Journal of biomechanics, 43*(15), 2993-2998.

Dingwell, J., Cusumano, J. P., Cavanagh, P., & Sternad, D. (2001). Local dynamic stability versus kinematic variability of continuous overground and treadmill walking. *J. Biomech. Eng., 123*(1), 27-32.

Dingwell, J. B., & Cusumano, J. P. (2000). Nonlinear time series analysis of normal and pathological human walking. *Chaos: An Interdisciplinary Journal of Nonlinear Science, 10*(4), 848-863.

Dingwell, J. B., & Marin, L. C. (2006). Kinematic variability and local dynamic stability of upper body motions when walking at different speeds. *Journal of biomechanics, 39*(3), 444-452.

Diss, C. E., & Parmar, A. (2021). Performance Decline in Master Endurance Runners. In *The Science and Practice of Middle and Long Distance Running* (pp. 328-341). Routledge.

Dixon, S. J., Collop, A. C., & Batt, M. E. (2000). Surface effects on ground reaction forces and lower extremity kinematics in running. *Medicine & Science in Sports & Exercise, 32*(11), 1919-1926.

Dobkin, B. H., & Dorsch, A. (2011). The promise of mHealth: daily activity monitoring and outcome assessments by wearable sensors. *Neurorehabilitation and neural repair, 25*(9), 788-798.

Dubost, V., Kressig, R. W., Gonthier, R., Herrmann, F. R., Aminian, K., Najafi, B., & Beauchet, O. (2006). Relationships between dual-task related changes in stride velocity and stride time variability in healthy older adults. *Human movement science, 25*(3), 372-382.

Ducharme, S. W., Liddy, J. J., Haddad, J. M., Busa, M. A., Claxton, L. J., & van Emmerik, R. E. (2018). Association between stride time fractality and gait adaptability during unperturbed and asymmetric walking. *Human movement science*, *58*, 248-259.

Dugan, S. A., & Bhat, K. P. (2005). Biomechanics and analysis of running gait. *Physical Medicine and Rehabilitation Clinics*, *16*(3), 603-621.

Edwards, R. B., Tofari, P. J., Cormack, S. J., & Whyte, D. G. (2017). Non-motorized treadmill running is associated with higher cardiometabolic demands compared with overground and motorized treadmill running. *Frontiers in physiology*, 914.

Eslami, M., Begon, M., Farahpour, N., & Allard, P. (2007). Forefoot–rearfoot coupling patterns and tibial internal rotation during stance phase of barefoot versus shod running. *Clinical Biomechanics*, *22*(1), 74-80.

Favre, J., Jolles, B. M., Aissaoui, R., & Aminian, K. (2008). Ambulatory measurement of 3D knee joint angle. *Journal of biomechanics*, *41*(5), 1029-1035.

Fellin, R. E., Rose, W. C., Royer, T. D., & Davis, I. S. (2010). Comparison of methods for kinematic identification of footstrike and toe-off during overground and treadmill running. *Journal of science and medicine in sport*, *13*(6), 646-650.

Ferber, R., Hreljac, A., & Kendall, K. D. (2009). Suspected mechanisms in the cause of overuse running injuries: a clinical review. *Sports Health*, *1*(3), 242-246.

Ferrari, A., Cutti, A. G., Garofalo, P., Raggi, M., Heijboer, M., Cappello, A., & Davalli, A. (2010). First in vivo assessment of "Outwalk": a novel protocol for clinical gait analysis based on inertial and magnetic sensors. *Medical & biological engineering & computing*, *48*, 1-15.

Fields, K. B., Sykes, J. C., Walker, K. M., & Jackson, J. C. (2010). Prevention of running injuries. *Current sports medicine reports*, *9*(3), 176-182.

Finch, C. F., & Cook, J. (2014). Categorising sports injuries in epidemiological studies: the subsequent injury categorisation (SIC) model to address multiple, recurrent and exacerbation of injuries. *British journal of sports medicine*, *48*(17), 1276-1280.

Firminger, C. R., Vernillo, G., Savoldelli, A., Stefanyshyn, D. J., Millet, G. Y., & Edwards, W. B. (2018). Joint kinematics and ground reaction forces in overground versus treadmill graded running. *Gait & posture*, *63*, 109-113.

Floría, P., Sánchez-Sixto, A., Ferber, R., & Harrison, A. J. (2018). Effects of running experience on coordination and its variability in runners. *Journal of Sports Sciences*, *36*(3), 272-278.

Floría, P., Sánchez-Sixto, A., Harrison, A. J., & Ferber, R. (2019). The effect of running speed on joint coupling coordination and its variability in recreational runners. *Human movement science*, *66*, 449-458.

Foch, E., & Milner, C. E. (2014). Frontal plane running biomechanics in female runners with previous iliotibial band syndrome. *Journal of Applied Biomechanics*, *30*(1), 58-65.

Folland, J. P., Allen, S. J., Black, M. I., Handsaker, J. C., & Forrester, S. E. (2017). Running technique is an important component of running economy and performance. *Medicine and science in sports and exercise*, *49*(7), 1412.

Frank, N. S., Callaghan, J. P., & Prentice, S. D. (2013). Lower limb kinematic variability associated with minimal footwear during running. *Footwear Science*, *5*(3), 171-177.

Frishberg, B. A. (1983). An analysis of overground and treadmill sprinting. *Medicine and science in sports and exercise*, *15*(6), 478-485.

Fu, W., Fang, Y., Liu, D. M. S., Wang, L., Ren, S., & Liu, Y. (2015). Surface effects on in-shoe plantar pressure and tibial impact during running. *Journal of Sport and Health Science*, *4*(4), 384-390.

Fuller, J. T., Amado, A., van Emmerik, R. E., Hamill, J., Buckley, J. D., Tsiros, M. D., & Thewlis, D. (2016). The effect of footwear and footfall pattern on running stride interval long-range correlations and distributional variability. *Gait & posture*, *44*, 137-142.

Fuller, J. T., Bellenger, C. R., Thewlis, D., Arnold, J., Thomson, R. L., Tsiros, M. D., Robertson, E. Y., & Buckley, J. D. (2017). Tracking performance changes with running-stride variability when athletes are functionally overreached. *International Journal of Sports Physiology and Performance*, *12*(3), 357-363.

Furlong, L.-A., & Egginton, N. L. (2018). Kinetic asymmetry during running at preferred and non-preferred speeds.

Gabbett, T. J. (2016). The training—injury prevention paradox: should athletes be training smarter and harder? *British journal of sports medicine*, *50*(5), 273-280.

Gallahue, D. L., & Ozmun, J. C. (2006). Understanding motor development: Infants, children, adolescents, adults. *(No Title)*.

Gallo, R. A., Plakke, M., & Silvis, M. L. (2012). Common leg injuries of long-distance runners: anatomical and biomechanical approach. *Sports Health*, *4*(6), 485-495.

Gao, Z., Mei, Q., Fekete, G., Baker, J. S., & Gu, Y. (2020). The effect of prolonged running on the symmetry of biomechanical variables of the lower limb joints. *Symmetry*, *12*(5), 720.

Gholami, M., Napier, C., & Menon, C. (2020). Estimating lower extremity running gait kinematics with a single accelerometer: A deep learning approach. *Sensors*, *20*(10), 2939.

Gidley, A. D., Lankford, D. E., & Bailey, J. P. (2020). The construction of common treadmills significantly affects biomechanical and metabolic variables. *Journal of Sports Sciences*, *38*(19), 2236-2241.

Gindre, C., Lussiana, T., Hebert-Losier, K., & Morin, J.-B. (2016). Reliability and validity of the Myotest® for measuring running stride kinematics. *Journal of Sports Sciences*, *34*(7), 664-670.

Girard, O., Millet, G. P., Slawinski, J., Racinais, S., & Micallef, J.-P. (2013). Changes in running mechanics and spring-mass behaviour during a 5-km time trial. *International journal of sports medicine*, 832-840.

Glazier, P. (2006). Measuring coordination and variability in coordination. *Movement system variability*, 167-181.

Goldberger, A. L., Amaral, L. A., Glass, L., Hausdorff, J. M., Ivanov, P. C., Mark, R. G., Mietus, J. E., Moody, G. B., Peng, C.-K., & Stanley, H. E. (2000). PhysioBank, PhysioToolkit, and PhysioNet: components of a new research resource for complex physiologic signals. *circulation*, *101*(23), e215-e220.

Goldberger, A. L., Amaral, L. A., Hausdorff, J. M., Ivanov, P. C., Peng, C.-K., & Stanley, H. E. (2002). Fractal dynamics in physiology: alterations with disease and aging. *Proceedings of the National Academy of Sciences*, *99*(suppl_1), 2466-2472.

Guignard, B., Ayad, O., Baillet, H., Mell, F., Simbana Escobar, D., Boulanger, J., & Seifert, L. (2021). Validity, reliability and accuracy of inertial measurement units (IMUs) to measure angles: application in swimming. *Sports Biomechanics*, 1-33.

Hafer, J. F., Brown, A. M., & Boyer, K. A. (2017). Exertion and pain do not alter coordination variability in runners with iliotibial band syndrome. *Clinical Biomechanics*, *47*, 73-78.

Hafer, J. F., Freedman Silvernail, J., Hillstrom, H. J., & Boyer, K. A. (2016). Changes in coordination and its variability with an increase in running cadence. *Journal of Sports Sciences*, *34*(15), 1388-1395.

Hafer, J. F., Provenzano, S. G., Kern, K. L., Agresta, C. E., Grant, J. A., & Zernicke, R. F. (2020). Measuring markers of aging and knee osteoarthritis gait using inertial measurement units. *Journal of biomechanics*, *99*, 109567.

Hamacher, D., Singh, N., Van Dieën, J. H., Heller, M., & Taylor, W. R. (2011). Kinematic measures for assessing gait stability in elderly individuals: a systematic review. *Journal of the Royal Society Interface*, *8*(65), 1682-1698.

Hamill, J., Palmer, C., & Van Emmerik, R. E. (2012). Coordinative variability and overuse injury. *Sports Medicine, Arthroscopy, Rehabilitation, Therapy & Technology*, *4*(1), 1-9.

Hamill, J., van Emmerik, R. E., Heiderscheit, B. C., & Li, L. (1999). A dynamical systems approach to lower extremity running injuries. *Clinical Biomechanics*, *14*(5), 297-308.

Hanley, B., & Tucker, C. B. (2018). Gait variability and symmetry remain consistent during high-intensity 10,000 m treadmill running. *Journal of biomechanics*, *79*, 129-134.

Harbourne, R. T., & Stergiou, N. (2009). Movement variability and the use of nonlinear tools: principles to guide physical therapist practice. *Physical therapy*, *89*(3), 267-282.

Hardin, E. C., Van Den Bogert, A. J., & Hamill, J. (2004). Kinematic adaptations during running: effects of footwear, surface, and duration. *Medicine & Science in Sports & Exercise*, *36*(5), 838-844.

Hausdorff, J., Zemany, L., Peng, C.-K., & Goldberger, A. (1999). Maturation of gait dynamics: stride-to-stride variability and its temporal organization in children. *Journal of applied physiology*, *86*(3), 1040-1047.

Hausdorff, J. M. (2005). Gait variability: methods, modeling and meaning. *Journal of neuroengineering and rehabilitation*, *2*(1), 1-9.

Hausdorff, J. M. (2007). Gait dynamics, fractals and falls: finding meaning in the stride-to-stride fluctuations of human walking. *Human movement science*, *26*(4), 555-589.

Hausdorff, J. M., Balash, J., & Giladi, N. (2003). Effects of cognitive challenge on gait variability in patients with Parkinson's disease. *Journal of geriatric psychiatry and neurology*, *16*(1), 53-58.

Hausdorff, J. M., Edelberg, H. K., Mitchell, S. L., Goldberger, A. L., & Wei, J. Y. (1997). Increased gait unsteadiness in community-dwelling elderly fallers. *Archives of physical medicine and rehabilitation*, *78*(3), 278-283.

Hausdorff, J. M., Mitchell, S. L., Firtion, R., Peng, C.-K., Cudkowicz, M. E., Wei, J. Y., & Goldberger, A. L. (1997). Altered fractal dynamics of gait: reduced stride-interval correlations with aging and Huntington's disease. *Journal of applied physiology*, *82*(1), 262-269.

Hausdorff, J. M., Nelson, M. E., Kaliton, D., Layne, J. E., Bernstein, M. J., Nuernberger, A., & Singh, M. A. F. (2001). Etiology and modification of gait instability in older adults: a randomized controlled trial of exercise. *Journal of applied physiology*, *90*(6), 2117-2129.

Hausdorff, J. M., Rios, D. A., & Edelberg, H. K. (2001). Gait variability and fall risk in community-living older adults: a 1-year prospective study. *Archives of physical medicine and rehabilitation*, *82*(8), 1050-1056.

Hawkins, T. (2019). *The Effects of Dual-Tasking on Gait Dynamics in Older Adults with Cognitive Impairment*

Hawkins, T. C., Samuel, R., Fiatarone Singh, M. A., Gates, N., Wilson, G. C., Jain, N., Meiklejohn, J., Brodaty, H., Wen, W., & Singh, N. (2019). Impairment of dual-task gait dynamics in older adults with mild cognitive impairment: Relationships to neuropsychological status, fitness and brain morphology. *medRxiv*, 19005249.

Hawley, J. A., & Noakes, T. D. (1992). Peak power output predicts maximal oxygen uptake and performance time in trained cyclists. *European journal of applied physiology and occupational physiology*, *65*, 79-83.

He, M., Qi, Z., Shao, Y., Yao, H., Zhang, X., Zhang, Y., Shi, Y., E, Q., Liu, C., & Hu, H. (2021). Quantitative evaluation of gait changes using APDM inertial sensors after the external lumbar drain in patients with idiopathic normal pressure hydrocephalus. *Frontiers in Neurology*, *12*, 635044.

He, Y., Chen, Y., Tang, L., Chen, J., Tang, J., Yang, X., Su, S., Zhao, C., & Xiao, N. (2024). Accuracy validation of a wearable IMU-based gait analysis in healthy female. *BMC Sports Science, Medicine and Rehabilitation*, *16*(1), 2.

Heck, H., Mader, A., Hess, G., Mücke, S., Müller, R., & Hollmann, W. (1985). Justification of the 4-mmol/l lactate threshold. *International journal of sports medicine*, *6*(03), 117-130.

Heesch, M. W., & Slivka, D. R. (2015). Running performance, pace strategy, and thermoregulation differ between a treadmill and indoor track. *The Journal of Strength & Conditioning Research*, *29*(2), 330-335.

Heiderscheit, B. C. (2000). Movement variability as a clinical measure for locomotion. *Journal of Applied Biomechanics*, *16*(4), 419-427.

Heiderscheit, B. C., Hamill, J., & Van Emmerik, R. (1999). Q-angle influences on the variability of lower extremity coordination during running. *Medicine and science in sports and exercise*, *31*(9), 1313-1319.

Heiderscheit, B. C., Hamill, J., & van Emmerik, R. E. (2002). Variability of stride characteristics and joint coordination among individuals with unilateral patellofemoral pain. *Journal of Applied Biomechanics*, *18*(2), 110-121.

Hein, T., Schmeltzpfenning, T., Krauss, I., Maiwald, C., Horstmann, T., & Grau, S. (2012). Using the variability of continuous relative phase as a measure to discriminate between healthy and injured runners. *Human movement science*, *31*(3), 683-694.

Helbostad, J. L., & Moe-Nilssen, R. (2003). The effect of gait speed on lateral balance control during walking in healthy elderly. *Gait & posture*, *18*(2), 27-36.

Helgerud, J., Høydal, K., Wang, E., Karlsen, T., Berg, P., Bjerkaas, M., Simonsen, T., Helgesen, C., Hjorth, N., & Bach, R. (2007). Aerobic high-intensity intervals improve V˙ O2max more than moderate training. *Medicine & Science in Sports & Exercise*, *39*(4), 665-671.

Herman, T., Giladi, N., Gurevich, T., & Hausdorff, J. (2005). Gait instability and fractal dynamics of older adults with a "cautious" gait: why do certain older adults walk fearfully? *Gait & posture*, *21*(2), 178-185.

Hollman, J. H., Kovash, F. M., Kubik, J. J., & Linbo, R. A. (2007). Age-related differences in spatiotemporal markers of gait stability during dual task walking. *Gait & posture*, *26*(1), 113-119.

Hollman, J. H., McDade, E. M., & Petersen, R. C. (2011). Normative spatiotemporal gait parameters in older adults. *Gait & posture*, *34*(1), 111-118.

Hollman, J. H., Watkins, M. K., Imhoff, A. C., Braun, C. E., Akervik, K. A., & Ness, D. K. (2016). A comparison of variability in spatiotemporal gait parameters between treadmill and overground walking conditions. *Gait & posture*, *43*, 204-209.

Horenstein, R. E., Goudeau, Y. R., Lewis, C. L., & Shefelbine, S. J. (2020). Using Magneto-Inertial Measurement Units to Pervasively Measure Hip Joint Motion during Sports. *Sensors*, *20*(17), 4970.

Hu, M., Kobayashi, T., Zhou, J., & Lam, W.-K. (2021). Current application of continuous relative phase in running and jumping studies: a systematic review. *Gait & posture*, *90*, 215-233.

Huang, W.-N. W., VanSwearingen, J. M., & Brach, J. S. (2008). Gait variability in older adults: observational rating validated by comparison with a computerized walkway gold standard. *Physical therapy*, *88*(10), 1146-1153.

Hulme, A., Nielsen, R. O., Timpka, T., Verhagen, E., & Finch, C. (2017). Risk and protective factors for middle-and long-distance running-related injury. *Sports medicine*, *47*, 869-886.

Iosa, M., Marro, T., Paolucci, S., & Morelli, D. (2012). Stability and harmony of gait in children with cerebral palsy. *Research in Developmental Disabilities*, *33*(1), 129-135.

Jakob, V., Küderle, A., Kluge, F., Klucken, J., Eskofier, B. M., Winkler, J., Winterholler, M., & Gassner, H. (2021). Validation of a sensor-based gait analysis system with a gold-standard motion capture system in patients with Parkinson's disease. *Sensors*, *21*(22), 7680.

Jones, A. M., & Doust, J. H. (1996). A 1% treadmill grade most accurately reflects the energetic cost of outdoor running. *Journal of Sports Sciences*, *14*(4), 321-327.

Jordan, K., Challis, J. H., & Newell, K. M. (2006). Long range correlations in the stride interval of running. *Gait & posture*, *24*(1), 120-125.

Jordan, K., Challis, J. H., & Newell, K. M. (2007a). Speed influences on the scaling behavior of gait cycle fluctuations during treadmill running. *Human movement science*, *26*(1), 87-102.

Jordan, K., Challis, J. H., & Newell, K. M. (2007b). Walking speed influences on gait cycle variability. *Gait & posture*, *26*(1), 128-134.

Junior, L. C. H., Costa, L. O. P., & Lopes, A. D. (2013). Previous injuries and some training characteristics predict running-related injuries in recreational runners: a prospective cohort study. *Journal of Physiotherapy*, *59*(4), 263-269.

Kahlon, A. S., Verma, K., Sage, A., Lee, S. C., & Behboodi, A. (2023). Enhancing wearable gait monitoring systems: identifying optimal kinematic inputs in typical adolescents. *Sensors*, *23*(19), 8275.

Kang, H. G., & Dingwell, J. B. (2008). Separating the effects of age and walking speed on gait variability. *Gait & posture*, *27*(4), 572-577.

Kelso, J., & Ding, M. (1993). Fluctuations, intermittency, and controllable chaos in biological coordination. *Variability and motor control*, *1993*, 291-316.

Kelso, J. S. (1995). *Dynamic patterns: The self-organization of brain and behavior*. MIT press.

Kerdok, A. E., Biewener, A. A., McMahon, T. A., Weyand, P. G., & Herr, H. M. (2002). Energetics and mechanics of human running on surfaces of different stiffnesses. *Journal of applied physiology*.

Klöpfer-Krämer, I., Brand, A., Wackerle, H., Müßig, J., Kröger, I., & Augat, P. (2020). Gait analysis–Available platforms for outcome assessment. *Injury*, *51*, S90-S96.

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

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

Ko, C.-Y., Chang, Y., Jeong, B., Kang, S., Ryu, J., & Kim, G. (2017). Effects of knee sleeves on coordination of lower-limb segments in healthy adults during level walking and one-leg hopping. *PeerJ*, *5*, e3340.

Kobsar, D., Charlton, J. M., Tse, C. T., Esculier, J.-F., Graffos, A., Krowchuk, N. M., Thatcher, D., & Hunt, M. A. (2020). Validity and reliability of wearable inertial sensors in healthy adult walking: A systematic review and meta-analysis. *Journal of neuroengineering and rehabilitation*, *17*, 1-21.

Kurz, M. J., Stergiou, N., Buzzi, U. H., & Georgoulis, A. D. (2005). The effect of anterior cruciate ligament recontruction on lower extremity relative phase dynamics during walking and running. *Knee Surgery, Sports Traumatology, Arthroscopy*, *13*, 107-115.

Lanovaz, J. L., Oates, A. R., Treen, T. T., Unger, J., & Musselman, K. E. (2017). Validation of a commercial inertial sensor system for spatiotemporal gait measurements in children. *Gait & posture*, *51*, 14-19.

Laursen, P. B., & Jenkins, D. G. (2002). The scientific basis for high-intensity interval training. *Sports medicine*, *32*(1), 53-73.

Lee, D.-c., Pate, R. R., Lavie, C. J., Sui, X., Church, T. S., & Blair, S. N. (2014). Leisure-time running reduces all-cause and cardiovascular mortality risk. *Journal of the American College of Cardiology*, *64*(5), 472-481.

Lieberman, D. E., Venkadesan, M., Werbel, W. A., Daoud, A. I., D'andrea, S., Davis, I. S., Mang'Eni, R. O., & Pitsiladis, Y. (2010). Foot strike patterns and collision forces in habitually barefoot versus shod runners. *Nature*, *463*(7280), 531-535.

Lilley, T., Herb, C. C., Hart, J., & Hertel, J. (2018). Lower extremity joint coupling variability during gait in young adults with and without chronic ankle instability. *Sports Biomechanics*, *17*(2), 261-272.

Lindsay, T. R., Noakes, T. D., & McGregor, S. J. (2014). Effect of treadmill versus overground running on the structure of variability of stride timing. *Perceptual and motor skills*, *118*(2), 331-346.

Lipsitz, L. A. (2002). Dynamics of stability: the physiologic basis of functional health and frailty. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, *57*(3), B115-B125.

Lipsitz, L. A., & Goldberger, A. L. (1992). Loss of'complexity'and aging: potential applications of fractals and chaos theory to senescence. *Jama*, *267*(13), 1806-1809.

Lopes, A. D., Costa, L. O. P., Saragiotto, B. T., Yamato, T. P., Adami, F., & Verhagen, E. (2011). Musculoskeletal pain is prevalent among recreational runners who are about to compete: an observational study of 1049 runners. *Journal of Physiotherapy*, *57*(3), 179-182.

Lopes, A. D., Hespanhol, L. C., Yeung, S. S., & Costa, L. O. P. (2012). What are the main running-related musculoskeletal injuries? A systematic review. *Sports medicine*, *42*, 891-905.

Lord, S., Howe, T., Greenland, J., Simpson, L., & Rochester, L. (2011). Gait variability in older adults: a structured review of testing protocol and clinimetric properties. *Gait & posture*, *34*(4), 443-450.

Lucas-Cuevas, Á. G., Quesada, J. I. P., Gooding, J., Lewis, M. G., Encarnación-Martínez, A., & Perez-Soriano, P. (2018). The effect of visual focus on spatio-temporal and kinematic parameters of treadmill running. *Gait & posture*, *59*, 292-297.

Lukšys, D., Jatužis, D., Jonaitis, G., & Griškevičius, J. (2021). Application of continuous relative phase analysis for differentiation of gait in neurodegenerative disease. *Biomedical signal processing and control*, *67*, 102558.

Luo, Z., Zhang, X., Wang, J., Yang, Y., Xu, Y., & Fu, W. (2019). Changes in ground reaction forces, joint mechanics, and stiffness during treadmill running to fatigue. *Applied Sciences*, *9*(24), 5493.

Maas, E., De Bie, J., Vanfleteren, R., Hoogkamer, W., & Vanwanseele, B. (2018). Novice runners show greater changes in kinematics with fatigue compared with competitive runners. *Sports Biomechanics*, *17*(3), 350-360.

Maki, B. E. (1997). Gait changes in older adults: predictors of falls or indicators of fear? *Journal of the American geriatrics society*, *45*(3), 313-320.

Maksud, M. G., Coutts, K. D., & Hamilton, L. H. (1971). Time course of heart rate, ventilation, and Vo2 during laboratory and field exercise. *Journal of applied physiology*, *30*(4), 536-539.

Malatesta, D., Simar, D., Dauvilliers, Y., Candau, R., Borrani, F., Préfaut, C., & Caillaud, C. (2003). Energy cost of walking and gait instability in healthy 65-and 80-yr-olds. *Journal of applied physiology*, *95*(6), 2248-2256.

Malina, R. M., Bouchard, C., & Bar-Or, O. (2004). *Growth, maturation, and physical activity*. Human kinetics.

Mancini, M., & Horak, F. B. (2016). Potential of APDM mobility lab for the monitoring of the progression of Parkinson's disease. *Expert review of medical devices*, *13*(5), 455-462.

Mangone, M., Marinelli, E., Santilli, G., Finanore, N., Agostini, F., Santilli, V., Bernetti, A., Paoloni, M., & Zaami, S. (2023). Gait analysis advancements: rehabilitation value and new perspectives from forensic application. *European Review for Medical and Pharmacological Sciences*, *27*(1), 3-12.

Mann, R., Malisoux, L., Brunner, R., Gette, P., Urhausen, A., Statham, A., Meijer, K., & Theisen, D. (2014). Reliability and validity of pressure and temporal parameters recorded using a pressure-sensitive insole during running. *Gait & posture*, *39*(1), 455-459.

Mann, R., Malisoux, L., Nührenbörger, C., Urhausen, A., Meijer, K., & Theisen, D. (2015). Association of previous injury and speed with running style and stride-to-stride fluctuations. *Scandinavian journal of medicine & science in sports*, *25*(6), e638-e645.

Mann, R., Malisoux, L., Urhausen, A., Statham, A., Meijer, K., & Theisen, D. (2015). The effect of shoe type and fatigue on strike index and spatiotemporal parameters of running. *Gait & posture*, *42*(1), 91-95.

Manupibul, U., Tanthuwapathom, R., Jarumethitanont, W., Kaimuk, P., Limroongreungrat, W., & Charoensuk, W. (2023). Integration of force and IMU sensors for developing low-cost portable gait measurement system in lower extremities. *Scientific Reports*, *13*(1), 10653.

Martin, J.-P., & Li, Q. (2017). Overground vs. treadmill walking on biomechanical energy harvesting: An energetics and EMG study. *Gait & posture*, *52*, 124-128.

Maruyama, H., & Nagasaki, H. (1992). Temporal variability in the phase durations during treadmill walking. *Human movement science*, *11*(3), 335-348.

Mason, R., Pearson, L. T., Barry, G., Young, F., Lennon, O., Godfrey, A., & Stuart, S. (2023). Wearables for running gait analysis: A systematic review. *Sports medicine*, *53*(1), 241-268.

Mc Ardle, R., Del Din, S., Galna, B., Thomas, A., & Rochester, L. (2020). Differentiating dementia disease subtypes with gait analysis: feasibility of wearable sensors? *Gait & posture*, *76*, 372-376.

McCrory, J. L., Moon, G. R., Leary, B. K., & Nguyen, A.-D. (2022). A Comparison Of Treadmill And Overground Running With The Use Of IMUs: 1886. *Medicine & Science in Sports & Exercise*, *54*(9S), 560.

Meardon, S. A., Hamill, J., & Derrick, T. R. (2011). Running injury and stride time variability over a prolonged run. *Gait & posture*, *33*(1), 36-40.

Mehdizadeh, S. (2018). The largest Lyapunov exponent of gait in young and elderly individuals: A systematic review. *Gait & posture*, *60*, 241-250.

Mehdizadeh, S., Arshi, A. R., & Davids, K. (2015). Quantifying coordination and coordination variability in backward versus forward running: Implications for control of motion. *Gait & posture*, *42*(2), 172-177.

Menz, H. B., Lord, S. R., & Fitzpatrick, R. C. (2003). Acceleration patterns of the head and pelvis when walking are associated with risk of falling in community-dwelling older people. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, *58*(5), M446-M452.

Mileti, I., Taborri, J., Rossi, S., Petrarca, M., Patanè, F., & Cappa, P. (2016). Evaluation of the effects on stride-to-stride variability and gait asymmetry in children with Cerebral Palsy wearing the WAKE-up ankle module. 2016 IEEE International Symposium on Medical Measurements and Applications (MeMeA),

Milgrom, C., Radeva-Petrova, D. R., Finestone, A., Nyska, M., Mendelson, S., Benjuya, N., Simkin, A., & Burr, D. (2007). The effect of muscle fatigue on in vivo tibial strains. *Journal of biomechanics*, *40*(4), 845-850.

Miller, J. R., Van Hooren, B., Bishop, C., Buckley, J. D., Willy, R. W., & Fuller, J. T. (2019). A systematic review and meta-analysis of crossover studies comparing physiological, perceptual and performance measures between treadmill and overground running. *Sports medicine*, *49*, 763-782.

Miller, R. H., Chang, R., Baird, J. L., Van Emmerik, R. E., & Hamill, J. (2010). Variability in kinematic coupling assessed by vector coding and continuous relative phase. *Journal of biomechanics*, *43*(13), 2554-2560.

Miller, R. H., Edwards, W. B., Brandon, S., Morton, A. M., & Deluzio, K. J. (2014). Why don't most runners get knee osteoarthritis? A case for per-unit-distance loads. *Medicine and science in sports and exercise*, *46*(3), 572-579.

Miller, R. H., Meardon, S. A., Derrick, T. R., & Gillette, J. C. (2008). Continuous relative phase variability during an exhaustive run in runners with a history of iliotibial band syndrome. *Journal of Applied Biomechanics*, *24*(3), 262-270.

Millet, G. Y., & Lepers, R. (2004). Alterations of neuromuscular function after prolonged running, cycling and skiing exercises. *Sports medicine*, *34*, 105-116.

Mitchell, C., McDonnell, S., Oganezova, K., Mockler, D., & Fleming, N. (2023). The effect of surface compliance on overground running biomechanics. A systematic review and meta-analysis. *Sports Biomechanics*, 1-25.

Mo, S. (2018). Evaluation of stride variability between experienced and novice runners during a prolonged run.

Mo, S., & Chow, D. H. (2018a). Accuracy of three methods in gait event detection during overground running. *Gait & posture*, *59*, 93-98.

Mo, S., & Chow, D. H. (2018b). Stride-to-stride variability and complexity between novice and experienced runners during a prolonged run at anaerobic threshold speed. *Gait & posture*, *64*, 7-11.

Mo, S., & Chow, D. H. K. (2019). Differences in lower-limb coordination and coordination variability between novice and experienced runners during a prolonged treadmill run at anaerobic threshold speed. *Journal of Sports Sciences*, *37*(9), 1021-1028.

Moe-Nilssen, R., & Helbostad, J. L. (2005). Interstride trunk acceleration variability but not step width variability can differentiate between fit and frail older adults. *Gait & posture*, *21*(2), 164-170.

Möhler, F., Fadillioglu, C., Scheffler, L., Müller, H., & Stein, T. (2022). Running-Induced Fatigue Changes the Structure of Motor Variability in Novice Runners. *Biology*, *11*(6), 942.

Mok, K.-M., Lee, J., Chung, M., & Hong, Y. (2009). A kinematic comparison of running on treadmill and overground surfaces. ISBS-Conference Proceedings Archive,

Morgan, D. W., Martin, P. E., & Krahenbuhl, G. S. (1989). Factors affecting running economy. *Sports Med*, *7*(5), 310-330.

Morin, J.-B., & Sève, P. (2011). Sprint running performance: comparison between treadmill and field conditions. *European journal of applied physiology*, *111*, 1695-1703.

Mullineaux, D. R., Clayton, H. M., & Gnagey, L. M. (2004). Effects of offset-normalizing techniques on variability in motion analysis data. *Journal of Applied Biomechanics*, *20*(2), 177-184.

Murray, M., Spurr, G., Sepic, S., Gardner, G., & Mollinger, L. (1985). Treadmill vs. floor walking: kinematics, electromyogram, and heart rate. *Journal of applied physiology*, *59*(1), 87-91.

Mutchler, J., Macias, K., Munkasy, B., Wilson, S. J., Garner III, J. C., & Li, L. (2020). Kinematic and coordination variability in runners with and without patellofemoral pain. *International Journal of Kinesiology & Sports Science*, *8*(3), 58.

Myburgh, K. H., Hutchins, J., Fataar, A. B., Hough, S. F., & Noakes, T. D. (1990). Low bone density is an etiologic factor for stress fractures in athletes. *Annals of internal medicine*, *113*(10), 754-759.

Nakayama, Y., Kudo, K., & Ohtsuki, T. (2010). Variability and fluctuation in running gait cycle of trained runners and non-runners. *Gait & posture*, *31*(3), 331-335.

Newell, K. M., & Vaillancourt, D. E. (2001). Dimensional change in motor learning. *Human movement science*, *20*(4-5), 695-715.

Nigg, B. M., De Boer, R. W., & Fisher, V. (1995). A kinematic comparison of overground and treadmill running. *Medicine and science in sports and exercise*, *27*, 98-98.

Norris, M., Anderson, R., & Kenny, I. C. (2014). Previous and potential use of inertial senors in longitudinal running gait analysis links with running Ireland injury.

Owings, T. M., & Grabiner, M. D. (2003). Measuring step kinematic variability on an instrumented treadmill: how many steps are enough? *Journal of biomechanics*, *36*(8), 1215-1218.

Owings, T. M., & Grabiner, M. D. (2004a). Step width variability, but not step length variability or step time variability, discriminates gait of healthy young and older adults during treadmill locomotion. *Journal of biomechanics*, *37*(6), 935-938.

Owings, T. M., & Grabiner, M. D. (2004b). Variability of step kinematics in young and older adults. *Gait & posture*, *20*(1), 26-29.

Panday, S. B., Pathak, P., Moon, J., & Koo, D. (2022). Complexity of Running and Its Relationship with Joint Kinematics during a Prolonged Run. *International Journal of Environmental Research and Public Health*, *19*(15), 9656.

Paquette, M., Milner, C., & Melcher, D. (2017). Foot contact angle variability during a prolonged run with relation to injury history and habitual foot strike pattern. *Scandinavian journal of medicine & science in sports*, *27*(2), 217-222.

Park, S., & Yoon, S. (2021). Validity evaluation of an inertial measurement unit (IMU) in gait analysis using statistical parametric mapping (SPM). *Sensors*, *21*(11), 3667.

Patra, R., Das, H. C., & Jena, S. (2022). Evaluation of the mechanical behaviour of a bipolar hip prosthesis under transient loading. *International Journal of Biomedical Engineering and Technology*, *39*(3), 314-326.

Pescatello, L. S. (2014). *ACSM's guidelines for exercise testing and prescription*. Lippincott Williams & Wilkins.

Peserico, C. S., & Machado, F. A. (2014). Comparison between running performance in time trials on track and treadmill. *Revista Brasileira de Cineantropometria & Desempenho Humano*, *16*, 456-464.

Pohl, M. B., Lloyd, C., & Ferber, R. (2010). Can the reliability of three-dimensional running kinematics be improved using functional joint methodology? *Gait & posture*, *32*(4), 559-563.

Politi, A. (2013). Lyapunov exponent. *Scholarpedia*, *8*(3), 2722.

Ponzio, D. Y., Syed, U. A. M., Purcell, K., Cooper, A. M., Maltenfort, M., Shaner, J., & Chen, A. F. (2018). Low prevalence of hip and knee arthritis in active marathon runners. *JBJS*, *100*(2), 131-137.

Prasanth, H., Caban, M., Keller, U., Courtine, G., Ijspeert, A., Vallery, H., & Von Zitzewitz, J. (2021). Wearable sensor-based real-time gait detection: A systematic review. *Sensors*, *21*(8), 2727.

Preatoni, E., Hamill, J., Harrison, A. J., Hayes, K., Van Emmerik, R. E., Wilson, C., & Rodano, R. (2013). Movement variability and skills monitoring in sports. *Sports Biomechanics*, *12*(2), 69-92.

Prejean, B. J., & Ricard, M. D. (2019). A quantification of lower-limb coordinative variability during running with different levels of midsole cushioning. *Footwear Science*, *11*(2), 93-104.

Prosser, L. A., Lauer, R. T., VanSant, A. F., Barbe, M. F., & Lee, S. C. (2010). Variability and symmetry of gait in early walkers with and without bilateral cerebral palsy. *Gait & posture*, *31*(4), 522-526.

Pugh, L. G. C. E. (1970). Oxygen intake in track and treadmill running with observations on the effect of air resistance. *The Journal of physiology*, *207*(3), 823-835.

Rebula, J. R., Ojeda, L. V., Adamczyk, P. G., & Kuo, A. D. (2013). Measurement of foot placement and its variability with inertial sensors. *Gait & posture*, *38*(4), 974-980.

Reisman, D. S., Wityk, R., Silver, K., & Bastian, A. J. (2007). Locomotor adaptation on a split-belt treadmill can improve walking symmetry post-stroke. *Brain, 130*(7), 1861-1872.

Riccio, G. E. (1993). Information in movement variability about qualitative dynamics of posture and orientation. *Variability and motor control*.

Richman, J. S., & Moorman, J. R. (2000). Physiological time-series analysis using approximate entropy and sample entropy. *American journal of physiology-heart and circulatory physiology*, *278*(6), H2039-H2049.

Riley, P. O., Dicharry, J., Franz, J., Croce, U. D., Wilder, R. P., & Kerrigan, D. C. (2008). A kinematics and kinetic comparison of overground and treadmill running. *Medicine and science in sports and exercise*, *40*(6), 1093.

Riley, P. O., Paolini, G., Della Croce, U., Paylo, K. W., & Kerrigan, D. C. (2007). A kinematic and kinetic comparison of overground and treadmill walking in healthy subjects. *Gait & posture*, *26*(1), 17-24.

Riva, F., Bisi, M. C., & Stagni, R. (2014). Gait variability and stability measures: Minimum number of strides and within-session reliability. *Computers in biology and medicine*, *50*, 9-13.

Robert-Lachaine, X., Mecheri, H., Larue, C., & Plamondon, A. (2017). Validation of inertial measurement units with an optoelectronic system for whole-body motion analysis. *Medical & biological engineering & computing*, *55*, 609-619.

Robert-Lachaine, X., Mecheri, H., Muller, A., Larue, C., & Plamondon, A. (2020). Validation of a low-cost inertial motion capture system for whole-body motion analysis. *Journal of biomechanics*, *99*, 109520.

Robertson, D. G. E., Caldwell, G. E., Hamill, J., Kamen, G., & Whittlesey, S. (2013). *Research methods in biomechanics*. Human kinetics.

Rodrigues, P., Chang, R., TenBroek, T., van Emmerik, R., & Hamill, J. (2015). Evaluating the coupling between foot pronation and tibial internal rotation continuously using vector coding. *Journal of Applied Biomechanics*, *31*(2), 88-94.

Rossi, S., Colazza, A., Petrarca, M., Castelli, E., Cappa, P., & Krebs, H. I. (2013). Feasibility study of a wearable exoskeleton for children: is the gait altered by adding masses on lower limbs? *PloS one*, *8*(9), e73139.

Rubenstein, L. Z., & Josephson, K. R. (2002). The epidemiology of falls and syncope. *Clinics in geriatric medicine*, *18*(2), 141-158.

Running, U. (2017). National Runner Survey.

Salarian, A., Russmann, H., Vingerhoets, F. J., Dehollain, C., Blanc, Y., Burkhard, P. R., & Aminian, K. (2004). Gait assessment in Parkinson's disease: toward an ambulatory system for long-term monitoring. *IEEE Transactions on Biomedical Engineering*, *51*(8), 1434-1443.

Satti, R., Abid, N.-U.-H., Bottaro, M., De Rui, M., Garrido, M., Raoufy, M. R., Montagnese, S., & Mani, A. R. (2019). The application of the extended Poincaré plot in the analysis of physiological variabilities. *Frontiers in physiology*, *10*, 116.

Schaafsma, J. D., Giladi, N., Balash, Y., Bartels, A. L., Gurevich, T., & Hausdorff, J. M. (2003). Gait dynamics in Parkinson's disease: relationship to Parkinsonian features, falls and response to levodopa. *Journal of the neurological sciences*, *212*(1-2), 47-53.

Schache, A. G., Blanch, P. D., Rath, D. A., Wrigley, T. V., Starr, R., & Bennell, K. L. (2001). A comparison of overground and treadmill running for measuring the three-dimensional kinematics of the lumbo–pelvic–hip complex. *Clinical Biomechanics*, *16*(8), 667-680.

Scheerder, J., Breedveld, K., & Borgers, J. (2015). Who is doing a run with the running boom? The growth and governance of one of Europe's most popular sport activities. In *Running across Europe: The rise and size of one of the largest sport markets* (pp. 1-27). Springer.

Schieb, D. A. (1986). Kinematic accommodation of novice treadmill runners. *Research quarterly for exercise and sport*, *57*(1), 1-7.

Schmitz, A., Russo, K., Edwards, L., & Noehren, B. (2014). Do novice runners have weak hips and bad running form? *Gait & posture*, *40*(1), 82-86.

Schubert, A. G., Kempf, J., & Heiderscheit, B. C. (2014). Influence of stride frequency and length on running mechanics: a systematic review. *Sports Health*, *6*(3), 210-217.

Schücker, L., & Parrington, L. (2019). Thinking about your running movement makes you less efficient: attentional focus effects on running economy and kinematics. *Journal of Sports Sciences*, *37*(6), 638-646.

Schütte, K. H., Aeles, J., De Beéck, T. O., van der Zwaard, B. C., Venter, R., & Vanwanseele, B. (2016). Surface effects on dynamic stability and loading during outdoor running using wireless trunk accelerometry. *Gait & posture*, *48*, 220-225.

Schwartz, M. H., Trost, J. P., & Wervey, R. A. (2004). Measurement and management of errors in quantitative gait data. *Gait & posture*, *20*(2), 196-203.

Seay, J. F., Van Emmerik, R. E., & Hamill, J. (2011). Low back pain status affects pelvis-trunk coordination and variability during walking and running. *Clinical Biomechanics*, *26*(6), 572-578.

Seay, J. F., Van Emmerik, R. E., & Hamill, J. (2014). Trunk bend and twist coordination is affected by low back pain status during running. *European journal of sport science*, *14*(6), 563-568.

Seifert, L., Button, C., & Davids, K. (2013). Key properties of expert movement systems in sport: an ecological dynamics perspective. *Sports medicine*, *43*, 167-178.

Seifert, L., Leblanc, H., Chollet, D., & Delignières, D. (2010). Inter-limb coordination in swimming: Effect of speed and skill level. *Human movement science*, *29*(1), 103-113.

Sekiya, N., Nagasaki, H., Ito, H., & Furuna, T. (1997). Optimal walking in terms of variability in step length. *journal of orthopaedic & sports physical therapy*, *26*(5), 266-272.

Selinger, J. C., Hicks, J. L., Jackson, R. W., Wall-Scheffler, C. M., Chang, D., & Delp, S. L. (2022). Running in the wild: Energetics explain ecological running speeds. *Current Biology*, *32*(10), 2309-2315. e2303.

Senanayake, D., Halgamuge, S., & Ackland, D. C. (2021). Real-time conversion of inertial measurement unit data to ankle joint angles using deep neural networks. *Journal of biomechanics*, *125*, 110552.

Sethi, D., Bharti, S., & Prakash, C. (2022). A comprehensive survey on gait analysis: History, parameters, approaches, pose estimation, and future work. *Artificial Intelligence in Medicine*, *129*, 102314.

Sheridan, P. L., Solomont, J., Kowall, N., & Hausdorff, J. M. (2003). Influence of executive function on locomotor function: divided attention increases gait variability in Alzheimer's disease. *Journal of the American Geriatrics Society*, *51*(11), 1633-1637.

Silvernail, J. F., Boyer, K., Rohr, E., Brüggemann, G.-P., & Hamill, J. (2015). Running Mechanics and Variability with Aging. *Medicine and science in sports and exercise*, *47*(10), 2175-2180.

Sinclair, J., Richards, J., Taylor, P. J., Edmundson, C. J., Brooks, D., & Hobbs, S. J. (2013). Three-dimensional kinematic comparison of treadmill and overground running. *Sports Biomechanics*, *12*(3), 272-282.

Sloot, L., Van der Krogt, M., & Harlaar, J. (2014). Effects of adding a virtual reality environment to different modes of treadmill walking. *Gait & posture, 39*(3), 939-945.

Sousa, A. S., Silva, A., & Tavares, J. M. R. (2012). Biomechanical and neurophysiological mechanisms related to postural control and efficiency of movement: a review. *Somatosensory & motor research*, *29*(4), 131-143.

Sousa, A. S., Silva, A., & Tavares, J. M. R. (2013). Interlimb relation during the double support phase of gait: an electromyographic, mechanical and energy-based analysis. *Proceedings of the Institution of Mechanical Engineers, Part H: Journal of Engineering in Medicine*, *227*(3), 327-333.

Sousa, A. S., & Tavares, J. M. R. (2012). Effect of gait speed on muscle activity patterns and magnitude during stance. *Motor Control*, *16*(4), 480-492.

Srinivasan, D., Samani, A., Mathiassen, S. E., & Madeleine, P. (2015). The size and structure of arm movement variability decreased with work pace in a standardised repetitive precision task. *Ergonomics*, *58*(1), 128-139.

Stergiou, N. (2004). *Innovative analyses of human movement: Analytical tools for human movement research*. Human Kinetics Champaign.

Stergiou, N., & Decker, L. M. (2011). Human movement variability, nonlinear dynamics, and pathology: is there a connection? *Human movement science*, *30*(5), 869-888.

Stöggl, T. L., & Sperlich, B. (2015). The training intensity distribution among well-trained and elite endurance athletes. *Frontiers in physiology*, *6*, 295.

Strohrmann, C., Harms, H., Kappeler-Setz, C., & Troster, G. (2012). Monitoring kinematic changes with fatigue in running using body-worn sensors. *IEEE transactions on information technology in biomedicine*, *16*(5), 983-990.

Strongman, C., & Morrison, A. (2020). A scoping review of non-linear analysis approaches measuring variability in gait due to lower body injury or dysfunction. *Human movement science*, *69*, 102562.

Taborri, J., Rossi, S., Palermo, E., Patanè, F., & Cappa, P. (2014). A novel HMM distributed classifier for the detection of gait phases by means of a wearable inertial sensor network. *Sensors*, *14*(9), 16212-16234.

Tamburini, P., Storm, F., Buckley, C., Bisi, M. C., Stagni, R., & Mazzà, C. (2018). Moving from laboratory to real life conditions: Influence on the assessment of variability and stability of gait. *Gait & posture*, *59*, 248-252.

Tao, W., Liu, T., Zheng, R., & Feng, H. (2012). Gait analysis using wearable sensors. *Sensors*, *12*(2), 2255-2283.

Taunton, J., Ryan, M., Clement, D., McKenzie, D., Lloyd-Smith, D., & Zumbo, B. (2003). A prospective study of running injuries: the Vancouver Sun Run "In Training" clinics. *British journal of sports medicine*, *37*(3), 239-244.

Terrier, P., & Dériaz, O. (2011). Kinematic variability, fractal dynamics and local dynamic stability of treadmill walking. *Journal of neuroengineering and rehabilitation*, *8*, 1-14.

Terrier, P., & Schutz, Y. (2003). Variability of gait patterns during unconstrained walking assessed by satellite positioning (GPS). *European journal of applied physiology*, *90*(5), 554-561.

Terrier, P., Turner, V., & Schutz, Y. (2005). GPS analysis of human locomotion: further evidence for long-range correlations in stride-to-stride fluctuations of gait parameters. *Human movement science*, *24*(1), 97-115.

Teufl, W., Miezal, M., Taetz, B., Fröhlich, M., & Bleser, G. (2018). Validity, test-retest reliability and long-term stability of magnetometer free inertial sensor based 3D joint kinematics. *Sensors*, *18*(7), 1980.

Teufl, W., Taetz, B., Miezal, M., Lorenz, M., Pietschmann, J., Jöllenbeck, T., Fröhlich, M., & Bleser, G. (2019). Towards an inertial sensor-based wearable feedback system for patients after total hip arthroplasty: Validity and applicability for gait classification with gait kinematics-based features. *Sensors*, *19*(22), 5006.

Tonoli, D. C., Cumps, E., Aerts, I., Verhagen, E., & Meeusen, R. (2010). Incidence, risk factors and prevention of running related injuries in long-distance running: a systematic review. *Sport & Geneeskunde*, *43*(5).

Toro, I. S., Weir, G., Amado, A., Van Emmerik, R., Ervilha, U., & Hamill, J. (2022). Is coordination variability using vector coding different in overground and treadmill walking and running? *Gait & posture*, *92*, 413-420.

Tschopp, M., & Brunner, F. (2017). Erkrankungen und Überlastungsschäden an der unteren Extremität bei Langstreckenläufern. *Zeitschrift für Rheumatologie*, *76*(5).

van den Bogert, A. J., Read, L., & Nigg, B. M. (1999). An analysis of hip joint loading during walking, running, and skiing. *Medicine and science in sports and exercise*, *31*(1), 131-142.

Van der Worp, H., Vrielink, J. W., & Bredeweg, S. W. (2016). Do runners who suffer injuries have higher vertical ground reaction forces than those who remain injury-free? A systematic review and meta-analysis. *British journal of sports medicine*, *50*(8), 450-457.

Van Emmerik, R. E., Rosenstein, M. T., McDermott, W. J., & Hamill, J. (2004). A nonlinear dynamics approach to human movement. *Journal of Applied Biomechanics*, *20*(4), 396-420.

Van Gent, R., Siem, D., van Middelkoop, M., Van Os, A., Bierma-Zeinstra, S., & Koes, B. (2007). Incidence and determinants of lower extremity running injuries in long distance runners: a systematic review. *British journal of sports medicine*, *41*(8), 469-480.

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

van Ingen Schenau, G. (1980). Some fundamental aspects of the biomechanics of overground versus treadmill locomotion. *Medicine and science in sports and exercise*, *12*(4), 257-261.

Vandenberg, J. M., George, D. R., O'Leary, A. J., Olson, L. C., Strassburg, K. R., & Hollman, J. H. (2015). The modified gait abnormality rating scale in patients with a conversion disorder: a reliability and responsiveness study. *Gait & posture*, *41*(1), 125-129.

VanSwearingen, J. M., Paschal, K. A., Bonino, P., & Chen, T.-W. (1998). Assessing recurrent fall risk of community-dwelling, frail older veterans using specific tests of mobility and the physical performance test of function. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, *53*(6), M457-M464.

VanSwearingen, J. M., Paschal, K. A., Bonino, P., & Yang, J.-F. (1996). The modified Gait Abnormality Rating Scale for recognizing the risk of recurrent falls in community-dwelling elderly adults. *Physical therapy*, *76*(9), 994-1002.

Verghese, J., Holtzer, R., Lipton, R. B., & Wang, C. (2009). Quantitative gait markers and incident fall risk in older adults. *Journals of Gerontology Series A: Biomedical Sciences and Medical Sciences*, *64*(8), 896-901.

Verghese, J., LeValley, A., Hall, C. B., Katz, M. J., Ambrose, A. F., & Lipton, R. B. (2006). Epidemiology of gait disorders in community-residing older adults. *Journal of the American Geriatrics Society*, *54*(2), 255-261.

Volman, M. C. J., & Geuze, R. H. (1998). Relative phase stability of bimanual and visuomanual rhythmic coordination patterns in children with a developmental coordination disorder. *Human movement science*, *17*(4-5), 541-572.

Wank, V., Frick, U., & Schmidtbleicher, D. (1998). Kinematics and electromyography of lower limb muscles in overground and treadmill running. *International journal of sports medicine*, *19*(07), 455-461.

Washabaugh, E. P., Kalyanaraman, T., Adamczyk, P. G., Claflin, E. S., & Krishnan, C. (2017). Validity and repeatability of inertial measurement units for measuring gait parameters. *Gait & posture*, *55*, 87-93.

Weiss, A., Brozgol, M., Dorfman, M., Herman, T., Shema, S., Giladi, N., & Hausdorff, J. M. (2013). Does the evaluation of gait quality during daily life provide insight into fall risk? A novel approach using 3-day accelerometer recordings. *Neurorehabilitation and neural repair*, *27*(8), 742-752.

Wheat, J. S. (2005). *The measurement of variability in coordination during locomotion*. Sheffield Hallam University (United Kingdom).

Wheat, J. S., Baltzopoulos, V., Milner, C. E., Bartlett, R. M., & Tsaopoulos, D. (2005). Coordination variability during overground, treadmill and treadmill-on-demand running. ISBS-Conference Proceedings Archive,

Whittle, C., Jobson, S. A., & Smith, N. (2022). Validity of Calculating Continuous Relative Phase during Cycling from Measures Taken with Skin-Mounted Electro-Goniometers. *Sensors*, *22*(12), 4371.

Wight, J. T., Garman, J. E., Hooper, D. R., Robertson, C. T., Ferber, R., & Boling, M. C. (2022). Distance running stride-to-stride variability for sagittal plane joint angles. *Sports Biomechanics*, *21*(8), 966-980.

Wille, C. M., Lenhart, R. L., Wang, S., Thelen, D. G., & Heiderscheit, B. C. (2014). Ability of sagittal kinematic variables to estimate ground reaction forces and joint kinetics in running. *journal of orthopaedic & sports physical therapy*, *44*(10), 825-830.

Williams, P. T. (2009). Lower prevalence of hypertension, hypercholesterolemia, and diabetes in marathoners. *Medicine and science in sports and exercise*, *41*(3), 523.

Willson, J. D., Loss, J. R., Willy, R. W., & Meardon, S. A. (2015). Sex differences in running mechanics and patellofemoral joint kinetics following an exhaustive run. *Journal of biomechanics*, *48*(15), 4155-4159.

Winogrodzka, A., Wagenaar, R. C., Booij, J., & Wolters, E. C. (2005). Rigidity and bradykinesia reduce interlimb coordination in Parkinsonian gait. *Archives of physical medicine and rehabilitation*, *86*(2), 183-189.

Winter, S., Gordon, S., & Watt, K. (2016). Effects of fatigue on kinematics and kinetics during overground running: a systematic review. *The Journal of Sports Medicine and Physical Fitness*, *57*(6), 887-899.

Woledge, R. C., Birtles, D. B., & Newham, D. J. (2005). The variable component of lateral body sway during walking in young and older humans. *The Journals of Gerontology Series A: Biological Sciences and Medical Sciences*, *60*(11), 1463-1468.

Wong, D. W.-C., Lam, W.-K., & Lee, W. C.-C. (2020). Gait asymmetry and variability in older adults during long-distance walking: Implications for gait instability. *Clinical Biomechanics*, *72*, 37-43.

Wyatt, H. E., Weir, G., Jewell, C., van Emmerik, R. E., & Hamill, J. (2021). Stable coordination variability in overground walking and running at preferred and fixed speeds. *Journal of Applied Biomechanics*, *37*(4), 299-303.

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

Xing, H., Hou, B., Lin, Z., & Guo, M. (2017). Modeling and compensation of random drift of MEMS gyroscopes based on least squares support vector machine optimized by chaotic particle swarm optimization. *Sensors*, *17*(10), 2335.

Xu, D., Quan, W., Zhou, H., Sun, D., Baker, J. S., & Gu, Y. (2022). Explaining the differences of gait patterns between high and low-mileage runners with machine learning. *Scientific Reports*, *12*(1), 2981.

Yahya, U., Senanayake, S. A., & Naim, A. G. (2022). Characterising leg-dominance in healthy netballers using 3D kinematics-electromyography features' integration and machine learning techniques. *International Journal of Biomedical Engineering and Technology*, *39*(1), 65-92.

Yamane, T., Yamasaki, Y., Nakashima, W., & Morita, M. (2023). Tri-axial accelerometer-based recognition of daily activities causing shortness of breath in COPD patients. *Physical Activity and Health*, *7*(1).

Yao, J., Guo, N., Xiao, Y., Li, Z., Li, Y., Pu, F., & Fan, Y. (2019). Lower limb joint motion and muscle force in treadmill and over-ground exercise. *Biomedical engineering online*, *18*(1), 1-12.

Zandbergen, M. A., Marotta, L., Bulthuis, R., Buurke, J. H., Veltink, P. H., & Reenalda, J. (2022). Effects of level running-induced fatigue on running kinematics: A systematic review and meta-analysis. *Gait & posture*.

Zhang, H., Song, Y., Li, C., Dou, Y., Wang, D., Wu, Y., Chen, X., & Liu, D. (2023). Validation of a Wearable System for Lower Extremity Assessment. *Orthopaedic Surgery*, *15*(11), 2911-2917.

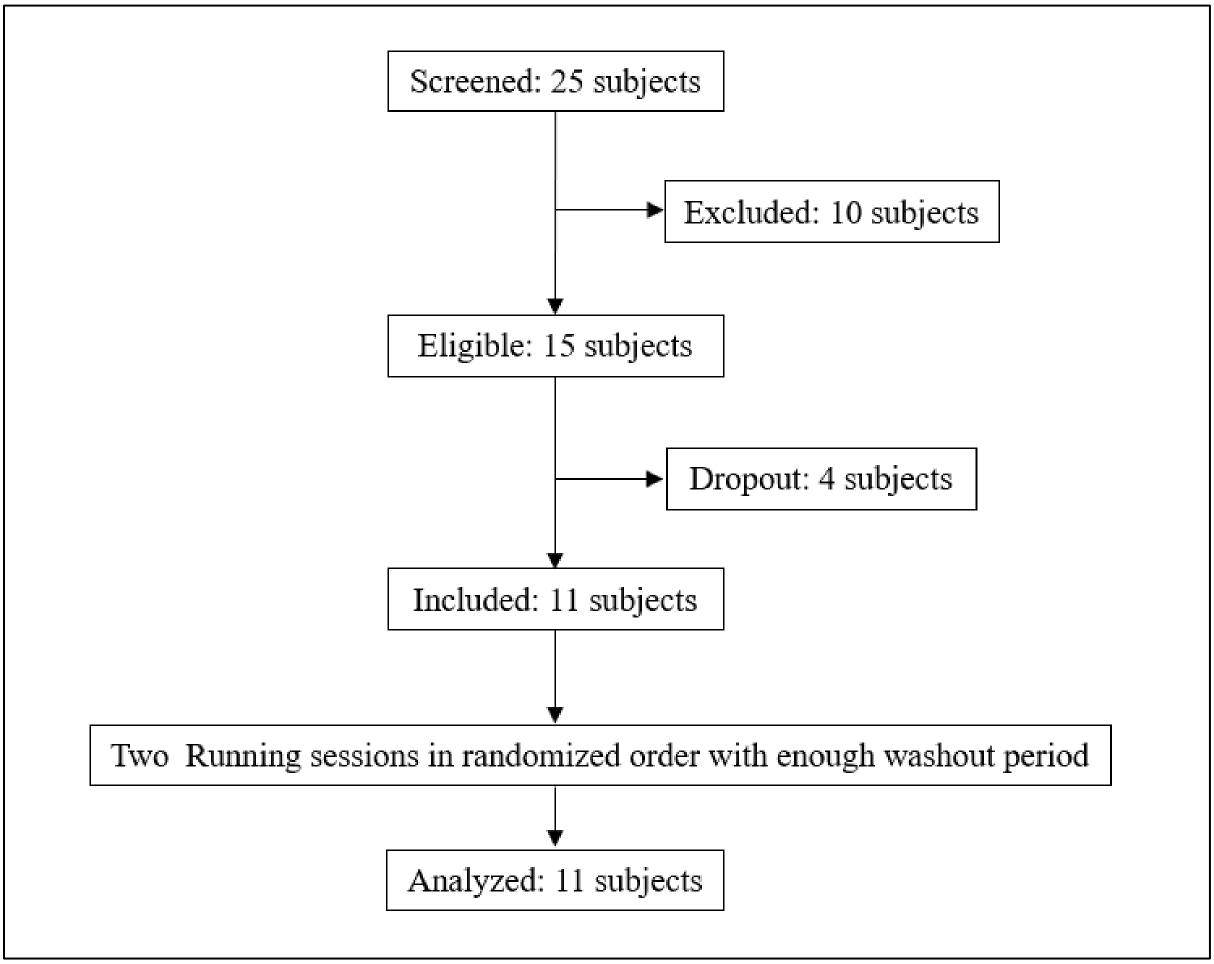
Zhang, J. H., McPhail, A. J., An, W. W., Naqvi, W. M., Chan, D. L., Au, I. P., Luk, A. T., Chen, T. L., & Cheung, R. T. (2017). A new footwear technology to promote non-heelstrike landing and enhance running performance: Fact or fad? *Journal of Sports Sciences*, *35*(15), 1533-1537.

Appendices

Appendix A: Demographic data of participants

Participant	**Gender**	**Age (years)**	**Height (cm)**	**Weight (kg)**
1	M	35	174	64.1
2	F	28	169	53.3
3	M	39	174	74.3
4	M	37	163	72.9
5	M	33	165	55.7
6	M	43	167	67.5
7	M	31	182	53.8
8	M	51	167	57.2
9	M	52	166	51.4
10	M	47	180	70.0
11	F	53	166	55.6
Mean	-	40.8	170.2	61.4
SD	-	8.9	6.3	8.5

Appendix B: CONSORT flow chart and study design: A crossover design with enough washout period to separate each running session



Appendix C: Data Collection Sheet

Stride-to-Stride Variability While Prolonged Running on a Treadmill and Over-Ground

Subject code:

Gender:

Age:

Height:

Leg length:

History of RRI or any other injury:

Running experience:

Data:

Initial and final 5 minutes – treadmill and over-ground running – to be downloaded from the IMU system and following to be further calculated

Stride time:

Joint angles:

Hip joint flexion/extension:

Knee joint flexion/extension:

Ankle joint plantarflexion/dorsiflexion:

Appendix D: Information and Consent to Participate Form

THE EDUCATION UNIVERSITY OF HONG KONG

Department of Health & Physical Education

INFORMATION AND CONSENT TO PARTICIPATE

Stride-to-Stride Variability While Prolonged Running on a Treadmill and Over-Ground

I ____________________ hereby consent to participate in the captioned research supervised by Prof. Chow, Hung Kay Daniel and conducted by Zaheen Ahmed Iqbal, who are staff and student of Department of Health & Physical Education in The Education University of Hong Kong.

I understand that information obtained from this research may be used in future research and may be published. However, my right to privacy will be retained, i.e., my personal details will not be revealed.

The procedure as set out in the **attached** information sheet has been fully explained. I understand the benefits and risks involved. My participation in the project is voluntary.

I acknowledge that I have the right to question any part of the procedure and can withdraw at any time without negative consequences.

Name of participant

Signature of participant

Date

INFORMATION SHEET

Stride-to-Stride Variability While Prolonged Running on a Treadmill and Over-Ground

You are invited to participate in a project supervised by Prof Chow, Hung Kay Daniel and conducted by Zaheen Ahmed Iqbal, who are staff and student of the Department of Health & Physical Education in The Education University of Hong Kong.

Long-distance running, as seen in marathons, have both health benefits and an increased risk of running-related injuries due to the strain on joints. Understanding how the body responds to different running environments is crucial for preventing injuries in long-distance runners. This doctoral thesis focuses on stride-to-stride variability during prolonged running, specifically looking at how stride-to-stride variability changes with running duration and surface type (treadmill vs. over-ground) using inertial measurement units in sagittal plane motions. The study hypothesizes that stride-to-stride variability varies with running duration and that there are differences in stride-to-stride variability between treadmill and over-ground running, with higher variability expected in over-ground running. It also predicts an interaction between running duration and surface type.

While extensive research has focused on gait parameters while treadmill running, limited work has addressed stride-to-stride variability particularly in joint angles and coordination comparing treadmill versus over-ground running. By examining stride-to-stride variability in stride time, lower limb joint angles, and joint coordination in natural running environments, this study seeks to provide insights into gait motor control under fatigue and tailor interventions for injury prevention and performance improvement for different groups of runners. This study is distinctive because it is one of the first to examine stride-to-stride variability in the natural environment outside the laboratory using inertial measurement units considering the factors like running duration and surface. The outcomes of this research could be instrumental in understanding gait motor control regulations, biomechanics of running, and developing strategies to prevent running-related injuries, enhance performance, and customize interventions for various runner groups to aid in rehabilitation.

Healthy recreational distance runners (between the ages of 18-40 years) who are willing to participate in prolonged running will be recruited from among university students and local residents. The inclusion criteria are as follows:

- You are a healthy adult aged 18 – 40 years
- You are recreational distance runners and are willing to participate in a prolonged running
- You should have no known running-related injuries or any other musculoskeletal injuries or pain during the last 6 months.

The exclusion criteria are as follows:

- Any known cardiovascular diseases or any other known diseases that would prevent you from participating in strenuous physical activities
- You answered "YES" to one or more questions of the Physical Activity Readiness Questionnaire (PAR-Q)
- You have any obvious anatomical abnormities such as genu valgus, genu varum, etc.

Your anthropometric data will be obtained during participant recruitment. An experienced therapist will perform a full body check. Data will be acquired using the Opal inertial measurement unit (IMU) systems (APDM, Inc., Portland, OR, USA). You will be required to run for 31 min at your preferred running speed on a treadmill and a 400-m outdoor track on separate days. The preferred running speed will be defined as the averaged self-reported comfortable running speeds. Prior to data collection, 7 IMUs will be affixed to your lower trunk (L5-S1 level) and bilateral hip, thigh, shank and foot. You will then perform warm-up exercises for 30 minutes to familiarize with the testing procedure. At the beginning of the data collection, you will be required to stand statically for 30 seconds for calibrating and initializing the IMU system. Data will be recorded continuously throughout the run at a sampling rate of 128 Hz.

This study will take place at the Education University of Hong Kong and requires 3 days with approximately 3 hours on each day.

The risk of this study is minimal as you will already have experience in long distance running. You may feel more than light discomfort during the full marathon, however, the discomfort will not be alien to you as you have experience in long distance running. You will not benefit from the experiment and the data of the study will be analyzed to test the hypothesis of this study. Your participation in the project is voluntary. You have every right to withdraw from the study at any time without negative consequences. All information related to you will remain confidential, and will be identifiable by codes known only to the researcher. As you may be aware, the results will be published in some academic journals (e.g. Journal of Sports Biomechanics).

If you would like to obtain more information about this study, please contact Prof Chow, Hung Kay Daniel at telephone number 2948-6421 or Zaheen Ahmed Iqbal at 2948-7448.

If you have any concerns about the conduct of this research study, please do not hesitate to contact the Human Research Ethics Committee by email at hrec@eduhk.hk or by mail to Research and Development Office, The Education University of Hong Kong.

Thank you for your interest in participating in this study.

Prof Chow, Hung Kay Daniel	Zaheen Ahmed Iqbal
Principal Investigator	PhD student
danielchow@eduhk.hk	[redacted]@s.eduhk.hk

香港教育大學

健康與體育學系

參與研究同意書

在跑步机和地面上长时间跑步时步幅变化

本人＿＿＿＿＿＿＿＿＿＿同意參加由周鴻奇教授負責監督，＿＿＿＿執行的研究項目。他們是香港教育大學健康與體育學系的教員/研究員。

本人理解此研究所獲得的資料可用於未來的研究和學術發表。然而本人有權保護自己的隱私，本人的個人資料將不會洩漏。

研究者已將所附資料的有關步驟向本人作了充分的解釋。本人理解可能會出現的風險。本人是自願參與這項研究。

本人理解我有權在研究過程中提出問題,並在任何時候決定退出研究，更不會因此而對研究工作產生的影響負有任何責任。

參加者姓名: ＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿

參加者簽名: ＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿

日期: ＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿＿

有關資料

在跑步机和地面上长时间跑步时步幅变化

您受邀参加一个由 Chow Hung Kay Daniel 教授指导、Zaheen Ahmed Iqbal 主持的项目，他们是香港教育大学健康与体育学系的教职员和学生。

正如马拉松比赛中所见，长跑既有益健康，又会因关节压力而增加跑步相关损伤的风险。 了解身体对不同跑步环境的反应对于预防长跑运动员受伤至关重要。 本博士论文重点关注长时间跑步期间的步幅变异性，特别是使用矢状面运动中的惯性测量单元，研究步幅变异性如何随跑步持续时间和表面类型（跑步机与地面）而变化。 该研究假设步幅变异性随跑步持续时间而变化，并且跑步机和地面跑步之间的步幅变异性存在差异，预计地面跑步的变异性更高。 它还预测跑步持续时间和表面类型之间的相互作用。

虽然广泛的研究集中在跑步机跑步时的步态参数，但有限的工作解决了步幅变化，特别是比较跑步机与地面跑步的关节角度和协调性。 通过检查自然跑步环境中步幅时间、下肢关节角度和关节协调性的步幅变异性，本研究旨在深入了解疲劳下的步态运动控制，并为不同群体制定预防损伤和提高表现的干预措施。 这项研究的独特之处在于，它是第一个利用惯性测量装置，考虑跑步时间和路面等因素，在实验室外的自然环境中检查步幅变化的研究之一。 这项研究的结果可能有助于理解步态运动控制规则、跑步的生物力学，并制定预防跑步相关损伤的策略，提高表现，并为不同跑步者群体定制干预措施以帮助康复。

从大学生和本地居民中招募身体健康、愿意参加长时间跑步的休闲长跑运动员（年龄在 18-40 岁之间）。纳入标准如下：

• 您是一名 18 – 40 岁的健康成年人

• 您是休闲长跑运动员并愿意参加长时间跑步

• 在过去 6 个月内，您应该没有已知的与跑步相关的损伤或任何其他肌肉骨骼损伤或疼痛。

排除标准如下：

• 任何已知的心血管疾病或任何其他已知的会妨碍您参加剧烈体力活动的疾病

• 您对身体活动准备问卷 (PAR-Q) 的一个或多个问题回答“是”

• 您有任何明显的解剖异常，例如膝外翻、膝内翻等。

您的人体测量数据将在参与者招募期间获得。经验丰富的治疗师将进行全身检查。将使用 Opal 惯性测量单元 (IMU) 系统（APDM, Inc.，波特兰，俄勒冈州，美国）采集数据。您将需要分别在跑步机和 400 米室外跑道上以您喜欢的跑步速度跑步 31 分钟。首选跑步速度将被定义为平均自我报告的舒适跑步速度。在收集数据之前，7 个 IMU 将固定在您的下躯干（L5-S1 水平）以及双侧臀部、大腿、小腿和足部。然后您将进行 30 分钟的热身练习，以熟悉测试程序。数据采集开始时，您需要静立 30 秒以校准和初始化

IMU 系统。 数据将以 128 Hz 的采样率在整个运行过程中连续记录。 这项研究将在香港教育大学进行，为期 3 天，每天约 3 小时。

这项研究的风险很小，因为您已经有长跑经验。 在全程马拉松过程中，你可能会感到轻微的不适，但是，由于你有长跑的经验，这种不适对你来说并不陌生。 您不会从实验中受益，研究的数据将被分析以检验本研究的假设。
您参与该项目是自愿的。 您完全有权随时退出研究，不会产生负面后果。
与您相关的所有信息都将保密，并且可以通过只有研究人员知道的代码来识别。 如您所知，研究结果将发表在一些学术期刊上（例如《运动生物力学杂志》）。

如果您想获得有关这项研究的更多信息，请联系 Chow Hung Kay Daniel 教授，电话号码 2948-6421 或 Zaheen Ahmed Iqbal，电话号码 2948-7448。

如果您对本研究的进行有任何疑问，请随时通过电子邮件 hrec@eduhk.hk 或邮寄至香港教育大学研究及发展办公室与人类研究伦理委员会联系。

感谢您有兴趣参与这项研究。

周鴻奇教授

首席研究員

Appendix E: Ethical Approval Letter

香港教育大學
The Education University of Hong Kong

9 November 2022

Mr Zaheen Ahmed IQBAL
Research Postgraduate Programmes
Graduate School

Dear Mr Iqbal,

Application for Ethical Review <Ref. no. 2022-2023-0044>

I am pleased to inform you that approval has been given by the Human Research Ethics Committee (HREC) for your research project:

Project title: Effects of Prolonged Running and Resulting Fatigue on Stride to Stride Variability

Ethical approval is granted for the project period from 9 November 2022 to 30 September 2024. If a project extension is applied for lasting more than 3 months, HREC should be contacted with information regarding the nature of and the reason for the extension. If any substantial changes have been made to the project, a new HREC application will be required.

Please note that you are responsible for informing the HREC in advance of any proposed substantive changes to the research proposal or procedures which may affect the validity of this ethical approval. You will receive separate notification should a fresh approval be required.

Thank you for your kind attention and we wish you well with your research.

Yours sincerely,

[redacted]

Patsy Chung (Ms)
Secretary
Human Research Ethics Committee

c.c. Prof CHOU Kee Lee, Chairperson, Human Research Ethics Committee

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