BIOMECHANICAL INVESTIGATION OF THE ILIOTIBIAL BAND IN RUNNING AND IMPLICATION FOR ILIOTIBIAL BAND SYNDROME

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BIOMECHANICAL INVESTIGATION OF THE ILIOTIBIAL BAND IN RUNNING AND IMPLICATION FOR ILIOTIBIAL BAND SYNDROME

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A thesis submitted in partial fulfillment of the requirements for the degree of Doctor of Philosophy

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ABSTRACT

The iliotibial band (ITB) is a thick band of connective tissue that runs along the outside of the thigh. It forms connections with the tensor fasciae latae muscle (TFL) and the gluteus maximus muscle (Gmax) in the hip region. The ITB then extends down the outside of the thigh, passing over the lateral femoral epicondyle. The ITB plays an important role in stabilizing the knee joint during various activities such as running and cycling. It functions as a tensioning element, delivering lateral stability to the knee and inhibiting excessive side-to-side motion. The ITB also interacts with various muscles, including the gluteus maximus (Gmax), gluteus medias (Gmed), tensor fasciae latae (TFL), vastus lateralis (VL), and biceps femoris (BF), to coordinate movement and maintain proper alignment of the lower extremity.

ITBS, a prevalent ailment, frequently affects individuals engaged in running activities. It is often characterized by pain on the outside of the knee, especially during activities that involve repetitive knee bending, such as running. ITBS can be caused by factors such as overuse, muscle imbalances, poor biomechanics, excessive training load, or inadequate stretching and conditioning.

Understanding the anatomy and biomechanics of the iliotibial band is important in

diagnosing and treating ITBS, as well as in optimizing performance and preventing injuries in individuals engaged in activities that place stress on the knee and hip joints.

Therefore, the current study aims to investigate the biomechanics of ITB during running and its implication for ITBS by using the musculoskeletal model with ITB and a subject-specific finite element (FE) model. The biomechanical study discussed the changes in the activation of ITB-related muscles and influential factors to the biomechanics of ITB during an exhaustive run. Furthermore, we studied the compressive pressures on the lateral femoral epicondyle applied by ITB where this area was proposed to be the 'pain zone' for ITBS under different running conditions.

In the initial segment, we investigated the impact of in-series musculature on the behavior of the ITB in healthy participants during a strenuous run. A total of twenty-five healthy participants, consisting of 15 males and 10 females, performed a 30-minute exhaustive run at a self-selected speed while wearing laboratory-provided footwear. Surface electromyography (EMG) was employed to capture muscle activities of ITB-related muscles, encompassing the TFL, Gmax, Gmed, BF, and VL. The findings indicated a declining trend in the maximum amplitudes of the TFL, Gmax, Gmed, and BF during the exhaustive run. However, the onset and offset of muscle activation remained consistent throughout the running session. These findings suggest that the behavior of the healthy ITB may be altered by the activities of the in-series musculature, potentially leading to increased compression forces applied to the lateral femoral epicondyle for knee joint stability during exhaustive running.

In the second part, we examined the effects of influential factors including exhaustion state and running speed on ITB strain and strain rate using the multibody musculoskeletal model. The strain rate in ITB has been identified as a key factor contributing to the development of ITBS. A total of 26 participants performed running trials at their normal preferred speed and a faster speed, followed by a 30-minute exhaustive treadmill run. Afterward, running trials with similar speeds to the speeds before the treadmill run were performed. The results revealed that both exhaustion states and running speeds significantly influenced ITB strain rate, with an increase observed after exhaustion for the similar speed condition before and after the exhaustive run and rapid increases in running speed. It is suggested that an exhaustion state and rapid speed changes may contribute to a higher ITB strain rate, emphasizing the importance of considering these factors in the prevention and treatment of ITBS. Running at a normal speed in a non-exhaustive state may be beneficial in managing this overuse injury.

In the third part, we aimed to investigate the interaction between ITB and the lateral femoral epicondyle using a subject-specific FE model. The compressive force between ITB and lateral femoral epicondyle, as well as the stress on the anterior cruciate ligament (ACL), were analyzed at different phases of the running cycle. The results showed that the peak values of compressive force and ACL stress occurred at a knee flexion angle of $20-30$ degrees. The study also highlighted the impingement zone between ITB and lateral femoral epicondyle, which occurs when ITB compresses against the femur during knee flexion of 30°. The findings contribute to understanding the biomechanics of ITB and its role in knee joint stability during running. The results enhance the understanding of the biomechanics of ITB during running by incorporating dynamic loading conditions. The findings have significant implications for injury prevention, rehabilitation strategies, and the customization of interventions for individuals with ITB-related problems. By considering the individual variability in biomechanics, this research provides valuable insights that can guide the development of personalized approaches to improve performance and reduce the risk of ITB-related injuries.

PUBLICATIONS

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BF	Biceps femoris muscle
EMG	Electromyography
Gmax	Gluteus maximus muscle
Gmed	Gluteus medius muscle
ITB	Iliotibial band
ITBS	Iliotibial band syndrome
TFL	Tensor fascia lata
TKEO	Teager-Kaiser energy operator
VL	Vastus lateralis muscle
FE	Finite element
MCL	Medial collateral ligament
LCL	Lateral collateral ligament
ACL	Anterior cruciate ligament

PCL	Posterior cruciate ligament

CHAPTER 1 INTRODUCTION

1.1 Background

Running has become exceedingly prevalent, with many individuals adopting regular running routines (Shipway & Holloway, 2010). Running is widely acknowledged as a pivotal component of a healthy lifestyle, and the absence of physical activity has been identified as a significant contributor to various chronic ailments (Booth et al., 2012). The benefits of running extend beyond physical health, significantly impacting mental well-being (Brene et al., 2007; Paluska & Schwenk, 2000). This popular activity has the potential to strengthen limbs and enhance blood circulation (Fields et al., 2010). Furthermore, as reported in extensive studies (Brene et al., 2007; Martinsen, 2008), running increases neurogenesis in hippocampus, making it a promising treatment for individuals experiencing depression. Various studies have explored the effects of different running durations and intensities on improving mood and overall mental health (Oswald et al., 2020).

Despite so many benefits of running, running-related injuries are very common (Malisoux et al., 2015). Physical activities, such as running, involve larger knee flexion that would lead to a higher knee joint reaction force than those with smaller knee flexion (Hart et al., 2022). The incidence rate of running-related injuries varies largely from 18.2% to 92.4% because the definitions of running-related musculoskeletal injuries are different as well as the subjects' characters (Alexandre et al., 2012). Amongst the injuries, overuse injuries are primarily injuries due to the repetitive overloading of the musculoskeletal structures over a long period of time (Saragiotto et al., 2014). Some studies have estimated that up to 70% of runners suffer an overuse injury (Ferber et al., 2009). The knee joint, the most common site, takes up nearly 50% of all overuse injuries (Rolf, 1995). A retrospective case-control analysis reported that patellofemoral pain syndrome is the most common running-related injury with an incidence rate of 16.5%, followed by ITBS with an incidence rate of 8.4% (Taunton et al., 2002).

Though overuse injuries happen a lot, few of them were explored clearly (Taunton et al., 2002). The cause of these overuse injuries is believed to be multifactorial (Ferber et al., 2009). Both extrinsic and intrinsic causes were reported to be related to overuse injuries (Gabriel & Marina, 2015). Extrinsic factors include running distance, ground stiffness, footwear, and intensity (Wilder & Sethi, 2004). Artificial surfaces as one of the extrinsic factors were explored to be related to the increased incidence of overuse injuries as they increase the mechanical stiffness (Pine, 2016). However, another study (Khaund & Sharon, 2005) revealed that the influences of the surfaces were not generalized. Meanwhile, the running pace is always reported to be one of the risk factors for overuse injuries (Lorimer & Hume, 2016). A rapid increase in running pace is generally associated with the possible incidence of injuries. A less pronated heel strike and a more lateral direction of roll-off were explored to be one factor for lower limb overuse injuries (Hesar et al., 2009). Different factors might lead to overuse injuries for different skill level runners according to their performance during a long-distance overground run (Winter et al., 2021). Overuse injuries are complex and multi-dimensional, which implies that various causes and prevention strategies should be investigated for targeted human groups. Though many studies have explored overuse injuries, risk factors, and biomechanical mechanisms remain ambiguous as not an isolated factor contributes to the injuries.

As mentioned earlier, ITBS stands out as among the most prevalent overuse injuries affecting the knee joint. It usually occurs in activities with repetitive motions of the knee joint, typically in running. The onset of ITBS is usually insidious with the characteristic of lateral knee pain during running, which could not be related to any traumatic events (Brushoj et al., 2008). Patients with ITBS feel pain in the location at the lateral femoral condyle (Lavine, 2010). Runners with or without ITBS reveal significant differences in kinematics and kinetics of lower limb joints. Female runners with ITBS demonstrated greater peak knee internal rotation angle and hip adduction angles compared to the control group, which results in a potential increase of stress in iliotibial band (ITB) (Ferber et al., 2010). Meanwhile, male runners with ITBS also revealed a significance increase in hip internal rotation and knee adduction angles, while ITB length showed no significant differences compared to the control group (Noehren et al., 2014). Moreover, during an exhaustive run, runners with ITBS exhibited abnormal segment coordination patterns, characterized by reduced variability in thigh adduction/abduction and tibia internal/external rotation couplings (Miller et al., 2008).

The classification of injury grades usually is according to the symptoms of ITBS (Lindenberg et al., 2016). Level 1 indicates that pain comes on after running but has no influence on the running distance and speed. Level 2 demonstrates that pain comes along with running without any restriction in distance or speed. Level 3 restricts running speeds or distance. Level 4 would prevent runners from running due to severe pain. The grading of the injury determines the treatments for ITBS.

Treatments for ITBS usually include conservative and surgical options (Beals & Flanigan, 2013). Conservative treatments incorporate resting, stretching and a rectification of running habits. The practical management to release the pain of ITBS patients could be divided into several phases (Fredericson & Weir, 2006). Activity modification or corticosteroid injection could be applied in case of severe pain in the acute phase (Gunter

& Schwellnus, 2004). Afterward, stretching of the iliotibial band (ITB) and hip strengthening exercises are recommended (McKay et al., 2020), which have been reported an improvement of the symptom. Strengthening exercises involve foam roll mobilization for ITB (Fredericson & Wolf, 2005). However, regardless of these studies, Ellis et al. highlighted the scare in the quality and quantity regarding the evaluation of conservative treatments of ITB (Ellis et al., 2007). For refractory cases, surgical treatments are usually applied to release the pathologic distal portion of ITB or resect the bursa underlying ITB. A surgical treatment was performed to transect the posterior half of ITB where it passes over the lateral femoral condyle, which produced a good result according to a retrospective study of 45 patients (Drogset et al., 1999). Hariri et al. (2009) reported a minimal 20-month follow-up of 12 patients with ITB bursectomy. ITB bursectomy removes the pain in ITBS patients, which allows them to involve physical activities. This technology does not need to transect ITB. However, whether a pathological bursa is in all ITBS patients or whether the bursa is the pain cause are still in debate (Hariri et al., 2009).

1.2 Formulation of Research Question

The function of ITB is the passive resistance to the hip adduction and internal rotation and internal rotation of the tibia (Reed et al., 2010). The ITB's resistance to the external adduction moment or its tensioning resulting from the compression force on the femur could aid in stabilizing the knee joint (Hutchinson et al., 2022). The aetiology of ITBS was controversial in many studies. For the early researchers, the excessive friction between ITB and lateral femoral condyle was believed to be aetiology caused by the inflammation of the soft tissues deep to ITB during repetitive flexion movements of knee joint (Muhle et al., 1999). However, this view was challenged in some studies (Fairclough et al., 2007; Geisler, 2021; Godin et al., 2017) as ITB was found that it compresses against the lateral femoral condyle firmly. Thus, some researchers inferred that ITBS is caused by the excessive compression between ITB and lateral femoral condyle (John Fairclough et al., 2006). Both theories indicated that the abnormal increases of the forces between and the lateral femoral condyle were the main aetiology.

The mechanical functions of ITB could be directly influenced by the gluteus maximus muscle (Gmax) that partly insets into ITB and the tensor fascia latae muscle (TFL) that fully inserts into ITB. Gmax pulls ITB to posterior to extend the hip, while TFL pills ITB to anterosuperior to flex the hip (Gottschalk et al., 1989). However, the force transmission pattern of Gmax and TFL through ITB still remains unclear during running or other physical activities. Greater hip adduction was reported to be one of the risk factors for ITBS resulting from the weakness of hip abductor strength (Grau et al., 2011). There is a common hypothesis that the increase of hip adduction might alter the attachment of ITB to the medial part causing in ITB strain, which leads to excessive compression between ITB and lateral femoral condyle. Participants with ITBS tend to decrease ITB strain rate with greater hip adduction compensation to relieve pain. However, hip abductor torques shows no differences for ITBS runners (Balachandar et al., 2019). Different muscle activities contribute to ITB strain rate, which might represent a proximal mechanism.

The cause of ITBS is very complex and multifactorial. Many studies reported variable contributing factors. Prospective analyses have indicated that the primary factor contributing to the development of ITBS is ITB strain and strain rate (Khan, 2009). The increase in ITB strain might deteriorate ITB during running (Khan, 2009) or apply more compression force to the lateral femoral condyle through ITB, which would induce pain (Foch et al., 2015). Though the biomechanical factors leading to ITB strain remain vague, many studies investigated three-dimensional kinematics that are considered to have connections with ITB strain. Ferber et al. (2010) demonstrated that ITB provides high rotational restraint for the knee joint, which indicates that excessive motions in the transverse plane of the knee joint would increase potential injuries. However, only a weak relationship was found between the peak stain and the kinematics of the lower limb joints (Tateuchi et al., 2015). On the contrary, the strain rate defined as the slope of the strain-time profile instead of the peak of strain was found to be one causative factor in the development of ITBS (Miller et al., 2007). Based on this theory, Boyer and Derrick (Boyer & Derrick, 2015) investigated that shortening of stride lengths might have a propensity for the decrease of ITB rate, which indicates a smaller possibility of the injury. In the meantime, it was discovered that increasing step width proves beneficial in preventing ITBS during running (Meardon et al., 2012). Biomechanical factors contributing to the increase of ITB strain rate should be investigated during running.

Based on the aforementioned studies, the research question of this study is: biomechanical cause for ITBS, the biomechanical factors of lower limb joints contributing to the increase of ITB rate, muscle force transmission between ITB and the in-series musculature, the mechanical role of ITB during running is still insufficient.

1.3 Objectives of This Study

Therefore, the objectives of this study are:

1) To analyze the impact of muscle contributions on the behavior of ITB during an exhaustive run.

2) To study the dynamics of ITB strain and strain rate throughout a long-distance running, utilizing a musculoskeletal model with ITB.

3) To explore the effects of ITB on the knee joint, employing a detailed finite element model.

1.4 Value and Significance

ITBS is a prevalent running-related musculoskeletal injury, causing significant discomfort for runners and hindering their commitment to a healthy lifestyle. With the growing number of individuals participating in running activities, it becomes crucial to delve into the mechanical behaviour of ITB and explore strategies for preventing ITBS. The primary goal of this study is to model the force transmission between ITB and the musculature in-series, unravelling its role in potential mechanical functions. The outcomes of this research aim to provide a more comprehensive understanding of the typical mechanics of ITB. Furthermore, we aim to investigate the factors contributing to the elevated ITB strain rate, recognized as a pivotal factor in the development of ITBS during running. These insights can be valuable in devising preventive measures for runners susceptible to ITBS. Additionally, the construction of a finite element model incorporating ITB will enable a thorough investigation of the biomechanics involved, potentially shedding light on the underlying causes of ITBS. Ultimately, the findings from this study can contribute to the clinical treatment of individuals dealing with ITBS, offering potential improvements for their condition.

1.5 Outline of the Dissertation

Chapter 1 provides the general overview of the iliotibial band, the significance of the iliotibial band during running, and ITB-related overuse injury (iliotibial band syndrome). The chapter states the current status of the relevant topics and proposes the questions that the current study aims to address. In the final section, the paper outlines the steps taken to bridge the gaps in existing knowledge and contribute to the field.

Chapter 2 is the literature review part. The first three sections summarize the anatomy of the iliotibial band, the coordination between the iliotibial band and knee/hip joints, and the muscle contributions to the biomechanics of the iliotibial band. The next section states the functions of the iliotibial band during running followed by reviewing the iliotibial band syndrome that is one prevalent overuse injury for runners.

Interventions for iliotibial band syndrome are also offered. The last two sections discuss the workflow of the investigation of the overuse injuries and the evaluation methods of the iliotibial band. The literature review serves the purpose of identifying research gaps and providing a comprehensive understanding of the current state of knowledge in the field. It helps to guide and inform the direction of the current study, highlighting areas that require further investigation and shedding light on potential avenues for exploration.

Chapter 3 is the overview of the adopted methods in the paper. It presents the evaluation tools utilized for analyzing different aspects of the iliotibial band biomechanics during running. Surface electromyography, musculoskeletal modeling, and finite element modeling are discussed as the key methods used for this investigation. By employing these evaluation tools, the study aims to gain valuable insights into the role of ITB in knee joint stability and its relevance to injury prevention and management during running activities.

Chapter 4 discusses the muscle activations of the muscles that could make contributions to the biomechanics of the iliotibial band, including tensor facia latae, gluteus maximus, gluteus medias, biceps femoris, and vastus lateralis. Surface electromyography is used to record the muscle activation patterns during the whole 30-minute run. The activations and fire timing of the muscles are analyzed to reveal their contributions to the iliotibial band.

Chapter 5 investigates the iliotibial band strain and strain rate before and after the exhaustive run. The impact of running speed on the iliotibial band is also under examination. To achieve this, a sophisticated musculoskeletal model that incorporates ITB is utilized to calculate the strain and strain rate of ITB during the running activity. The researchers aim to understand how the biomechanical properties of ITB change as a result of an exhaustive run and how running speed impacts these alterations. By analyzing the strain and strain rate data, the study seeks to provide a comprehensive understanding of the mechanical responses of ITB under different running conditions.

Chapter 6 analyzes the interaction between the iliotibial band and the lateral femoral epicondyle during running. By incorporating subject-specific finite element modeling, the chapter aims to gain insights into the biomechanical behavior of ITB and its effect on the lateral femoral epicondyle during the running gait. The simulations allow for a detailed investigation of how ITB compresses against the femur at various phases of the running cycle.

Chapter 7 addresses the limitations of the research conducted and explores potential avenues for future advancements in the field.

CHAPTER 2 LITERATURE REVIEW

2.1 Iliotibial Band

The ITB is a fibrous fascial tissue that extends from the tensor fascia latae muscle to the lateral tibia. It has been associated with the erection of humans from the view of the current evolutionary form. ITB passively resists internal rotation and adduction of the hip joint. And it also plays an important role in the movement of internal rotation and anterior translation of the tibia. ITB coordinated with the gluteus maximus muscle increases the stability of the hip joint. Due to the large force transmission through ITB when involved in multiple activities, ITB could be easily vulnerable to injuries. Investigation of the biomechanics of ITB could provide more insights into the injury mechanism and precautions.

2.1.1 Anatomy

The fascia latae is the deep facia in the thigh, which encases the lower limb muscles as shown in Figure 1. The ITB is a tough longitudinal band of the deep fascia (Flato et al., 2017). Proximately, ITB splits into superficial and deep layers, coalescing into the tensor facia latae muscle, the gluteus maximus muscle, and the gluteus medias muscle at the level of the greater trochanter (Muhle et al., 1999). TFL enclosed by ITB to the iliac crest inserts into ITB directly, while Gmax and Gmed partially insert into ITB. Even a partial proportion of insertion is substantial enough to consider it as an insertional tendon of these muscles (Hutchinson et al., 2022). The attachments of the fascia of the tensor fascia latae are the iliac crest (Falvey et al., 2010), anterior superior iliac spine (Birnbaum et al., 2004), and the capsule of the hip joint (Falvey et al., 2010). Due to these attachments of ITB, it significantly contributes to stabilizing the pelvis and lower extremities (Hammer et al., 2012). The function of ITB could be compared to both a ligament and a flat tendon connecting the pelvis to the tibia (Hammer et al., 2012).

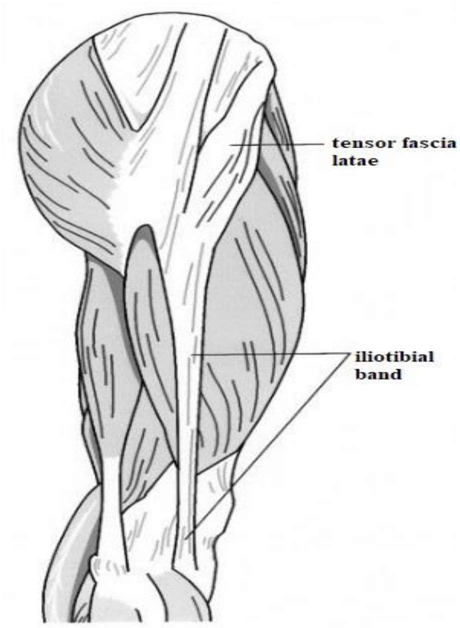


Figure 1 Lateral illustration of ITB (Muhle et al., 1999)

2.1.2 Distal Insertions

The unique functions of ITB lie in its distal insertions. Figure 2 shows the distal insertions of ITB. Superficial tibial insertion of ITB is at the Gerdy's tubercle and the deeper layer attaching on the lateral tibial tubercle of the proximal tibia (Mansour et al., 2014). The deep fibres of femoral insertions of ITB are at the distal femur bordering the biceps femoris muscle via two bundles of Kaplan fibres (Godin et al., 2017). The Kaplan functioning as a dynamic ligamentous junction connects ITB to the distal lateral femoral condyle. It redirects the forces from the TFL and Gmax (Hirschmann & Muller, 2015).

The patellar attachments of ITB are connected to the lateral aspect of the patella and the patellar tendon. Various distal attachments of ITB might represent that multiple potential force transmission patterns during different postures and physical activities.

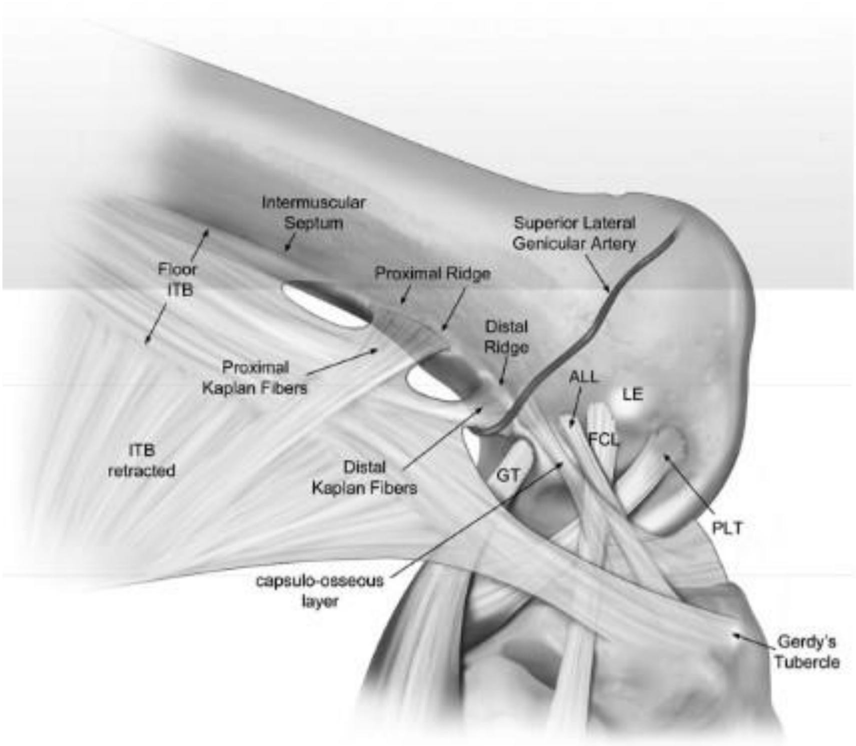


Figure 2 Illustration of the distal insertion of ITB. ALL, anterolateral ligament; FCL, fibular collateral ligament; GT, lateral gastrocnemius tendon; LE, lateral epicondyle; PLT, popliteus tendon (Godin et al., 2017)

2.1.3 Mechanical properties

The mechanical properties of ITB play a big important role in its functions during movements. Seeber et al. (2020) tested ten ITB of human cadaveric specimens including 3 males and 7femalese as shown in Figure 3. The mean initial length of ITB is 420.9 ± 55.2 mm without any force exertion. The ITB could resist 872.8 ± 285.9 N during tension-to-failure testing. Elongation of ITB is 35.9 ± 11.7 mm at the peak load, which reveals a 9.0 ± 3.9% deformation compared to the initial length. The mean stiffness of ITB is 27.2 ± 4.5 N/mm. Human ITB could resist substantial tensile forces according to the isolated human ITB testing. Wilhelm et al. (2017) illustrated an elongation response of a complex structure of ITB and tensor fascia latae muscle in-vitro. The proximal region of the complex structure sustains greater elongation in response to the clinical stretch force. Table 1 shows elastic modulus measured ex vivo (Derwin et al., 2008; Hammer et al., 2012; Steinke et al., 2012).

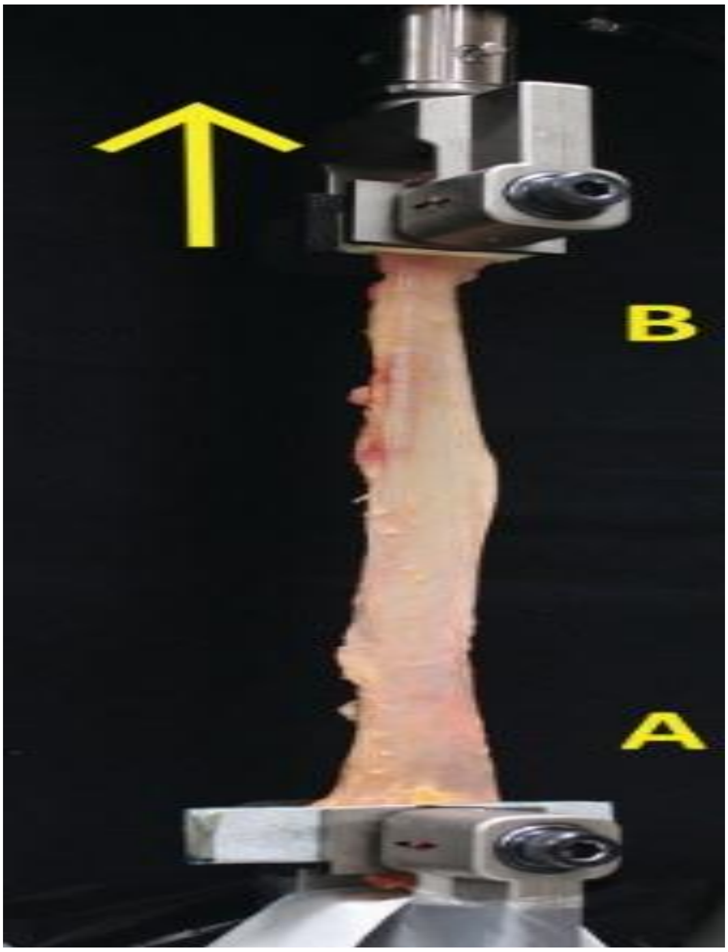


Figure 3 ITB mounting in the Materials Testing System. B: distal ITB; A: proximal ITB, Arrow represents the applied force direction (Seeber et al., 2020).

Table 1 Summary of elastic moduli of ITB

Authors	Elastic modulus (GPa)	Conclusions	Scenario
Steinke et al. (2012)	0.397 ± 0.152	chemical fixation influences elastic modulus	Ex vivo
Hammer et al. (2012)	0.084 ± 0.030 (young) 0.369 ± 0.191 (old)	Lower Elastic modulus (less stiff) in younger than in older ones	Ex vivo
Derwin et al. (2008)	0.392 ± 0.189	Processing would not change the properties of ITB	Ex vivo

The human lower limb allows for energy storage release with the assistance of spring-like tendons (Ker et al., 1987; Thorpe et al., 1999). The ITB is one kind of energy-save structure. Eng et al. (2015) explored the energy storage function of ITB using a three-dimensional musculoskeletal geometry model. The ITB was pulled by TFL and Gmax muscles to store at least 5% of all the positive work during self-selected pace running (Eng et al., 2015). Due to variability insertion in the Gmax, the posterior ITB was suggested to transform larger forces than the anterior part of ITB inserting into the TFL. Although it is not yet validated, the energetic contribution role of ITB has been well accepted (Hutchinson et al., 2022). Measuring the energetic storage of soft tissues directly remains challenging due to the complexity of their insertions and a large number of degrees of freedom involved. The function of ITB in the kinematic contributions to movements is also difficult to measure.

2.2 ITB and Joints

ITB proximately originates from the TFL and partly inserts into the Gmax muscles and inserts at the Gerdy's tubercle of the tibia and lateral femoral epicondyle. Due to the unique anatomical characteristics of ITB, ITB contributes to the movements of the hip joint and knee joint. The hip joint plays a significant role in the generation and transmission of forces during physical activities. Additionally, the hip joint is a constrained articulation of the concave socket of the pelvis and the spherical head of the proximal femur. The knee joint comprises of the distal femur, proximal tibia, and patella, which is one major load-bearing joint that transfers significant forces. Due to its distinct functions, ITB is essential in the force transmission of the lower extremities.

2.2.1 Hip Joint and ITB

The hip joint is a ball-and-socket joint that connects the pelvis to the femur (Polkowski & Clohisy, 2010). It is one of the largest and most important joints in the human body that is responsible for supporting the weight of the upper body and facilitating the movement of the lower limbs. Hip joint dysfunction can lead to pain, stiffness, and decreased mobility, affecting the quality of life for those who suffer from it. Therefore, understanding the anatomy, biomechanics, and pathology of the hip joint is crucial in diagnosing and treating hip-related disorders.

As a synovial joint, the hip joint is surrounded by a synovial membrane that produces synovial fluid to lubricate and nourish the joint. The joint is composed of the acetabulum that is the socket-shaped cavity in the pelvis, and the femoral head that is the ball-shaped end of the femur. The acetabulum is deepened by a fibrocartilaginous rim called the acetabular labrum, which increases the stability of the joint by creating a suction effect that helps hold the femoral head in place. The femoral head is covered by articular cartilage, which provides a smooth, low-friction surface for the bones to move against each other. The joint capsule surrounds the joint and is strengthened by several ligaments, including the iliofemoral ligament, pubofemoral ligament, and ischiofemoral ligament. The muscles surrounding the hip joint, including the gluteal, adductors, and hip flexors, play an important role in stabilizing and moving the joint.

The hip joint is a complex joint that allows for a wide range of motion while maintaining stability. During hip flexion, the femoral head moves forward and upward in the acetabulum, while the femoral head moves backward and downward during hip extension. The femoral head also rotates in the acetabulum during internal and external rotation of the hip. The orientation of the acetabulum and the angle of the femoral neck also affect the biomechanics of the joint. The acetabulum is angled slightly anteriorly, which contributes to the joint's stability by preventing the femoral head from sliding forward. The angle of the femoral neck, known as the neck-shaft angle, varies between individuals and affects the load distribution across the joint. A larger neck-shaft angle results in a greater proportion of the load being transmitted through the femoral head and neck, while a smaller angle results in a greater proportion of the load being transmitted through the shaft of the femur.

In the proximal insertion, ITB connects to the iliac crest and the anterior iliac spine, and passes over the greater trochanter without fixation to the femur (Birnbaum et al., 2004), which is shown in Figure 4. The femoral shaft bends stress under the human body's own weight, which leads to the compression force in the medial part and tensile force in the lateral of the femur. The tension counteracts the bending forces applied to the femur to reduce the loads of the bone. The TFL and Gmax muscles pull ITB with the intensification of the simultaneous contraction of the vastus lateralis muscle. The increase tension in ITB could reduce the force across the femur (Birnbaum et al., 2004). An enlargement of the posterior portion of ITB might lead to external snapping hip. It is caused by ITB sliding over the great trochanter of the femur during the extension and flexion of the hip (Zhang et al., 2021). The force angle of ITB applied to the greater trochanter puts an effect on the hip joint force toward the femoral head. The ITB was revealed to have a decisive effect on the centralization of the hip joint (Birnbaum et al., 2004). Hip joint abduction relax ITB, whereas the adduction of the hip joint tightens ITB. The ITB assists hip joint abduction and rotation and allows for pelvic standing asymmetry (Rajakulasingam et al., 2021).

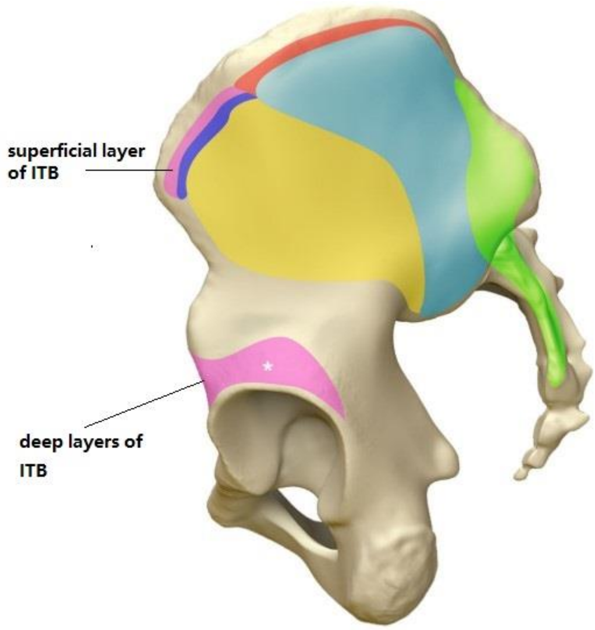


Figure 4 Illustration of ITB connections with pelvis

2.2.2 Knee Joint and ITB

The knee joint is a complex joint that is composed of bones, cartilage, ligaments, tendons, and other soft tissues. The anatomy of the knee joint is important for knee joint disorders. The knee joint comprises three bones: the femur, tibia, and patella. The femur is the thigh bone, and it articulates with the tibia to form the main hinge joint of the knee joint. The femur has a large, rounded end called the femoral condyle, which articulates with the tibial plateau, a flat surface on the top of the tibia. The femur also has a bony prominence on its anterior surface called the patellar surface, which provides a groove for the patella to move within. The tibia is the shin bone, and it forms the lower half of the knee joint. The tibial plateau is divided into two regions, the medial and lateral tibial plateaus. The medial tibial plateau is larger and more curved than the lateral tibial plateau. The tibia also has a bony prominence on its anterior surface called the tibial tuberosity, which provides attachment for the patellar tendon. The patella is a small, triangular bone that sits in front of the knee joint. It is embedded within the tendon of the quadriceps muscle group, which extends from the femur to the tibia.

The knee joint is cushioned by two types of cartilage: articular cartilage and meniscal cartilage. Articular cartilage is a smooth and white tissue that covers the surface of the femoral condyles and the tibial plateaus. It provides a low-friction surface for the bones to slide against during movement. The menisci are two crescent-shaped pieces of cartilage that sit between the femoral condyles and the tibial plateaus. They act as shock absorbers, distributing the weight of the body across the knee joint and protecting the articular cartilage from excessive stress.

The knee joint is upheld by numerous ligaments, serving to stabilize and curtail excessive movement of the joint. The medial collateral ligament (MCL) and the lateral collateral ligament (LCL) are positioned on the sides of the knee joint, acting to restrict excessive medial and lateral movement, respectively. The anterior cruciate ligament (ACL) and the posterior cruciate ligament (PCL) are situated within the joint, functioning to restrain excessive anterior and posterior movement, respectively. The knee joint is also supported by several tendons, which attach muscles to bones. The quadriceps tendon connects the quadriceps muscle group to the patella, while the patellar tendon links the patella to the tibial tuberosity. The hamstring tendons attach the hamstring muscle group to the tibia.

Due to the complex structures of the knee joint, it performs various functions such as supporting the weight of the body, facilitating movement, and absorbing shock. One of the primary functions of the knee joint is to support the weight of the body. The joint is designed to withstand large amounts of force and distribute it evenly across the joint. The weight-bearing capacity of the knee joint is due to the structure and composition of its bones, cartilage, ligaments, and tendons. The articular cartilage, which covers the ends of the bones, is particularly important in distributing the load evenly across the joint.

The knee joint enables movement through its hinge-like structure, permitting flexion and extension. In addition, the knee joint permits a limited degree of rotation and lateral movement. The movement of the knee joint is controlled by a complex system of muscles, ligaments, and tendons. The quadriceps muscle group, which is located on the front of the thigh, is responsible for extending the knee joint, while the hamstring muscle group, which is located on the back of the thigh, is responsible for flexing the knee joint.

Another crucial role of the knee joint is to dampen shock. The joint is subject to large amounts of impact during activities such as running and jumping, and the shock-absorbing capacity of the joint is essential to prevent injury. The menisci, which are crescent-shaped pieces of fibrocartilage that sit between the femoral condyles and the tibial plateaus, are particularly important in absorbing shock. The menisci act as cushions, distributing the weight of the body across the knee joint and reducing the impact on the articular cartilage.

The movement patterns of the knee joint are influenced by a variety of factors such as joint structure, muscle strength, and joint loading. During activities such as walking and running, the knee joint undergoes a series of complex movements involving flexion, extension, and rotation. The degree of knee flexion during these activities varies depending on factors such as walking speed and terrain.

The ITB is consistently regarded as the stabilizer of the knee joint in the anterolateral direction (Rajakulasingam et al., 2021). Figure 5 shows the insertions of ITB in the knee joint (Luyckx et al., 2010). Various attachment sites at the knee joint for the ITB have been documented. A direct insertion onto Gerdy's tubercle at the anterolateral tibia plateau has been observed. Due to these insertions, ITB could be involved into various knee joint movements (Huser et al., 2017) and exposed to considerable mechanical stress from the relevant muscles (Otsuka et al., 2020).

Due to distinct anatomical structures of ITB insertions into knee joints, ITB was suggested to have a significant role in the knee stability (Birnbaum et al., 2004; Merican & Amis, 2009; Stecco et al., 2013; Vieira et al., 2007). Knee stability was defined by the concept of ‘primary workers’ and ‘secondary helpers’ that is the soft tissue structures resistance to the loads from the knee joint. At the anterolateral direction, ITB contributes to the knee stability as a secondary restraint to the joint rotation laxity (Monaco et al., 2014). The ITB supports the knee joint and prevent dislocation and subluxation (Suero et al., 2013). The ITB do not function to prevent the anterior dislocation in the full extension of the knee joint, whereas ITB insertions could decrease the anterior displacement of the tibia in an anterior cruciate ligament deficient knee joint after knee joint flexes over 30° (Yamamoto et al., 2006). The shortage of these findings is that all the results were tested in the cadaveric studies, thus the forces applied on ITB is completely passive. The substantial muscle forces coming from Gmax and TFL are ignored, which is contradictory to the real physical activities. Therefore, the force transmission patterns between ITB and adjacent muscles should be investigated to reveal the real role in the knee stabilization during dynamic activities.

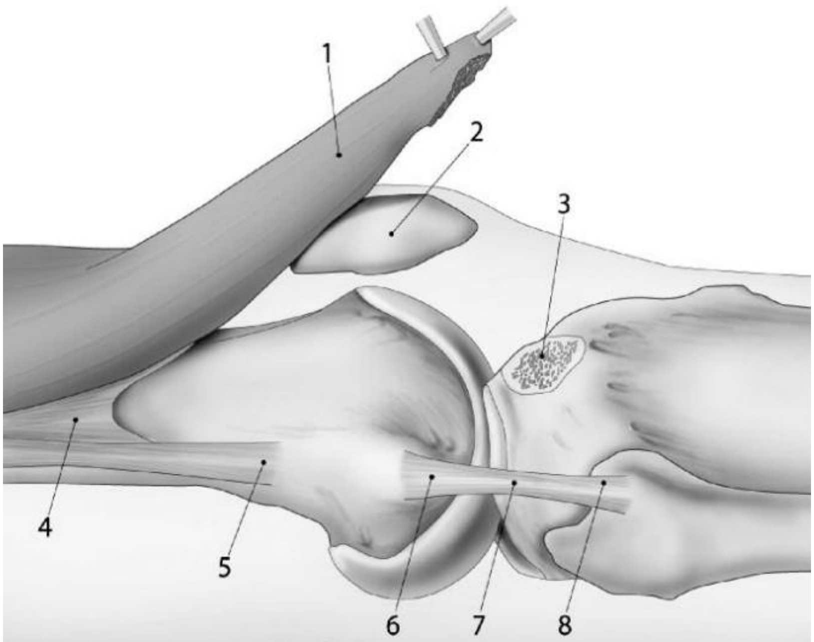


Figure 5 Illustration of the relationship between ITB and knee joint. 1: superficial layer of ITB, 2: patella, 3: ITB insertion onto the Gerdy's tubercle, 4: ITB insertion at the linea aspera, 5: insertion at the lateral epicondylar, 6: lateral epicondylar, 7: fibular collateral ligament, 8: fibula head (Luyckx et al., 2010).

The ultimate elongation of ITB was identified to be only half of the anterolateral capsule at the knee joint before rupture, which reveals that ITB would incur rupture injuries before the capsule (Rahnemai-Azar et al., 2016). Furthermore, ITB is much stiffer and thicker than the anterolateral capsule. ITB was conformed to firmly anchor the end of femur, which indicates that it could not roll over the lateral epicondyle during knee joint movements (Fairclough et al., 2007). Therefore, the distal part from the Gerdy's tubercle to the lateral epicondyle of ITB should be treat as a ligament, while its proximal part connecting to muscles could be regarded as tendons. The ITB will be tense as the knee joint bears loads. John Fairclough et al. (2006) identified tension of ITB under different positions of the knee joint. As the knee flexes progressively, the tension will transit from the anterior to the posterior region. The ITB would be tensioned by the posterior movement of the tibia. Given that ITB is essential in the force transmission of the knee joint, it incurs injuries easily.

2.3 Muscles and ITB

The ITB inserts into the TFL fully and Gmax partly (Godin et al., 2017). In the proximal position, ITB receives tension from the TFL, Gmax (Flato et al., 2017), and gluteal aponeurrosis (over the gluteus medius (Gmed) muscles). At the distal location, the contraction of the vastus lateralis (VL) and biceps femoris (BF) muscles resulted in the distortion of the ITB (Besomi et al., 2021). During running, ITB works in coordination with these muscles to ensure stability for the knee and hip joints (Foch, Aubol, & Milner, 2020). Figure 6 illustrates ITB and its adjacent muscle landmarks in the lateral view.

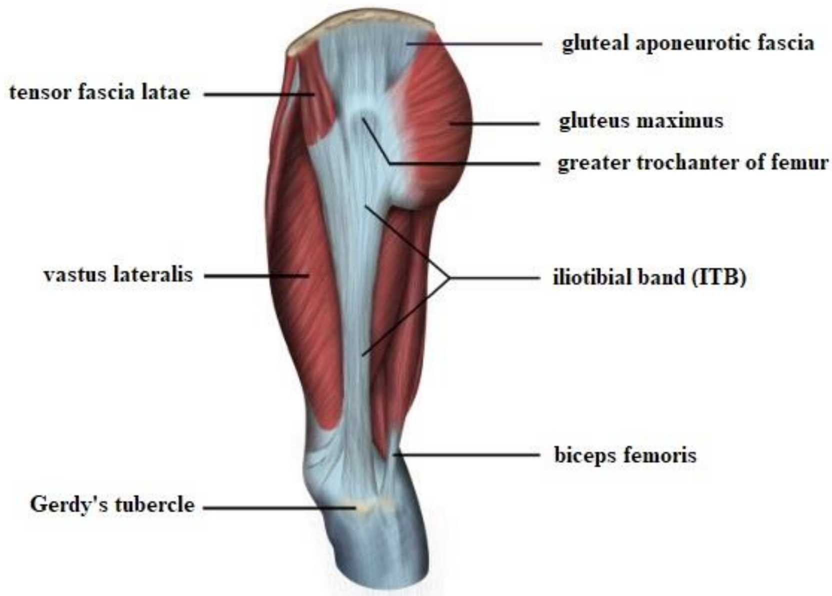


Figure 6 Lateral illustration of ITB and adjacent landmarks (Flato et al., 2017)

2.3.1 Tensor fasica latae muscle

The TFL muscle originates from the anterior superior iliac spine and inserts into the anterior iliac crest with a fusiform (Grimaldi et al., 2009). The length of the TFL is only 15cm, lying between the superficial and deep layers of ITB. It descends along with the anterolateral direction of the hip and pelvis. The entire TFL inserts into the anterior part of ITB as shown in Figure 7.

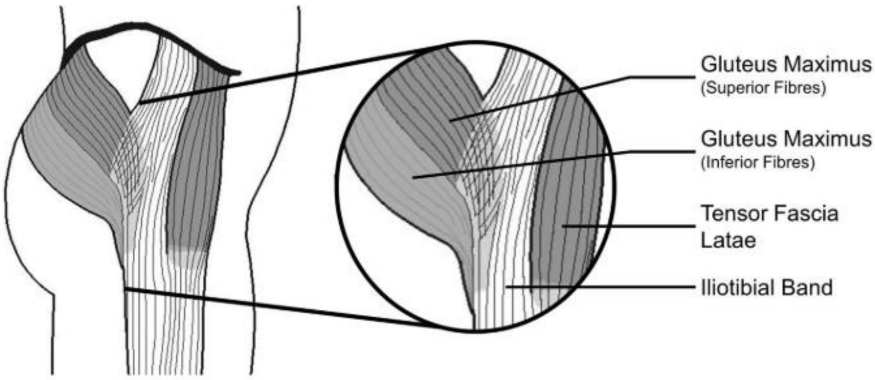


Figure 7 Illustration of the attachments of TFL (Hutchinson et al., 2022)

TFL muscle plays an important role in movements in the sagittal and transverse planes (Gottschalk et al., 1989). The activation of TFL might influence the hip joint movements. Due to the anatomical insertion to ITB, augmented activation of the TFL could enhance ITB tension (Baker et al., 2018). Thereby, TFL activity contributes to the patellofemoral joint loading through ITB tension. Generally, the TFL contributes to internal rotation and flexion of the hip joint, along with stabilization of the knee joint. Though there is a debate in the TFL's role in the abduction of the hip joint, TFL was concluded to primarily involve in the hip abduction due to a high surface electromyographic activities in an isolated abduction test (Gottschalk et al., 1989). The TFL exerts forces on the knee joint through its connection with ITB (Terry et al., 1993). Although there is a close connection between TFL and the knee joint, the TFL is not supposed to actively contribute to the movements of the knee joint but functioning via ITB. The TFL contributes to the stabilization of the pelvis in the frontal plane (Pierrynowski, 2011). The mechanical function of the TFL is impacted by the positions of the hip and knee joints, given its shared insertions with the Gmax into the ITB. Anteromedial muscle fibres of TFL were revealed to contribute the hip flexion, whereas the posterolateral fibres contribute the hip abduction and the rotation based on the EMG signal analyses (Cho et al., 2018).

2.3.2 Gluteus maximus muscle

The Gmax muscle is regarded to be the largest muscle in the human body (Carvalhais et al., 2013; Stecco et al., 2013). The upper insertion of the Gmax originates from the iliac crest, and the lower part attaches to the inferior sacrum and upper lateral coccyx (Grimaldi et al., 2009). The muscle fibres are arranged in layers and run in different directions, giving the muscle its characteristic shape and function as shown in Figure 8. One of the primary functions of the Gmax is hip extension. The lower part below the hip rotation centre acts to be the primarily hip extensor due to its large moment arm for the extension (Jaegers et al., 1992). This occurs when the thigh bone moves backward from the hip joint, as in walking, running, or climbing stairs. The Gmax contracts to pull the thigh bone back, creating a powerful force that propels the body forward. This function is particularly important in activities that require explosive movements, such as sprinting or jumping. Hip abduction refers to the movement of the thigh bone away from the midline of the body, as inside stepping or lateral lunges. Due to the upper insertion into the linea aspera of the femoral bone above the hip rotation centre the Gmax largely contributes to the abduction of the hip joint. The gluteus maximus muscle also plays a role in hip external rotation, which occurs when the thigh bone rotates outward from the hip joint. This movement is essential for activities such as turning, pivoting, and twisting, which are common in many sports and physical activities. The gluteus maximus muscle contracts to pull the thigh bone outward, helping to stabilize the hip joint and prevent injury. Due to the insertions of Gmax, it corporates with vastus lateralis and biceps to stabilize the fascia latae during movements

(Stecco et al., 2013). Finally, the gluteus maximus muscle also plays a vital role in the stabilization of the hip joint. This muscle helps to keep the thigh bone cantered in the hip socket, preventing it from sliding too far forward or backward during movement. Additionally, the gluteus maximus muscle helps to maintain proper alignment of the pelvis, reducing the risk of injury and promoting optimal biomechanics during physical activity.

Figure 8 Posterior view of the thigh with Gmax and ITB (Stecco et al., 2013)

Many musculoskeletal lower limb injuries were revealed to be associated with the weakness and imbalanced strength in the Gmax (Macadam & Feser, 2019). Strengthening of the Gmax has been suggested to beneficial to many musculoskeletal injuries, such as patellofemoral pain (Earl & Hoch, 2011; Jankowski, 2012), hamstring injuries (Wagner et al., 2010), and Achilles tendinopathy (Franettovich et al., 2014). Furthermore, the management of hip abduction and external rotation is suggested to be one significant approach for knee rehabilitation (Powers, 2010).

2.3.3 Gluteus medium muscle

The Gmed has a broad and fun shape, inserting into the superior ilium and attaching to the lateral side of the greater trochanter as shown in Figure 9 (Reiman et al., 2012). The Gmed could be separated into anterior part, middle part, and posterior part. The Gmed could initiate hip abduction with the assistance of the anterior and middle parts. The posterior part of the Gmed extends, abducts, and rotates the hip joint (Reiman et al., 2012).

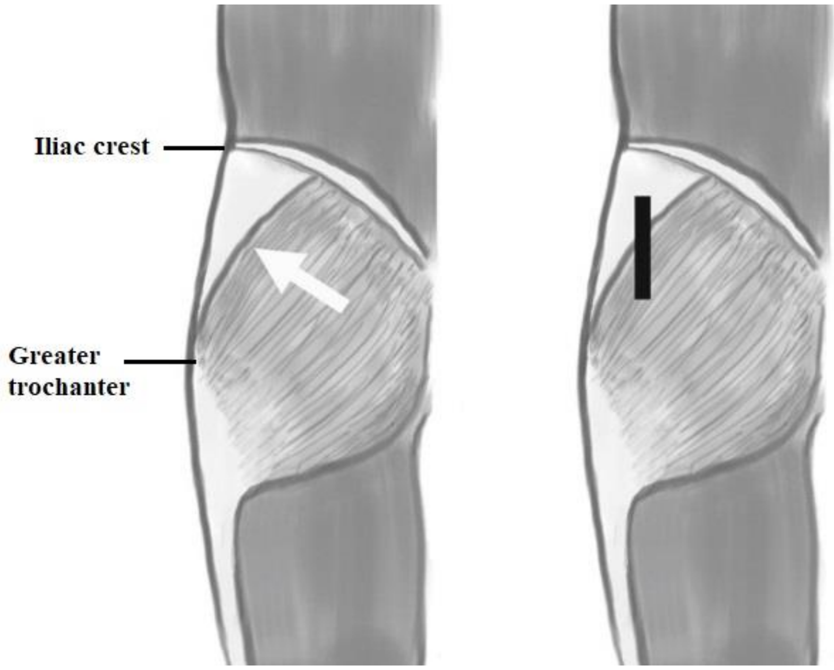


Figure 9 Schematic of the Gmed (Kim et al., 2016)

The primary function of the Gmed is hip abduction, which involves moving the leg away from the midline of the body. This movement is essential for maintaining balance and stability during physical activities, particularly when standing on one leg. The Gmed contracts to lift the leg outward, preventing the opposite hip from dropping and maintaining proper alignment of the pelvis. The Gmed also plays a role in hip extension, which involves moving the leg backward from the hip joint. This movement is crucial in various activities, including walking, running, and climbing stairs. The Gmed also contracts to help propel the body forward and maintain proper alignment of the hip joint. The final function of the Gmed is to stabilize the femur and pelvis when bearing loads during activities. The muscle activities of the Gmed measured by surface EMG revealed that it comes to the highest amplitude during the stance of gait. During running, Gmed keep the pelvis level when raising the opposite leg. Due to its large anatomical size, the Gmed could generate substantial force (Ward et al., 2010).

The Gmed is involved in the stance phase of gait, which is the period when the foot is in contact with the ground. Weakness of the Gmed can lead to compensatory movement patterns, such as increased lateral pelvic tilt and increased hip adduction, to maintain balance during gait. These compensations can result in altered gait patterns, including decreased step length, decreased walking speed, and increased energy expenditure.

The Gmed is also important for maintaining proper alignment of the lower extremity. Weakness of this muscle can lead to hip drop on the opposite side, which can result in increased knee valgus and decreased foot and ankle stability. These altered alignment patterns can lead to increased stress on the knee joint, which can increase the risk of injury. The Gmed plays an essential role in controlling the position of the femur bone during dynamic activities such as running and jumping. Weakness of this muscle can result in increased femoral internal rotation and adduction, which can increase the load on the knee joint. Weakness of the Gmed has been associated with an increased risk of lower extremity injuries, including patellofemoral pain syndrome, ITBS, and ACL injuries.

2.3.4 Vastus lateralis muscle

The VL muscle as a part of the quadriceps muscle group acts to be the extensor of the knee joint (Becker et al., 2009; Bohm et al., 2018; Miyamoto et al., 2019; Staron et al., 2000). The VL muscle could be split into four anatomical parts. The superficial proximal part of VL inserts to the superficial of the aponeurosis. The deep proximal part is under the deep aspect of the VL muscle and inserts into the proximal aponeurosis. The central partition locating in the middle is the dominant part of the VL muscle. The deep distal part of the

VL muscle has an additional attachments into ITB as shown in Figure 10 (Becker et al., 2010). The superficial aponeurosis originates from the greater trochanter and enwraps the proximal portion of the VL. The deep aponeurosis derived from the middle of the muscle and attaches to the patella (Toia et al., 2015). It was reported to be involved in the pathogenesis concerning the knee joint due to its distinct anatomy insertions (Neptune et al., 2000), such as ITBS (Vieira et al., 2007).

Figure 10 The lateral view of the distal insertions of ITB and VL (Becker et al., 2010)

The VL muscle is a powerful knee extensor and plays a significant role in force production during lower limb movements. It works in conjunction with the other quadriceps muscles to generate force and torque around the knee joint, which is necessary for movements such as running, jumping, and kicking.

The VL muscle also plays a role in joint stabilization, particularly during weight-bearing activities. As the body weight is transferred through the knee joint, the VL muscle helps to stabilize the joint and prevent excessive movement. This function is critical for preventing knee injuries, such as ligament tears and meniscus injuries.

In conclusion, the VL muscle plays a crucial role in knee biomechanics, particularly during the stance phase of the gait cycle. It helps to control knee flexion and extension during weight-bearing activities, such as walking and running. The VL muscle also works with other muscles in the lower limb, such as the hip abductors and adductors, to maintain proper alignment of the knee joint and prevent excessive medial or lateral movement.

2.3.5 Biceps femoris muscle

The BF muscle is a component of the hamstring group, which comprises three muscles. It is located on the back of the thigh and is responsible for several important functions, including hip extension, knee flexion, and external rotation of the hip. The BF muscle consists of two heads: the long head and the short head. These two heads work together to provide various functions, including knee flexion and hip extension, and they play a crucial role in the overall movement and stability of the lower extremity. Certainly, the long head of the biceps femoris muscle originates from the ischial tuberosity of the pelvis, whereas the short head originates from the linea aspera of the femur. These distinct points of origin allow the two heads to act in coordination to facilitate different movements and functions, contributing to the muscle's overall role in the lower limb biomechanics. The two heads of the BF muscle then merge to form a common tendon, which inserts onto the head of the fibula and the lateral condyle of the tibia as shown in Figure 11.

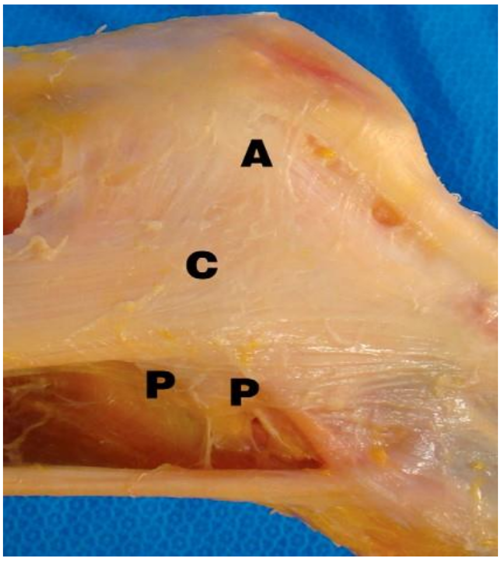


Figure 11 The structure of ITB and BF. A: attachment to the patella, C: attachment to the Gerdy's tubercle, P: attachments to the BF muscle

The BF muscle is one of the primary hip extensors. During activities such as walking, running, and jumping, the BF works in conjunction with the other hip extensors, such as the gluteus maximus muscle and the hamstrings, to generate force and propel the body forward.

The BF muscle is also a key knee flexor. During activities such as squatting, lunging, and jumping, the BF works in conjunction with the other knee flexors, such as the hamstrings and the gastrocnemius, to generate force and stabilize the knee joint. The BF muscle also plays a role in externally rotating the hip. This function is particularly important during activities such as cutting and pivoting, which require quick changes in directions. The BF muscle is essential in athletic performance, particularly in sports that require explosive lower limb movements. Strong, well-conditioned BF muscles can improve sprinting speed, jumping ability, and agility, while also decreasing the risk of injury.

2.4 Functions during running

2.4.1 Gait circle during running

The gait cycle during running is a highly coordinated and complex sequence of movements that involves the interaction of various muscles and joints. The gait cycle is comprised of two phases: the stance phase and the swing phase.

The stance phase begins when the foot contacts the ground and ends when the foot leaves the ground. This phase can be further divided into five sub-phases, which include initial contact, loading response, mid-stance, terminal stance, and pre-swing. During the initial contact sub-phase, the foot makes contact with the ground, and the heel begins to lower. As the knee begins to flex, the hip simultaneously starts to extend. The muscles of the hip, thigh, and calf work together to control the motion of the leg and absorb the shock of impact. In the loading response sub-phase, the foot continues to flatten, and the body weight is transferred to the foot. Following knee flexion, the hip also continues to extend. The muscles together to control the motion of the leg and stabilize the body. During the mid-stance sub-phase, the foot is flat on the ground, and the body weight is directly over the foot. The terminal stance sub-phase begins when the heel begins to rise, and the foot begins to lift off the ground. The muscles propel the body forward.

The swing phase begins when the foot leaves the ground and ends when the foot makes contact with the ground again. This phase can be further divided into three sub-phases, which include initial swing, mid-swing, and terminal swing. During the initial swing sub-phase, the leg begins to swing forward. The knee continues to flex, and the hip begins to flex. In the mid-swing sub-phase, the knee reaches its maximum flexion, and the hip continues to flex. The muscles work together to control the motion of the leg and prepare it for the terminal swing sub-phase. During the terminal swing sub-phase, the leg begins to extend, and the foot prepares to contact the ground again. The muscles propel the leg forward and prepare it for the next stance phase.

The biomechanics of the gait cycle during running are complex and involve the interaction between the body and the ground. When the foot contacts the ground, a vertical force is applied to the foot, and the body weight is supported by the leg. This force causes the foot to flatten, and the muscles of the leg and foot work together to absorb the shock of impact.

As the body moves forward over the foot, a horizontal force is generated, which propels the body forward. The hip, knee, and ankle joints work together to control the motion of the leg and stabilize the body. The muscles generate the force required to propel the body forward and maintain balance during the gait cycle.

In summary, the gait cycle during running is a complex and highly coordinated sequence of movements that involves the interaction of various muscles and joints. Understanding the biomechanics of the gait cycle is important for improving athletic performance, preventing injuries, and developing rehabilitation programs for individuals with gait disorders. Further research is needed to investigate the complex interactions between the body and the ground during the gait cycle and to develop effective interventions for individuals with gait disorders.

2.4.2 Kinematics

Runners with ITBS commonly feel pain in the lateral femoral condyle during long distance running. Grau et al. (2008) examined 52 healthy runners and 18 ITBS runners with matched sex, height, and mass, and they found runners with ITBS exhibit lower amplitude of hip adduction angle and eversion angle of the subtalar joint. Runners with

ITBS were also found to exhibit abnormal segmental coordination during fatigue running, particularly in couplings involving tibia internal/external rotation and thigh abduction/adduction (Miller et al., 2008). In another study, Ferber et al. (2010) revealed that competitive female runners with ITBS exhibited a significant increase in the peak hip adduction angle and internal rotation angle of the knee joint compared to healthy female runners. Grau et al. (2011) also examined the kinematics of ITBS runners under the condition of running barefoot at 3.3 m/s along the runway. Runners with ITBS indicated a decrease of the hip adduction and range of hip joint motion, and also a lower velocity of the hip flexion and knee flexion. A principal components analysis approach was applied to discover the differences between healthy runners and ITBS runners (Foch & Milner, 2014b), which indicate a less hip adduction angle in ITBS runners than that of the healthy runners during stance phase. However, another study (Foch et al., 2020) demonstrated that no significant differences between ITBS female runners and controls were discovered in the peak hip adduction angle, and peak flexion of the hip joint, and peak internal rotation of the hip joint except for an significant decrease of the hip adduction excursion. Meanwhile. Foch and Milner (2019) revealed that female ITBS runners demonstrate a higher coordination variability in the frontal plane pelvis-frontal plane thigh and frontal plane hip-transvers plane hip.

Hunter et al. (2014) examined biomechanics of one recreational female ITBS runners during nine real-time running-retraining sessions. As the running time for ITBS runner increases, the pelvis external rotation angle decreases. In another study, Ober test was performed to 17 ITBS runners and 17 healthy runners, which indicated that ITBS runners indicate a significantly lower Ober measurement, and a weaker hip external rotators, and a bigger knee adduction angle, and a greater peak angle of the hip internal rotation (Noehren et al., 2014). Running to fatigue was exposed to decrease the peak amplitude of the hip adduction angles in ITBS runner compared to healthy control group (Brown et al., 2016). Tateuchi et al. (2016) adopted the shear-wave modulus to measure the shear elastic modulus in ITB to identify ITB stiffness under 7 single leg standing conditions. They revealed ITB stiffness increases significantly when the pelvic stays in the posterior tilted position, contralateral drop position, and contralateral posterior rotated position, which indicated an increasing possibility of ITBS. Dodelin et al. (2018) used orthotic insoles to treat three runners with ITBS in three-week training. After the training, all ITBS runners felt less pain significantly and they showed a decrease in the mean peak of the internal rotation of the hip and knee joint. Table 2 summarized the kinematics of the runners with ITBS.

Table 2 A summary of the kinematics of ITBS runners

Authors	Participants details	Protocol	Outcome variables	Significant findings
Grau et al. (2008)	52 healthy runners, 18 runners with ITBS	Run barefoot at self-selected speed	Hip Abduction and Adduction, tibia rotation, Eversion and Inversion of the Foot.	A lesser peak of hip adduction angle in ITBS runner (p=0.008); A slight external rotation of the tibia in relation to the foot (p=0.008); Less eversion angle of the subtalar joint in ITBS group.
Miller et al. (2008)	Eight healthy runners; eight runners with ITBS	Run to exhaust within 20min at a self-comfortable constant speed	Lower limb kinematic couplings	Runners with ITBS have less continuous relative phase variability in tibia rotation-rearfoot motion and rearfoot motion-thigh ab/adduction, and more variability in knee joint extension/flexion-foot ad/abduction.
Ferber et al. (2010)	Thirty-five female healthy runners and 35 ITBS age-matched	Running with the lab-providing neutral shoes	Peak angles of the hip joint, knee joint, and ankle joint	ITBS group revealed a significant increase of the peak angle of the knee internal rotation and hip adduction.

	runners			
Bauer and Duke (2011)	20 ITBS runners including 12 females and 8 males; 20 runners without ITBS including 9 females and 11 males	Treadmill running with subject's own shoes. Three minutes for each treadmill condition: uphill graded 5°, downhill graded 5°, and level ground	Knee flexion, tibial rotation	ITBS runner shows a lower knee flexion angle compared to the control group
Grau et al. (2011)	18 ITBS runner (13 males and 5 females), 18 healthy runners (13 males and 5 females)	Barefoot running at 12 km/h	Maximum angle of the lower extremities' joints, maximum velocity of the lower limb joint motion.	Runners with ITBS show less hip adduction than the control group and also lower maximum velocity of the hip joint and knee flexion
Foch and Milner (2014b)	Forty runners at the age 18-45, including 20 healthy runners and 20 ITBS runners	Overground running at 3.5 ± 0.18 m/s	Lower limb joint angles	Runners with previous ITBS show less hip adduction during stance phase
Hunter et al. (2014)	One recreational female ITBS runner	Treadmill running	Peak angles in the transverse plane of the foot, knee, and pelvis	The external rotation angle of the pelvis decrease.
Noehren et al. (2014)	Thirty-four participants including 17 ITBS runners and 17 healthy runners	Overground running	Kinematics of the hip and knee joint	Runners with a history of ITBS exhibit a greater adduction angle of the knee joint and hip internal rotation angles
Brown et al. (2016)	Twenty uninjured runners and 12 female ITBS runners	Treadmill running to fatigue	Peak angle of the hip adduction and internal rotation	Fatigue leads to a decrease of the peak amplitude of the hip adduction angle in ITBS runner compared to control group
Tateuchi et al. (2016)	Fourteen healthy participants	Single leg standing under 7	Shear elastic modulus, angles of the hip and	The stiffness of ITB increases significantly when

		conditions including one control condition: normal single leg standing	joint angles	the pelvic tilts posteriorly including hip extension, contralaterally drops including hip adduction, and contralaterally posterior rotates including hip external rotation
Dodelin et al. (2018)	Three runners with ITBS	Three weeks of training with orthotic insoles	Pain scale, peak angle of the hip and knee joint, peak hip adduction angle	Mean pain scale decreases significantly, mean peak internal rotation of the hip joint and knee joint decrease significantly. The hip adduction shoes no effects.
Foch and Milner (2019)	18 ITBS female runners and 18 healthy female runners	Overground running	The coordination patterns of the lower limb	Female runners with ITBS shows greater coordination variability in the frontal-transvers hip plane and in frontal pelvis plane-frontal thigh plane
Foch et al. (2020)	Thirty female participants including 15 controls	A 30-minute treadmill running with a moderate pace	Hip neuro-mechanics	Only hip adduction excursion shows a significant decrease and no significant effects in peak hip adduction angle, peak hip flexion, and hip internal rotation angle
Hamstra-Wright et al. (2020)	Nine female runners with ITBS and eight healthy runners	Treadmill running	Discrete joint angles	Runners with ITBS shows an greater peak hip adduction compared to the control group during running.

Although there are many kinematics studies investing ITBS, the kinematics of the lower limb joints appear to be considerably contradictory. Furthermore, whether the changes in the running kinematics are the causes of ITBS or the compensatory strategies (Miller et al., 2007) for accommodation and relief of pain remains unclear.

2.4.3 Kinetics

The kinetics of the hip, knee, and ankle joints of ITBS runners appear to show considerable differences when compared to healthy runners (Miller et al., 2007; Worp et al., 2012b). Despite many studies investigating the differences in the kinetics between ITBS runners and the healthy runners, contradictory results were concluded in some other studies (Brown et al., 2016; Ferber et al., 2010; Foch & Milner, 2014a, 2014b). Noehren et al. (2007) recruited a large group of runners to study running-related injuries following two-year research. The incidence of ITBS was revealed to be 16% among all the injuries. Then they conducted a kinetic assessment of ITBS runners, of which they found no significant differences between ITBS runners and the control group (Noehren et al., 2007). Grau et al. (2008) found no significant differences in the foot plantar pressure distribution between ITBS group and the matching control group. To display the contradictory results from previous studies, a summary of the kinetics of the runners with ITBS was displayed in Table 3.

Table 3 A summary of the kinetics of ITBS runners

Authors	Participants details	Protocol	Outcome variables	Significant findings
Fredericson et al. (2002)	Five male elite-level distance runners	Three conditions of ITB stretches	Hip adduction moment, knee adduction moment	An overhead arm extension stretches ITB length mostly
Noehren et al. (2007)	18 ITBS runners and 18 healthy runners with matched age and sex	Overground running	Hip adduction moment, knee external rotation moment, rearfoot inversion moment	No significant differences between ITBS group and the control group
Grau et al. (2008)	52 healthy runners as the control group, 18 with ITBS	Barefoot running in the laboratory	Plantar pressure distribution	No significant differences between the control group and ITBS group
Ferber et al.	35 female	Overground	The peak of the	No significant

(2010)	runners with ITBS and 35 healthy age-matched female runners	running	hip abductor moment, peak of the external knee moment	differences were found between the two group
Foch and Milner (2014a)	17 runners with ITBS, 17 healthy runners	Overground running trials	Peak of the external knee adduction moment	Runners with ITBS shows the similar peak compared to the control group
Foch and Milner (2014b)	Forty runners including 20 ITBS runners	Overground running trials	Frontal plane knee moment	No differences were found between the two groups
Tateuchi et al. (2015)	Sixteen healthy volunteers	Single leg standing	external joint moment of hip and knee joints	External adduction moments of the hip and knee joint increase significantly in the posture with pelvic and trunk inclination toward the contralateral side of the standing leg
Brown et al. (2016)	12 runners with ITBS, 20 healthy runners	Treadmill running to fatigue	Peak of the hip abduction moment and external rotation moment of the lower limb joints	No differences between the two groups
Shen et al. (2019)	Thirty recreational runners including 15 ITBS runners and 15 healthy runners	Two overground running trials with an interval of eight weeks	Hip abductor moment	The decrease of the hip abductor moment might be a reasonable strategy to avoid ITBS

2.4.4 Muscle contributions

Runners sustain repetitive impact forces of about one to three times of the body weight. They have been suggested to be one of the major reasons for running-related injuries. As the lower limb consists of major muscle groups, they play a big important role in force transmission. Muscle activity is very complicated in a running stride, which functions to stabilize the lower extremities and promote the movement task. Muscle activity pattern refers to the sum of a group of muscles for a specific function, as illustrated in Figure 12. Muscles activate to prepare the locomotor for landing shortly before ground contact. The shortening of the muscle-tendon unit is suggested to have minor significance during the precontact phase according to the tuning function of the muscle activities. During the ground contact phase, muscles activate to execute the specific movement task while they produce joint torques to move and adjust joint stiffness. The joint torques for movements are provided by shortening the muscle-tendon unit. Muscle tuning is necessary to understand one specific control mechanism of muscle groups (Nigg & Wakeling, 2001).

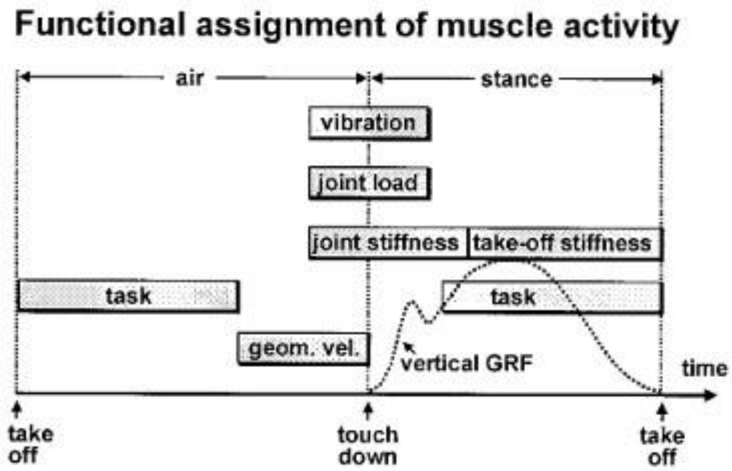


Figure 12 Illustration of the muscle activities during running (Nigg & Wakeling, 2001)

Muscle output, influenced by the muscle structure and activation characteristics, can be modified under various musculoskeletal conditions. The amplitude and timing parameters of specific muscles can be recorded during functional movements.

The ITB inserts into the TFL and the Gmax, extending into the lateral tibial condyle (Godin et al., 2017). In the proximal position, ITB receives tension from TFL, Gmax (Flato et al., 2017), and gluteal aponeurrosis (over the gluteus mediums muscles (Gmed)). Contraction of the VL and BF muscles induces distortion of the ITB at the distal position

(Besomi et al., 2021) as shown in Figure 13. The ITB works in conjunction with these muscles to ensure stability in joints during physical acitivities (Foch et al., 2020). Deficits of the hip abductors were suggested to be the main cause of the development of ITBS (Khaund & Sharon, 2005). However, Grao et.al. examined the muscle strength of the hip abductors between healthy runners and ITBS runners and suggested no significant effects (Grau et al., 2008). Additionally, ITBS runners showed no weak hip abductors (Messier et al., 1995). Given that the ITB inserts into the TFL, it experiences tension from the TFL muscle. Baker et.al.(Baker et al., 2018) revealed a significant increase in EMG amplitude in the TFL in ITBS group comparing the values at three minutes and thirty minutes during a running. On the contrary, no significant differences were investigated in a thirty-minute running test (Besomi et al., 2020).

Different muscle coordination functions were performed during running, which might be largely influenced by fatigue (Brown et al., 2016; Cowley & Gates, 2017). Thus, the examination of the effect of one single muscle on the tension of ITB might be deficient, which might induce these controversial statements in these studies.

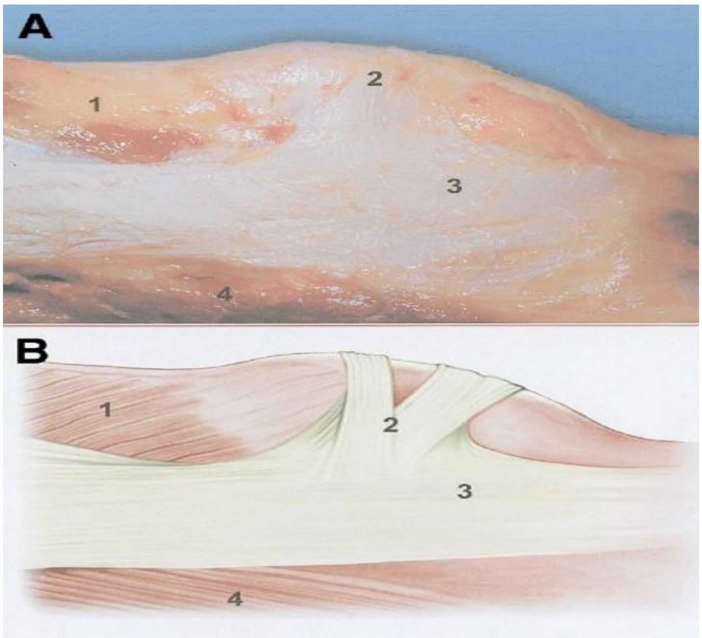


Figure 13 Distal attachments of ITB at the knee joint: (1) vastus lateralis muscle, (2) superficial oblique retinaculum, (3) superficial layer of ITB, (4) biceps femoris muscle (Vieira et al., 2007)

Muscle activation strategies have been reported to be related to overuse injuries(Cowley & Gates, 2017; Hug & Tucker, 2017). Stretch-shorting cycle activities of fatigue muscles have been related to the development of overuse injuries in lower limb (Debenham et al., 2016). Surface EMG was usually adopted to study muscle activities (Gazendam & Hof, 2007; Schache et al., 2014). EMG profiles and the activation timing of muscles could identify the variation in muscle coordination in a given motor (Hug, 2011). Modified muscle force transmission would influence the mechanical properties of ITB from an altered muscle coordination pattern (Wilke et al., 2019). Muscles near the fibular head also have interactions with ITB, which might join with the tightness of ITB (Wilke et al., 2019). Nevertheless, the functional linkage of the muscles interacted with TFL remains unclear during continuous fatigue running.

Runners engage in repetitive movements (Gaudreault et al., 2018). The excessive and abnormal increase of compression in ITB was indicated to be one of the most causes of ITB. As fatigue influences muscle activities (Edwards, 2018) and the pain caused by ITBS appears after a period of running, the current study hypothesizes that during fatigue running muscle activities would alter to be dysfunctional, which intensifies ITB tension. Muscle coordination dysfunction would predispose to ITBS under repetitive and cyclic tension in ITB. Table 4 summarized the muscle contributions to ITB from previous studies.

Table 4 A summary of muscle contributions to ITB

Authors	Participants details	Muscles Involved and Protocol	Outcome variables	Significant findings

Baker et al. (2018)	15 ITBS runners, 15 healthy runners. Each group has 7 females and 8 males	TFL, Gmax, Gmed; Treadmill running for 30 minutes	EMG data was collected at 3 and 30 minutes and then normalized to the maximal voluntary contraction	Only the activity of TFL at 3 minutes was increased significantly in ITBS group compared to the control group
Brown et al. (2019)	Twenty healthy runners and 12 ITBS runners	Gmed, TFL. Running to fatigue	Onset activation timing	The onset of the muscle activation shows no significant differences between the two groups
Foch et al. (2020)	Thirty female participants including 15 controls	Gmed; A thirty-minute moderate-pace treadmill running	The magnitude of the Gmed and duration of the activity	Both the average of the magnitude and the duration of the Gmed muscle activities reveal no significant differences between the two groups
Besomi et al. (2021)	Thirteen runners without pain	TFL, Gmax, Gmed, VL, BF; a start position in standing and then follow a slow weight shift towards the test leg	Activities of the five muscles	No significant differences were observed

2.4.5 Strain and strain rate of ITB

ITBS leads to lateral knee pain in runners (Falvey et al., 2010; Strauss et al., 2011). The pain was suggested to develop from strain in ITB as a result of the friction when ITB slides over the lateral femoral condyle (Foch & Milner, 2014a; Khan, 2009). Due to the anatomical attachment of ITB to the lateral femoral condyle of the tibia, excessive rearfoot eversion was postulated to increase ITB strain as the tibial internal rotation increases (Ferber et al., 2010). Furthermore, ITB strain and strain rate would increase when the peak of the hip adduction (Tateuchi et al., 2015) and knee internal rotation increase (Meardon et al., 2012). Excessive knee internal rotation might enlarge the torsion strain in ITB (Foch et al., 2015). However, Charles et al. demonstrated a weak correlation between the strain rate of the hip abductor muscles and the hip adduction and internal rotation of the knee joint (Charles & Rodgers, 2020). Runners might adopt a compensatory running pattern to place less pain strain to reduce pain, which illustrates the decrease of the hip adduction angle during a prolonged run (Hutchinson et al., 2022).

In addition to the kinematic factors, strain in ITB might increase due to the weakness of the hip external rotation strength (Noehren et al., 2014). Meanwhile, Kim et al. (2020) indicated that muscle activities could play an important role in ITB strain rate (Kim et al., 2020). The increase in ITB strain has a positive on the medial stability of the patellofemoral joint (Merican et al., 2009). Concurrently, as the load on ITB increases, the loads transmitted through the medial patellofemoral joint are observed to decrease significantly, as demonstrated by a cadaveric model under a non-weight-bearing condition (Gadikota et al., 2013).

Strain and tension in ITB is suggested to be one main causing factor for ITBS, which are influenced by the in-series musculature during different phases of the running stride (Hutchinson et al., 2022). A deeper understanding of force transmission within ITB and the tension and strain it sustains would provide the necessary basis and evidence to address the lack of rationale for ITBS.

2.5 ITBS

ITBS is an overuse injury that was demonstrated a maximal impingement zone at about 20~30° of the knee flexion, which causes by the repetitive compression of ITB at the lateral femoral condyle as shown in Figure 14 (Fredericson & Wolf, 2005). In ITBS runners, magnetic resonance imaging demonstrated that the distal of ITB has been thickened and the space between ITB and the lateral femoral condyle is inflamed and full of fluid (Khaund & Sharon, 2005). ITBS runners have a main symptom with sharp pain at the lateral knee joint while they are free of pain at the start of the running but feel pain suddenly after a few distances. The symptom would not exist after a long-distance run, but it returns to disturb running next running. In some severe cases, the pain would inhibit their running.

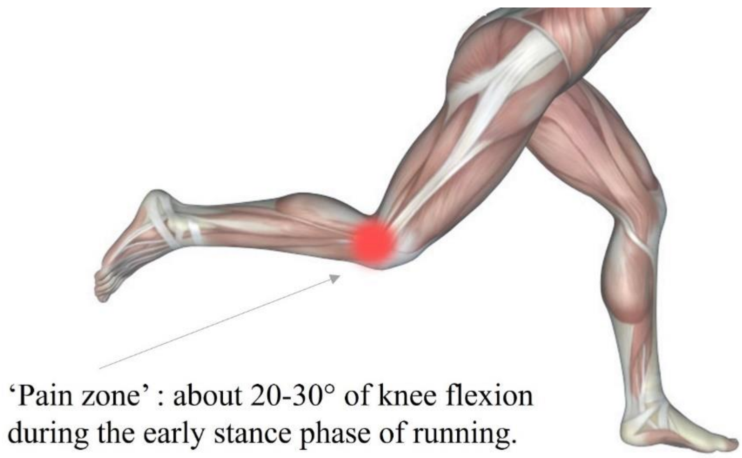


Figure 14 Illustration of pain position for runners with ITBS

2.5.1 Epidemiology

The incidence of ITBS varies depending on the population being studied and the definition used for diagnosis. The prevalence of ITBS has been reported to vary widely. Studies (Worp et al., 2012a) have reported incidence rates ranging from 5% to 14% in runners and up to 24% in long-distance cyclists . In a study of recreational runners, the prevalence of ITBS was found to be 12% (Jelsing et al., 2013). Another study reported a prevalence of 22% in a group of long-distance cyclists (Franco et al., 1997). Female athletes exhibit a higher prevalence of ITBS compared to male athletes. One study found that female runners had a higher prevalence of ITBS compared to male runners (Kim et al., 2020).

2.5.2 Biomechanical risk factors

Multiple biomedical risk factors including intrinsic factors and extrinsic risk factors have been demonstrated to have the potential predisposition to ITBS (Peterson et al., 2022). Intrinsic risk factors, such as age, gender, and body mass index (BMI), as well as anatomical and physiological characteristics, can play a significant role in the development of ITBS.

Age and gender have been identified as intrinsic risk factors for ITBS. Studies have shown that ITBS is more common in female athletes than in males, with a female-to-male ratio of approximately 2:1 (Foch et al., 2015). This may be due to differences in lower extremity alignment and neuromuscular control between males and females. In addition, older athletes may be more prone to developing ITBS due to changes in muscle mass, strength, and flexibility that occur with aging.

Anatomical characteristics, such as leg length discrepancies and abnormal hip and knee joint alignment, can also contribute to the development of ITBS. Leg length discrepancies can lead to uneven loading of the lower extremities, which can result in increased stress on the iliotibial band and surrounding structures (Hamill et al., 2008). Abnormal hip and knee joint alignment, such as excessive femoral anteversion or tibial torsion, can also lead to increased stress on the iliotibial band and surrounding structures.

One study found that runners with ITBS had a significantly greater degree of femoral anteversion compared to healthy controls, and runners with ITBS had a greater degree of internal tibial rotation during the stance phase compared to healthy controls (Noehren et al., 2014). In addition, runners with ITBS have been found to have a narrower Q angle (Orchard et al., 1996).

Physiological characteristics, such as muscle strength and flexibility, can also contribute to the development of ITBS. Muscular weakness in the hip abductors and external rotators (Fredericson et al., 2000), which are responsible for stabilizing the pelvis and controlling hip movement during running, can lead to increased stress on the iliotibial band and surrounding structures (Fairclough et al., 2007). Reduced flexibility of the iliotibial band and surrounding structures can also lead to increased tension and irritation. Several studies have found a link between muscle strength and ITBS. One study found that runners with ITBS had significantly weaker hip abductor and external rotator muscles compared to healthy controls (Ferber et al., 2010). Another study found that ITBS was associated with reduced activation of the gluteus medius muscle during running (Schreiber & Louw, 2011), indicating a potential link between muscle activation patterns and the development of ITBS.

Body mass index (BMI), a measure of body fat based on height and weight, has also been identified as an intrinsic risk factor for ITBS. Studies have shown that a higher BMI is associated with an increased risk of developing ITBS (Noehren et al., 2007). This may be due to increased stress on the lower extremities and increased tension on the iliotibial band and surrounding structures. While intrinsic risk factors, such as age, gender, and anatomical and physiological characteristics, can contribute to the development of ITBS, extrinsic risk factors related to training and equipment can also play a significant role (Taunton et al., 2002).

Training factors, such as volume, intensity, and surface, have been identified as extrinsic risk factors for ITBS (Noehren et al., 2014). Increases in training volume and intensity without proper rest and recovery can lead to increased stress on the lower extremities, including the iliotibial band. Running on hard or uneven surfaces can also contribute to the development of ITBS. One study found that runners with ITBS had significantly higher weekly training volumes compared to healthy controls (Baker et al., 2011). In addition, running on banked or uneven surfaces has been found to increase the risk of developing ITBS, potentially due to increased stress on one leg compared to the other (Taunton et al., 2002).

Equipment factors, such as shoes and orthotics, can also contribute to the development of ITBS. Worn-out or inappropriate shoes can lead to improper foot alignment and increased stress on the lower extremities. Orthotics, such as arch supports, can help correct foot alignment and reduce stress on the iliotibial band.

2.5.3 Diagnosis

Diagnosis of ITBS can be challenging, as it shares symptoms with several other knee conditions, including patellofemoral pain syndrome, meniscal tears, and lateral collateral ligament injuries. The diagnostic methods for ITBS mainly include patient history, physical examination, imaging studies, etc.

A thorough patient history is an important first step in diagnosing ITBS. The clinician takes a detailed history of the patient's symptoms, including the location, onset, and duration of pain (Baker & Fredericson, 2016). ITBS is typically characterized by pain on the outside of the knee joint, which may worsen with activity lasting, particularly when running downhill or on uneven surfaces. The patient may also report a snapping or popping sensation in the knee as they bend and straighten the leg (Baker et al., 2011). Previous injuries or surgeries that the patient may have had, as well as any relevant medical history also matter. ITBS is more common in athletes, particularly runners, and cyclists, and may be more likely to occur in individuals with certain anatomical variations, such as leg length discrepancies or abnormal foot mechanics.

A physical examination is an essential approach to the diagnostic process for ITBS. A thorough examination of the knee joint, including inspection, palpation, and range of motion testing should be performed. The examination should begin with observation of the patient's gait and posture, looking for any abnormalities or asymmetries. Palpation of the lateral femoral condyle, where the iliotibial band attaches to the knee joint, can help identify tenderness and inflammation in the area. Examination for any signs of swelling or redness in the knee joint should be also required. Range of motion testing can help identify any limitations or pain in knee flexion and extension. The patient may experience pain or tightness when the knee is bent to 30 degrees or less (Louw & Deary, 2014). In addition, provocative tests, such as the Ober test or the Noble compression test, can help diagnose ITBS by reproducing the patient's symptoms.

Imaging studies, such as MRI or ultrasound, could be useful in confirming the diagnosis of ITBS and ruling out other possible causes of knee pain, such as meniscal tears or ligament injuries (Friede et al., 2020; Goh et al., 2003; Villanueva et al., 2021). MRI can be useful in identifying inflammation and thickening of the iliotibial band, as well as any other soft tissue abnormalities in the knee joint (Wang et al., 2008). Ultrasound is a less expensive and non-invasive alternative to MRI and can also be used to visualize ITB and surrounding structures (Friede et al., 2021).

2.5.4 Current treatment modalities

ITBS is a common overuse injury that affects runners and other athletes who engage in activities that involve repetitive knee flexion and extension. There are several effective treatments for ITBS, ranging from rest and rehabilitation exercises to more advanced interventions like corticosteroid injections and surgery (Khaund & Sharon, 2005).

Rest is often the first step in treating ITBS (Fredericson & Wolf, 2005). It needs to avoid activities that aggravate the condition and give the body time to heal itself. During this time, rehabilitation exercises can be used to strengthen the muscles around the knee and hip, which can help to alleviate the pressure on the iliotibial band (Strauss et al., 2011). Runners with ITBS could gradually return to activities as symptoms subside.

Foam rolling is a form of self-myofascial release that involves using a foam roller to apply pressure to the iliotibial band (Balachandar et al., 2019). This can help to reduce tension and inflammation and increase blood flow to the affected area. Foam rolling was found to be an effective intervention for reducing pain and improving function in patients with ITBS. It is recommended that foam rolling be used as part of a comprehensive treatment plan for ITBS.

Stretching is another efficient treatment that can help lengthen the iliotibial band and reduce tension (Baker et al., 2011). Common stretches for ITBS include the standing IT band stretch and the seated butterfly stretch. Standing ITB stretching needs to stand with feet shoulder width, and then reach right arm over the head and lean to the left side. Seated ITB stretching treatment requires sitting on the ground with legs extended straight, and then cross right leg over the left leg with the right foot on the ground, and then slowly twisting torso to the right (Strauss et al., 2011). Strengthening exercises can help to improve the strength and stability of the muscles around the knee and hip, which can help to reduce the strain on the iliotibial band.

Cross-training involves engaging in low-impact activities that can help to maintain cardiovascular fitness while reducing the strain on the knee and hip (Baker et al., 2011; Fredericson et al., 2000). Cross-training activities include swimming, cycling, and using an elliptical machine. Cross-training was found to be an effective treatment for ITBS, with patients reporting reduced pain and improved function. Additionally, cross-training be used as part of a comprehensive treatment plan for ITBS, particularly for patients who are unable to engage in high-impact activities due to their condition.

In some cases, medications may be used to help manage the pain and inflammation associated with ITBS. Nonsteroidal anti-inflammatory drugs such as ibuprofen and naproxen can help to reduce pain and swelling (Gunter & Schwellnus, 2004). However, these medications should be used with caution, as they can have side effects such as stomach irritation.

Surgery intervention for ITBS is rare and is typically only considered for cases that are severe and unresponsive to conservative treatments (Strauss et al., 2011). The two main surgical interventions for ITBS are iliotibial band release and bursectomy (Fairclough et al., 2007). Iliotibial band release involves cutting the iliotibial band to relieve tension and reduce friction on the bursa. Iliotibial band release is a safe and effective treatment for ITBS, with patients reporting significant improvement in pain and function. However, caution should be noted that the procedure should only be considered as a last resort after conservative treatments have failed (McKay et al., 2020). Bursectomy involves removing the inflamed bursa to reduce pain and inflammation (Beals & Flanigan, 2013). Bursectomy was found to be an effective treatment for ITBS, with patients reporting significant improvement in pain and function. However, bursectomy is associated with a high risk of complications and should only be performed by experienced surgeons.

In conclusion, there are several effective treatments available, ranging from rest and rehabilitation exercises to more advanced interventions like corticosteroid injections and surgery. Conservative treatments such as rest, foam rolling, stretching, strengthening exercises, and cross-training should be tried first before considering more invasive options like surgery. It is important for patients to work with their healthcare provider to develop a comprehensive treatment plan that addresses their individual needs and goals. By following a structured treatment plan, many patients with ITBS can successfully manage their symptoms and return to their desired level of physical activity.

2.6 Investigation of overuse injuries

Overuse injuries are a common problem among athletes and non-athletes, and they can affect any part of the body that is subjected to repetitive stress (Hesar et al., 2009). These injuries are caused by micro-damage to bones, muscles, tendons, or ligaments, resulting from repetitive stress without sufficient time for healing and recovery (Wilder & Sethi, 2004). Overuse injuries can be painful and debilitating, and they can affect a person's ability to participate in sports and daily activities. They are also known as repetitive strain injuries, cumulative trauma disorders, and chronic pain syndromes. Overuse injuries can be classified as acute or chronic, depending on the duration and severity of the injury.

Overuse injuries are caused by repetitive stress on a particular tissue, and they can result from a variety of activities, including running and cycling (Edwards, 2018). Overuse injuries are often caused by excessive training or improper technique. The Figure 15 shows a framework to investigate overuse injuries. The causes of overuse injuries include: repetitive motion: overuse injuries often result from repetitive motions, such as running; lack of rest: overuse injuries can occur when a person does not allow enough time for the body to rest and recover between activities; poor technique: overuse injuries can result from poor technique, such as improper running form; inadequate equipment: overuse injuries can occur when a runner uses equipment that is not properly fitted or is worn out; age: as a person ages, the body becomes less able to withstand repetitive stress, increasing the risk of overuse injuries; gender: women are more prone to certain overuse injuries, such as ITBS; body weight: excess body weight puts additional stress on the joints and can increase the risk of overuse injuries; training intensity: overtraining and excessive intensity of training can increase the risk of overuse injuries; previous injuries: previous injuries can weaken the affected area, increasing the risk of overuse injuries (Poppel et al., 2021; Rolf, 1995; Wilder & Sethi, 2004; Wilgen & Verhagen, 2012; Wilke et al., 2019).

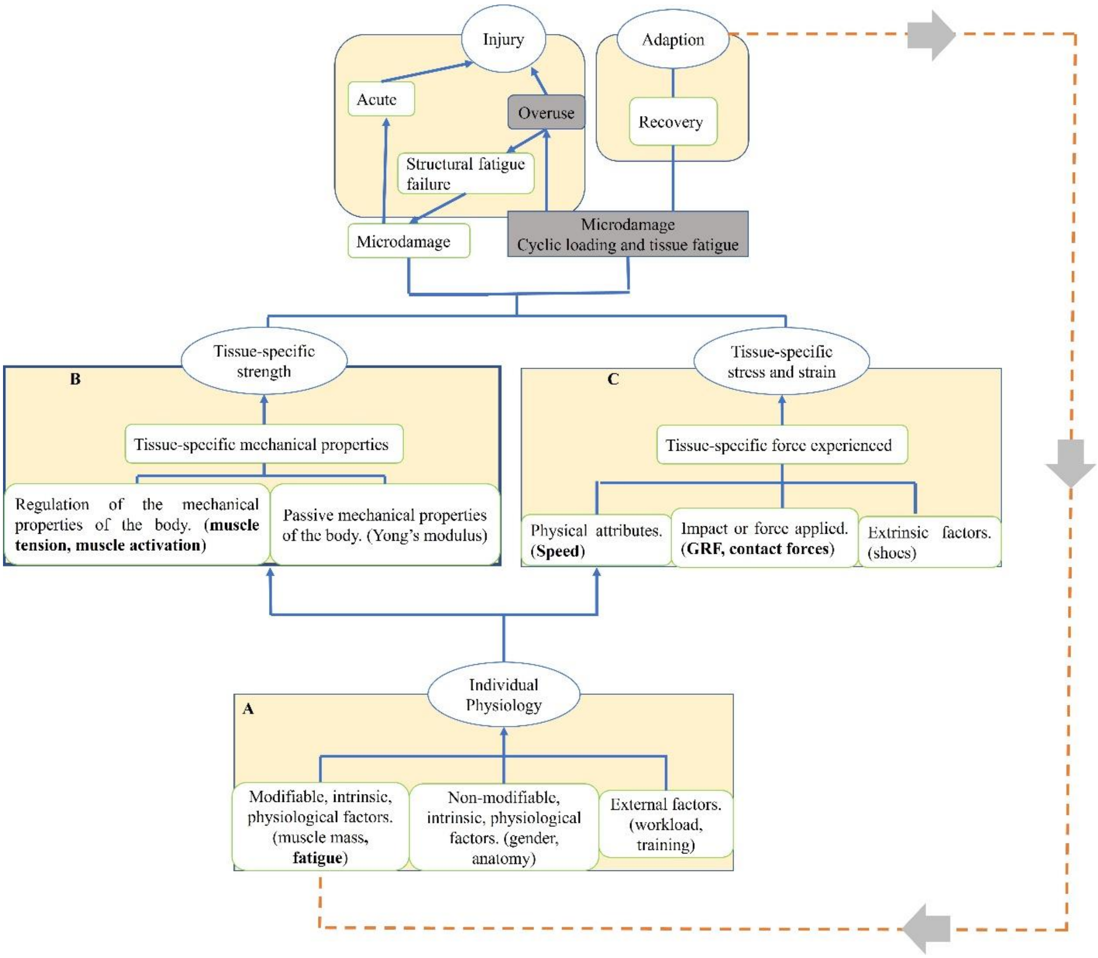


Figure 15 Schematic graph of strain-related overuse injury

ITBS is a typical overuse injury. The investigation could follow the framework shown in Figure 15. External factors, such as training intensity and fatigue state, put important influences on ITB strain and strain rate. Furthermore, physical factors, such as speed, could also affect ITB properties. Running has the potential to modify the behavior of the ITB by altering muscle activation patterns.

2.7 Biomechanical evaluation methods

2.7.1 Ultrasound evaluation

Ultrasound has been proposed to have a credible imaging modality for ITB that consists of superficial soft tissues, which is more convenient and cost-effective than magnetic resonance imaging (Flato et al., 2017). To obtain a consistent measure, the scanning protocol was usually followed that the scanning plane is longitudinal with the location at the femoral epicondyle 2cm above the joint while participants stay supine and knee flexion (Gaudreault et al., 2018). Wang et al. (2008) examined the properties of ITB for forty-four healthy young subjects including 23 men and 21 women using ultrasound as shown in Figure 16. The results showed that the width of ITB is about 5.3 mm and the thickness is about 1.9 mm using ultrasound while the width is 5.6 mm and the thickness is about 1.5 mm using magnetic resonance imaging. In addition, in another previous study (Goh et al., 2003), ITB thickness ranges from 1.3 to 2.5 mm at the level of the lateral femoral epicondyle with an overall averaged thickness of 3.4 mm. Moreover, a negative relationship was revealed between the thickness of ITB and the subject's age and weight. The thickness tends to decrease with an increase in the age of the subjects. The stiffness of ITB was examined for 14 recreational runners with ITBS and 14 healthy runners in a previous study (Friede et al., 2020), of which the shear wave elastography data were shown in Figure 17. According to the results, there were no significant differences in ITB tension between the control group and ITBS runners.

Figure 16 Ultrasound image of the knee with part of ITB (Wang et al., 2008)

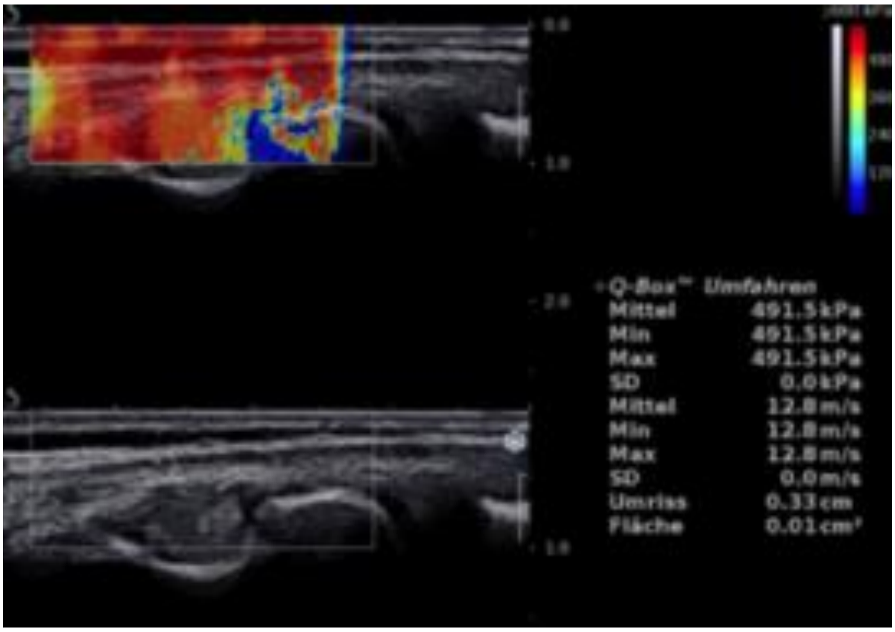


Figure 17 Shear wave elastography image of ITB

In addition to the static measurement, ultrasound could also be applied to examine the dynamic evaluation of ITB. Hutchinson et al. (2022) placed an ultrasound transducer at the distal lateral thigh along with ITB to track the displacement of ITB as shown in Figure 18. Similarly, another study adopted an ultrasound unit to measure the strain of the distal ITB in both weight-bearing and non-weight-bearing conditions (Kim et al., 2020). Five different postures. Neutral, knee flexion, hip adduction, hip adduction with knee flexion, and standing, were performed to collect the data. It was proposed that women with genu varum alignment have a higher incidence of ITBS due to the higher ITB rate under the weight-bearing condition. Table 5 lists the morphometric and dynamic outcomes related to ITB using ultrasound.

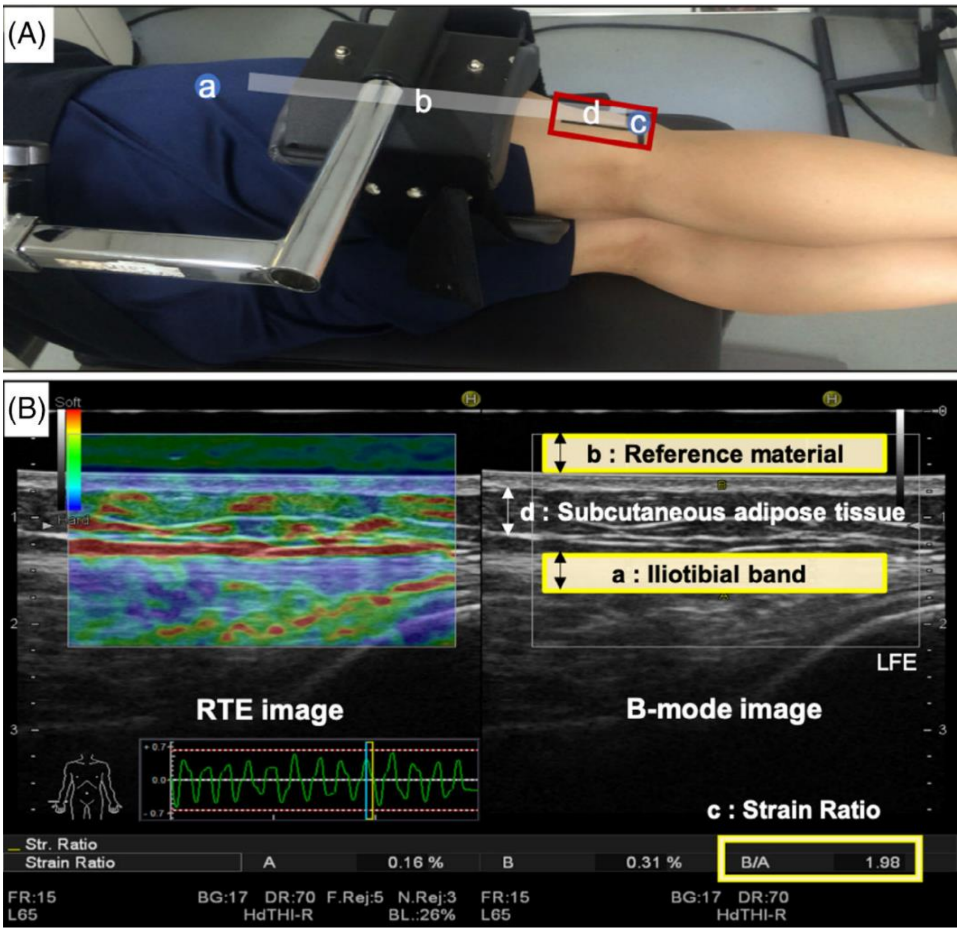


Figure 18 Measurement of the distal strain in ITB. A: measurement under a non-weight-bearing condition; B, calculation of ITB strain (Kim et al., 2020)

Table 5 Outcomes related to ITB using ultrasound.

Authors	Subjects	Protocol	Outcomes
Goh et al. (2003)	31 asymptomatic subjects including 13 men and 18 women	An ultrasound scanner with an 8-15 MHz transducer were used to measure the thickness at the level of the lateral femoral epicondyle.	The thickness of ITB ranges from 1.3 to 2.6 mm at the level of lateral femoral epicondyle. No significant difference in ITB thickness between women and men.
Wang et al. (2008)	Forty-four healthy subjects including 23 men and 21 women	Band width of ITB was assessed using ultrasound under the Ober manoeuvre in three positions: hip adduction, neutral, weight-bearing hip adduction	The band width of ITB was associated with the stretching force during hip adduction significantly.
Tateuchi et al. (2016)	Fourteen healthy subjects	Participants performed one-leg standing under 7 conditions. Then shear-wave elastography was used to measure shear elastic modulus in ITB	In the position of pelvic posterior tilt, pelvic drop, pelvic posterior rotation, the stiffness in ITB significantly increased when compared to the single leg standing.
Friede et al. (2020)	14 recreational runners with ITBS, 14 healthy controls	Ultrasound shear wave elastography	No significant differences were found in ITB tension between ITBS runners and healthy controls
Kim et al. (2020)	Forty-four healthy athletes including 21 men and 23 women	ITB strain measured by the ultrasound real-time elastography	Women with genu varum group revealed an increase in ITB strain under the weight-bearing condition.
Otsuka et al. (2020)	Twelve healthy males (*in vivo*) and twelve male cadavers	Young’s modulus of ITB in the longitudinal direction.	The stiffness of ITB is associated with the increase of hip extension angle.

2.7.2 Surface electromyography evaluation of muscles

Surface EMG is a technique used to measure the electrical activity of muscles using electrodes placed on the skin surface (Hug, 2011). The use of EMG in the evaluation of lower limb muscles has become increasingly popular in clinical settings and research studies (Bartlett et al., 2014; Mann et al., 1986; Pandy & Andriacchi, 2010).

The EMG technique has been widely used to evaluate the function of lower limb muscles in a range of clinical settings. One of the main clinical applications of EMG is in the diagnosis and treatment of neuromuscular disorders (Fagan & Delahunt, 2008). By measuring muscle activation patterns, muscle fire timing, and muscle coordination during various activities, EMG can provide valuable information about muscle function and identify any abnormalities (Kyrolainen et al., 2005).

EMG has been applied to assess lower limb muscle function following musculoskeletal injuries such as ankle sprains, knee ligament injuries, and hip fractures (Distefano et al., 2009). For example, EMG was used to evaluate the strength and activation patterns of the quadriceps and hamstrings muscles in patients with knee injuries such as ACL tears (Kakehata et al., 2021). Similarly, EMG would also be adopted to assess the function of the gluteus maximus muscle in patients with hip disorders such as femoroacetabular impingement (FAI) (Casartelli et al., 2011; Lawrenson et al., 2020). In addition to clinical applications, EMG is also used to investigate the effects of various interventions such as exercise, stretching, and neuromuscular electrical stimulation (NMES) on muscle function (Hsu et al., 2020; Reiman et al., 2012). For example, EMG has been used to investigate the effects of resistance training on muscle activation patterns and muscle strength in healthy individuals and patients with various neuromuscular disorders (Fagan & Delahunt, 2008; Nielsen et al., 2014).

The quadriceps muscles are a group of four muscles located on the front of the thigh that are responsible for extending the knee joint. EMG has been extensively used to evaluate the function of the quadriceps muscles in patients with knee injuries such as ACL tears and patellofemoral pain syndrome (PFPS) (Fagan & Delahunt, 2008). In patients with ACL tears, EMG has been used to evaluate the timing and activation patterns of the quadriceps muscles during various activities such as squatting and jumping (Myer et al., 2005; Simonsen et al., 2000). Abnormal quadriceps activation patterns have been observed in patients with ACL tears, which may contribute to knee instability and increased risk of re-injury.

The gluteus maximus muscle is the largest muscle in the buttocks that is responsible for hip extension and external rotation (Lieberman et al., 2006). EMG has been used to evaluate the function of the gluteus maximus muscle in patients with hip disorders such as FAI and hip osteoarthritis (Diamond et al., 2017). In patients with FAI, EMG has been used to evaluate the strength and activation patterns of the gluteus maximus muscle during various activities such as squatting and walking (Rutherford et al., 2018).

The gastrocnemius muscle is a muscle located in the back of the lower leg that is responsible for plantarflexion of the ankle joint. The function of the gastrocnemius muscle was examined in patients with ankle injuries such as ankle sprains and Achilles tendonitis (Darendeli et al., 2023). In patients with ankle sprains, EMG has been used to evaluate the strength and activation patterns of the gastrocnemius muscle during various activities such as walking, side-hop and running (Koldenhoven et al., 2016; Yoshida et al., 2018).

The tibialis anterior muscle is a muscle located in the front of the lower leg that is responsible for dorsiflexion of the ankle joint. The function of the tibialis anterior muscle was investigated in patients with foot disorders such as flat feet and plantar fasciitis (Nardo et al., 2013). In patients with flat feet, EMG has been used to evaluate the strength and activation patterns of the tibialis anterior muscle for children with flat feet following one year (Caravaggi et al., 2018). Abnormal tibialis anterior activation patterns might contribute to foot pain.

One of the main advantages of EMG is its non-invasive nature (Hermens et al., 1999). Unlike invasive techniques such as needle electromyography, EMG does not require the insertion of needles into the muscle, making it a more comfortable and less invasive technique for patients (Elsais et al., 2020). Moreover, EMG is a relatively simple and cost-effective technique that can be easily performed in clinical settings (Liu & Liu, 2016). Another advantage of EMG is its ability to provide real-time data on muscle activity during various activities such as walking, running, jumping, and squatting (Bartlett et al., 2014; Chumanov et al., 2011). This real-time feedback can be used to identify any abnormalities in muscle activation patterns or timing and can help to guide rehabilitation programs.

Despite its advantages, EMG has several limitations that need to be considered when interpreting the results. One of the main limitations of EMG is the potential for cross-talk from neighboring muscles (Kong et al., 2010). Cross-talk occurs when the electrical activity of one muscle is detected by the electrodes placed on another muscle, leading to inaccurate readings (Albertus-Kajee et al., 2011; Selvanayagam et al., 2012). Therefore, proper electrode placement is critical for accurate EMG measurements. Another limitation of EMG is the potential for measurement errors due to factors such as electrode placement, skin preparation, and muscle fatigue (Olin & Gutierrez, 2013). For example, if the electrodes are not placed correctly or if the skin is not prepared properly, the quality of the EMG signal may be compromised. Similarly, if the muscles being evaluated are fatigued, the EMG signals may be affected. In addition to these technical limitations, the interpretation of EMG signals can also be challenging due to the complex nature of muscle activation patterns. Therefore, EMG should be used in conjunction with other assessment tools to provide a more comprehensive evaluation of muscle function.

In conclusion, EMG evaluation of lower limb muscles is a non-invasive and cost-effective technique that can provide valuable information about muscle function in a range of clinical settings. The technique has been widely used to evaluate the function of lower limb muscles in patients with various musculoskeletal injuries and neuromuscular disorders. While EMG has several advantages such as real-time feedback and non-invasiveness, there are also several limitations that need to be considered when interpreting the results. Proper electrode placement, skin preparation, and muscle contraction level are important factors that need to be standardized to ensure accurate and reliable EMG measurements.

2.7.3 Motion capture system

Running is a complex and dynamic process that involves the interaction of various musculoskeletal structures, including the lower limb joints, muscles, and tendons (Taunton et al., 2002). Proper running biomechanics is essential for optimal performance and injury prevention, as deviations from normal mechanics can lead to increased stress on the joints and tissues, resulting in overuse injuries (Poppel et al., 2021). Motion capture systems have been widely used to evaluate running biomechanics, providing objective and detailed measurements of joint angles, kinematics, and kinetics during the gait cycle (Fukuchi et al., 2017; Riley et al., 2008).

Motion capture systems are a set of tools and techniques that allow for the measurement and analysis of human movement (Mundermann et al., 2006). These systems typically consist of multiple high-speed cameras that capture the motion of reflective markers attached to the participant's body (Barnes & Kilding, 2015). The cameras record the position of the markers in three-dimensional space, which is then used to reconstruct the movement of the participant's body. Motion capture systems can provide detailed measurements of joint angles, kinematics, and kinetics during various tasks such as walking, running, and jumping.

Running biomechanics refers to the movement patterns and mechanics of the lower limb joints, muscles, and tendons during the gait cycle (Paschalis et al., 2007). During the stance phase, the foot remains in contact with the ground, providing stability and support for the body. In contrast, the swing phase is the period when the foot is off the ground, allowing for forward movement and preparation for the next step. The stance phase can be further divided into several sub-phases, including initial contact, loading response, midstance, terminal stance, and pre-swing.

During the stance phase, the lower limb joints undergo a series of movements that absorb and generate forces, allowing for efficient propulsion and shock absorption (Jin & Hahn, 2018; Silder et al., 2015). The ankle joint undergoes plantarflexion at initial contact, followed by dorsiflexion during loading response and midstance. The knee joint undergoes flexion during loading response and midstance, followed by extension during terminal stance (Teng & Powers, 2015). The hip joint undergoes flexion during initial contact, followed by extension during midstance and terminal stance (Nigg et al., 2017). During the swing phase, the lower limb joints undergo movements that prepare the limb for the next stance phase (Kenneally-Dabrowski et al., 2019).

Motion capture systems have been used extensively to evaluate running biomechanics, focusing on various aspects of running biomechanics, including the effects of footwear, running speed, and training interventions on joint mechanics and muscle activity

(Firminger et al., 2017). Footwear has been shown to have a significant impact on running biomechanics, with different shoe types affecting joint mechanics and muscle activity (Brund et al., 2017; Hannigan & Pollard, 2020). Motion capture studies have investigated the effects of minimalist shoes, cushioned shoes, and orthotics on running biomechanics (Hoffman et al., 2015). Minimalist shoes are designed to mimic barefoot running, with a thin sole and no heel cushioning (Bergstra et al., 2015). Studies have shown that minimalist shoes lead to increased dorsiflexion and plantarflexion range of motion at the ankle joint, increased knee flexion during stance phase, and decreased impact forces compared to traditional cushioned shoes (Firminger & Edwards, 2016). Cushioned shoes, on the other hand, are designed to provide increased cushioning and shock absorption(Goss et al., 2015). Studies indicated that cushioned shoes lead to decreased ankle dorsiflexion and plantarflexion range of motion, increased knee flexion during the swing phase, and increased impact forces compared to minimalist Orthotics, which are shoe inserts designed to modify foot function, have also been investigated for their effects on running biomechanics (Bonacci et al., 2018; Roca-Dols et al., 2018). Studies have shown that orthotics can lead to decreased pronation and supination at the ankle joint, decreased knee adduction and internal rotation, and decreased vertical ground reaction forces (Hollander et al., 2015; Paquette et al., 2017).

Motion capture studies could also study the effects of running speed on joint mechanics and muscle activity during the gait cycle (Hunter et al., 2019). Studies have shown that as running speed increases, there is an increase in hip flexion and extension range of motion, a decrease in knee flexion range of motion, and an increase in ankle plantarflexion range of motion (Kruk & Reijne, 2018). Additionally, there is an increase in knee extensor and ankle plantarflexor muscle activity, while hip flexor muscle activity decreases with increasing speed (Shah et al., 2020). These findings suggest that as running speed increases, there is a shift towards more energy-efficient running mechanics, with increased reliance on the ankle plantarflexors and decreased reliance on the hip flexors.

Training interventions have been investigated for their effects on running biomechanics, with studies examining the effects of strength training, plyometric training, and gait retraining on joint mechanics and muscle activity (Beckman et al., 2016; Denadai et al., 2017; Lum et al., 2019; Roper et al., 2016). Strength training has been shown to improve running mechanics by increasing joint stability and reducing joint loading (Blagrove et al., 2018; Silva et al., 2015). Studies have shown that strength training can lead to decreased knee adduction and internal rotation, decreased hip adduction, and decreased vertical ground reaction forces during running. Plyometric training involving high-impact exercises that emphasize explosive movements, has also been investigated for its effects on running biomechanics (Giovanelli et al., 2017). Studies have shown that plyometric training can lead to increased knee and ankle joint stiffness, decreased knee and ankle joint angles, and increased muscle activity in the ankle plantarflexors. Gait retraining, which involves modifying running mechanics through visual, auditory, or tactile feedback, has also been investigated for its effects on running biomechanics (Chan et al., 2018; Miller et al., 2021). Studies have shown that gait retraining can lead to improved joint mechanics, decreased impact forces, and decreased risk of injury (Gaudette et al., 2022; Vikmoen et al., 2017).

Motion capture systems provide objective and detailed measurements of joint angles, kinematics, and kinetics during running, allowing for a better understanding of running biomechanics. Studies have shown that footwear, running speed, and training interventions can significantly impact joint mechanics and muscle activity during running. The use of motion capture systems can provide valuable information for the prevention and management of running-related injuries, as well as improving running performance. Further research is needed to establish standardized protocols for motion capture measurements and to determine the most effective interventions for improving running biomechanics.

2.7.4 Musculoskeletal multibody model

Running is a complex physical activity that requires the coordinated interaction of various body parts and physiological systems. Biomechanical analysis of running can provide insights into the underlying mechanisms of running performance, injury prevention, and rehabilitation. One way to better understand the biomechanics of running is to use musculoskeletal multibody models, which simulate the motion of the musculoskeletal system during running (Trinler et al., 2019). Two commonly used software packages for musculoskeletal multibody modeling are OpenSim and AnyBody (Alexander et al., 2021; Trinler et al., 2019).

OpenSim is an open-source software suite developed by the National Center for Simulation in Rehabilitation Research at Stanford University (Lai et al., 2017). It provides tools for building, simulating, and analyzing musculoskeletal models of the human body. OpenSim includes a library of anatomical models that can be customized to simulate specific individuals, as well as tools for modeling muscle forces and joint dynamics as shown in Figure 19 (Nitschke et al., 2020). The software also provides visualization tools for examining the motion of the musculoskeletal system during running (Verheul et al., 2023).

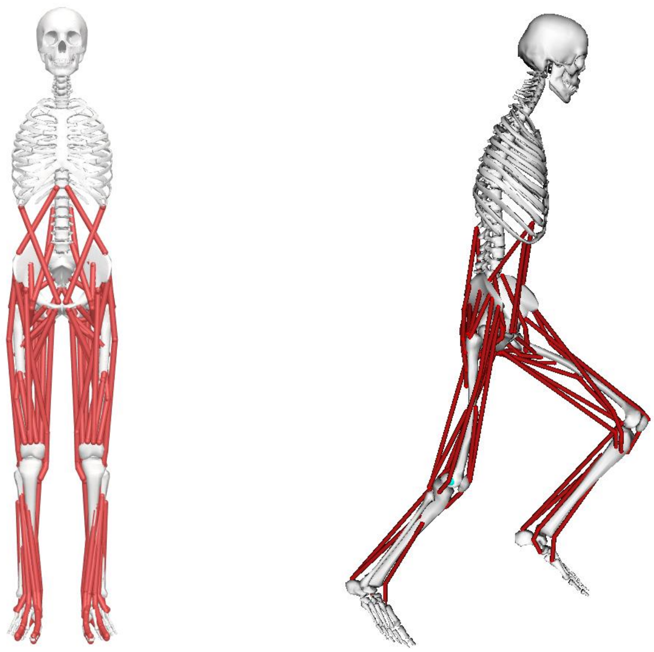


Figure 19 Musculoskeletal model in OpenSim

Many studies have used OpenSim to investigate the biomechanics of running. For example, a study by Delp et al. used OpenSim to simulate the muscle forces and joint dynamics during running, and found that the ankle plantarflexors and knee extensors are the primary contributors to running propulsion (Delp et al., 2007). The study also found that the hip extensors play a role in controlling the forward motion of the body during running. Another study by Hamner et al. used OpenSim to investigate the effect of muscle properties on running mechanics (Hamner & Delp, 2013). The study found that muscle properties such as maximum isometric force and optimal fiber length have a significant impact on running performance, and that these properties may be modulated through training or other interventions. A more recent study (Alexander et al., 2021) used OpenSim to investigate the effect of running speed on muscle activation patterns during running. The study found that faster running speeds are associated with increased muscle activation in the ankle dorsiflexors, knee extensors, and hip flexors. The study also found that slower running speeds are associated with greater activation of the ankle plantarflexors and knee flexors.

AnyBody is another software package for musculoskeletal multibody modeling as shown in Figure 20. It was developed by researchers at Aalborg University in Denmark and is designed to provide a more user-friendly interface for building and simulating musculoskeletal models. AnyBody uses a combination of inverse dynamics and optimization techniques to simulate the motion and forces of the musculoskeletal system during running. Several studies have used AnyBody to investigate the biomechanics of running. For instance, a study (Schache et al., 2014) used AnyBody to simulate the muscle forces and joint torques during running and found that the hip flexors and knee extensors play important roles in running mechanics. The study also found that the hip extensors are important for generating horizontal propulsion during running.

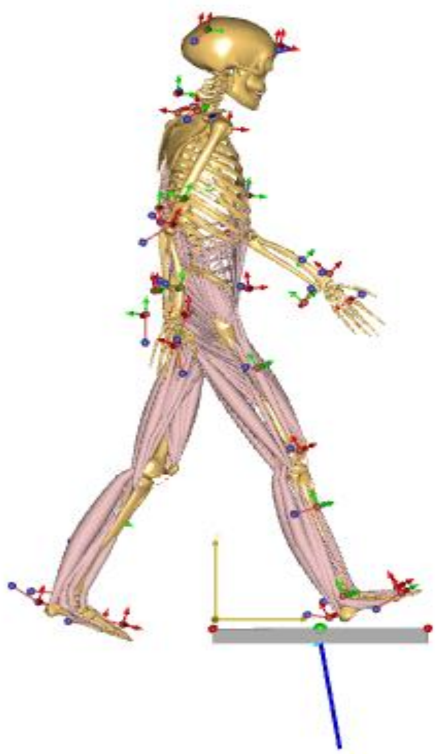


Figure 20 Musculoskeletal model in Anybody

Both OpenSim and AnyBody have been used extensively in studies investigating the biomechanics of running. While the two software packages have some similarities in terms of the types of models that can be created and the analyses that can be performed, there are also some important differences between the two. One key difference between OpenSim and AnyBody is the way in which they simulate muscle forces. OpenSim uses a Hill-type muscle model, which is a relatively simple model that assumes that muscle force is proportional to muscle activation and muscle length. AnyBody, on the other hand, uses a more complex muscle model that takes into account factors such as muscle architecture, force-length-velocity properties, and activation dynamics. This more complex model may provide a more accurate representation of muscle function during running. Another difference between OpenSim and AnyBody is the way in which they handle the optimization problem that arises when simulating the motion and forces of the musculoskeletal system. OpenSim uses a sequential optimization approach, in which the motion of the musculoskeletal system is first predicted based on prescribed joint angles, and then the muscle forces are optimized to match the predicted motion. AnyBody uses a simultaneous optimization approach, in which the motion and muscle forces are optimized together. The simultaneous approach may provide a more accurate representation of the dynamic interactions between the different components of the musculoskeletal system during running.

Overall, both OpenSim and AnyBody have been used successfully in studies investigating the biomechanics of running. The choice of software package may depend on the specific research question being addressed, as well as the expertise and resources available to the researcher.

CHAPTER 3 OVERVIEW

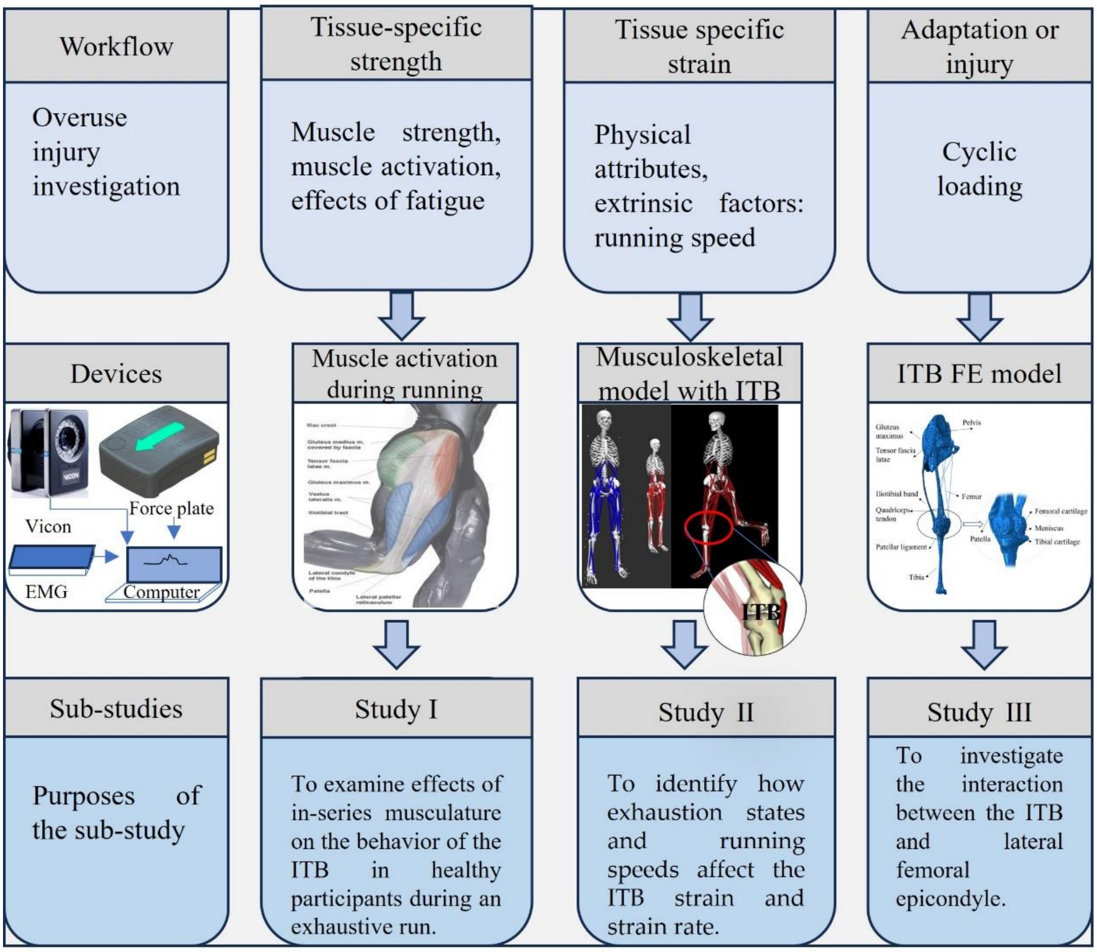
3.1. Overview of the sub-studies

In the first study, we examined ITB-related muscle activities during a prolonged running. The underlying rationale of the study is that the behaviours of ITB might be altered due to the change in the muscle activities during the whole running. Throughout the entire running duration, EMG signals from the relevant muscles were recorded and analysed. The effects of the alteration in the muscle activities on the biomechanics of ITB could provide insights into the implications of ITBS.

In the second study, we utilized a musculoskeletal model incorporated ITB to examine the changes in ITB strain and strain rate before and after an exhaustive run. We aimed to investigate whether there is a higher increase in ITB strain rate after an exhaustive run as the high ITB strain rate was proposed to be primary factor to the development of ITBS. Additionally, we examined the speed effects on ITB biomechanics. Both the exhaustive running state and running speed are external factors.

In the third study, we employed a ITB finite element model to investigate the interactions between ITB and the lateral femoral epicondyle. There was proposed a pain zone around the knee flexion angle of 30° that would occur to ITBS runners. Compression force applied to the lateral femoral epicondyle by the ITB can trigger pain. Thereby, the established ITB finite element model was adopted to simulate the compression force.

3.2 Method overview



3.2.1 Motion capture system

Experimental equipment

VICON is a camera-based motion capture system that uses specialized cameras and reflective markers to track the movement of objects or people in 3D space. The system consists of several components, including high-speed cameras, specialized software, and reflective markers. The VICON system uses high-speed cameras to capture the movement of reflective markers in real-time. As shown in Figure 21, these cameras are typically placed around the space where the movement is being captured, and they can capture up to several hundred frames per second. Reflective markers are placed on the objects or people being tracked. These markers reflect the light emitted by the cameras, allowing the system to track their movement in 3D space. VICON offers a range of different marker sizes and shapes to suit different applications. The cameras capture the movement of the reflective markers in real-time, and the software processes the data to create a 3D model of the movement, which has high accuracy and precision.

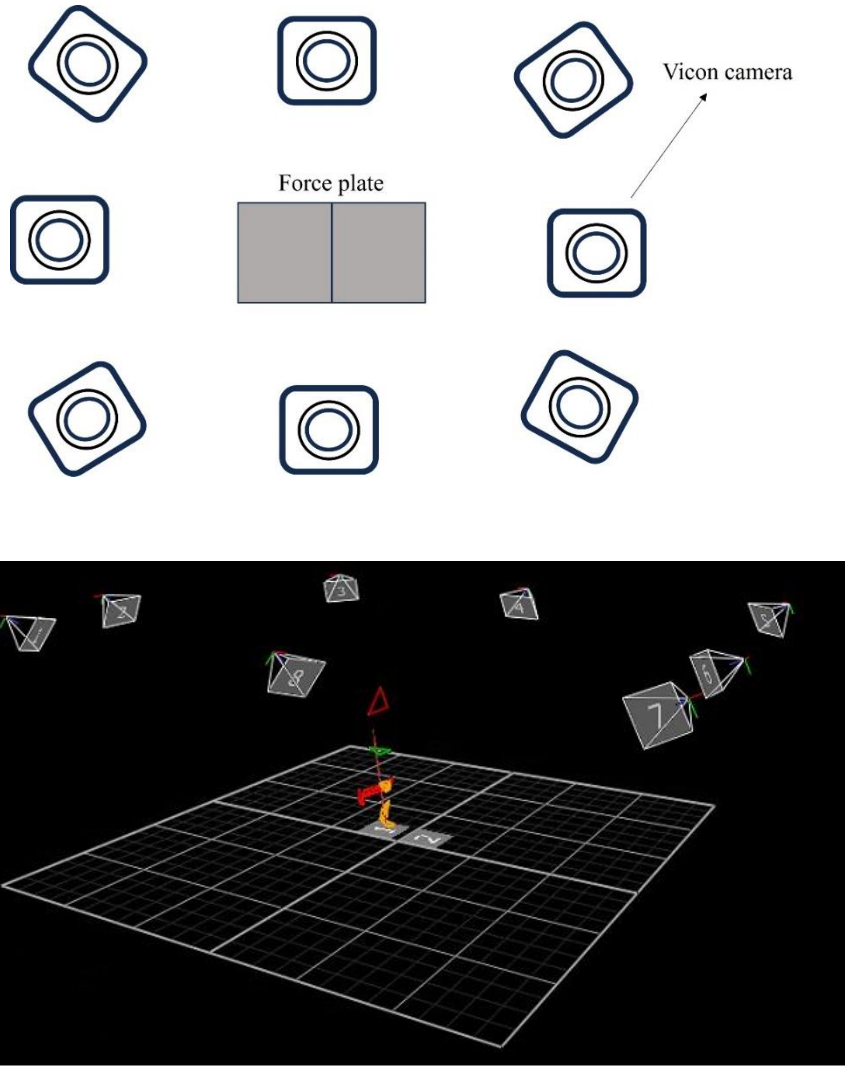


Figure 21 Vicon system with force plates

The workflow of the VICON includes calibration of the whole system, marker attachment, motion capture, and data processing. VICON provides calibration tools to ensure that the cameras are accurately tracking the movement of the reflective markers. These tools include specialized calibration markers and a calibration wand. After calibration, the VICON motion capture system works by using high-speed cameras to capture the movement of reflective markers that have been attached to the bony landmarks in 3D space. The VICON software then uses this information to create a 3D model of the movement. The resulting data can be analyzed using motion analysis tools and real-time tracking tools. The data can also be exported to other software packages, such as OpenSim, for further processing.

Lower limb coordinate system

In biomechanics, a coordinate system is a set of reference axes that are used to describe the position and movement of body segments during physical activities. The use of a standardized coordinate system is essential for accurate and reproducible measurements of joint angles and other biomechanical variables.

The lower limb coordinate system (Figure 22) used in biomechanics relies on anatomical landmarks and serves as a reference for measuring joint angles during various activities such as walking, running, and jumping. The widely adopted convention for the lower limb coordinate system is known as the Cardan/Euler angle convention. This convention employs three orthogonal rotational axes to define the position and motion of the lower limb. These axes are named the craniocaudal axis, mediolateral axis, and anteroposterior axis. The flexion-extension axis corresponds to movement in the sagittal plane, while the abduction-adduction axis pertains to movement in the frontal plane. Lastly, the internal-external rotation axis relates to movement occurring in the transverse plane. By employing this mathematical system, researchers can accurately describe and analyze the intricate movements of the lower limb.

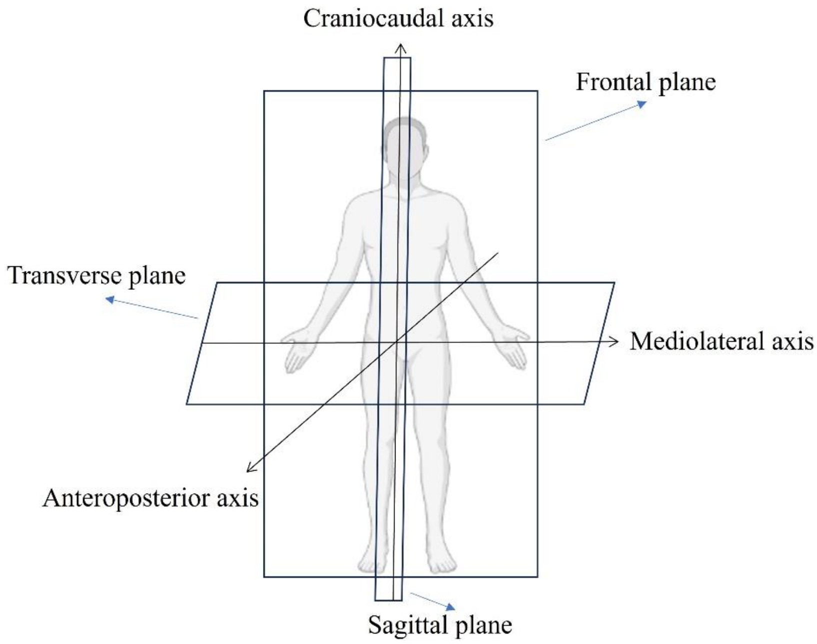


Figure 22 Illustration of coordinate system

When it comes to measuring joint angles using the lower limb coordinate system, the axes are aligned with the anatomical axes of the specific joint under consideration. For instance, when measuring knee flexion-extension, the flexion-extension axis is aligned with the longitudinal axis of both the femur and the tibia. By aligning the axes accordingly, the joint angle can be calculated based on the orientation of the segment relative to the fixed axes of the coordinate system.

The local coordinate system of the pelvis

The local coordinates of the pelvis establish a coordinate system that remains fixed to the pelvis itself. This system is utilized to describe the orientation and motion of the pelvis in relation to the rest of the body during human movement. The accurate measurement of hip joint angles and other biomechanical variables heavily relies on employing the local coordinates of the pelvis. Typically, the local coordinate system of the pelvis incorporates three orthogonal axes that align with the anatomical axes of the pelvis. These axes are defined as follows:

1). Anterior-Posterior Axis (x-axis): It spans from the anterior superior iliac spine to the posterior superior iliac spine. This axis represents the sagittal plane of the body and corresponds to the forward-backward movement.

2). Superior-Inferior Axis (y-axis): It runs from the center of the sacrum to the midpoint of the line connecting the two ASISs. This axis represents the coronal plane of the body and aligns with the up-down movement.

3). Medial-Lateral Axis (z-axis): It is perpendicular to the x and y axes and passes through the center of the pelvis. This axis represents the transverse plane of the body and corresponds to rotational movement.

By establishing the local coordinate system of the pelvis in this manner, orientation and movement of the pelvis can be precisely quantified during various activities, enabling accurate analysis and assessment of hip joint angles and related biomechanical parameters.

The local coordinate system of the femur

Local coordinates of the femur refer to a coordinate system that is fixed to the femur, which can be used to describe the orientation and movement of the femur relative to the pelvis and the rest of the body during human movement. The use of local coordinates of the femur is critical for the accurate measurement of hip joint angles. The local coordinate system of the femur typically uses three orthogonal axes that are aligned with the anatomical axes of the femur. The axes are defined as follows: The proximal-distal axis (y-axis) is oriented from the center of the femoral head to the center of the femoral condyles. It represents the sagittal plane of the body and is aligned with the direction of forward-backward movement. The medial-lateral axis (z-axis) is oriented from the midpoint of the line connecting the two femoral epicondyles to the midpoint of the femoral neck. It represents the coronal plane of the body and is aligned with the direction of medial-lateral movement. The anterior-posterior axis (x-axis) is oriented perpendicular to the x and y axes and passes through the center of the femur. It represents the transverse plane of the body and is aligned with the direction of rotational movement.

The local coordinate system of the tibia

Local coordinates of the tibia provide a fixed coordinate system that enables the accurate measurement of knee joint angles and other biomechanical variables during human movement. This coordinate system is aligned with the anatomical axes of the tibia and serves to describe the orientation and movement of the tibia relative to the femur and the rest of the body.

The local coordinate system of the tibia consists of three orthogonal axes. The proximal-distal axis (y-axis) extends from the center of the tibial plateau to the center of the tibial plafond, representing the sagittal plane and forward-backward movement. Perpendicular to the x-axis, the medial-lateral axis (z-axis) passes through the center of the tibial plateau, representing the coronal plane and medial-lateral movement. The anterior-posterior axis (x-axis), perpendicular to both the x and y axes, passes through the center of the tibia, representing the transverse plane and rotational movement. For instance, knee joint angles can be precisely measured. For example, when assessing knee flexion-extension, the x-axis aligns with the longitudinal axis of the tibia. The angle is determined by evaluating the orientation of the tibia in relation to the fixed axes of the femur coordinate system.

The local coordinate system of the foot

Coordinates of the foot are a coordinate system that is fixed to the foot and is used to describe the orientation and movement of the foot relative to the lower leg and the rest of the body during human movement. The use of local coordinates of the foot is critical for the accurate measurement of ankle and foot joint angles and other biomechanical variables. The local coordinate system of the foot typically uses three orthogonal axes that are aligned with the anatomical axes of the foot. The axes are defined as follows: The longitudinal axis (x-axis) is oriented from the center of the heel to the center of the forefoot. It represents the sagittal plane of the body and is aligned with the direction of forward-backward movement. The transverse axis (y-axis) is oriented perpendicular to the x-axis and passes through the center of the foot. It represents the coronal plane of the body and is aligned with the direction of medial-lateral movement. The sagittal axis (z-axis) is oriented perpendicular to the x and y axes and passes through the center of the foot. It represents the transverse plane of the body and is aligned with the direction of rotational movement.

Ankle and foot joint angles can be measured using the local coordinates of the foot, of which the axes are aligned with the anatomical axes of the ankle joint. For example, when measuring ankle dorsiflexion-plantarflexion, the x-axis is aligned with the longitudinal axis of the foot, and the angle is calculated based on the orientation of the foot relative to the fixed axes of the lower leg coordinate system.

3.2.2 Surface electromyography (EMG)

Muscle fibres excite via neural control in muscle physiology. Muscle fibre membrane stays at a resting potential when an ionic equilibrium forms between the inner and outer sides of the muscle cell. The resting potential states at about -80 to -90 mv when the muscle is not contracted. Activation triggered by the central nervous system or reflexes would result in the transmission of excitation along the nerve. Subsequently, following the release of transmitter substances at the motor end phase, the endplate potential is generated. It is presumed that any healthy muscle engaged in any type of muscle contraction would be associated with the mechanism.

The signal of EMG is a transforming form of the action potentials. Bipolar electrode produces a potential difference along the surface of the muscle fibre. EMG measurements would utilize a differential amplification for the kinesiological signals. The contraction process and the resultant force output of the activating muscles are the two primary control strategies. For the raw signals, soft tissues would apply a low pass filter effect. Consequently, surface EMG is not the original firing and amplitudes characters of the muscles. EMG signals directly reflects the characteristics of the measured muscle.

Raw EMG signals of the three bursts of the Gmed during running is given in Figure 23. When the muscle is unfired, a noise-free baseline for the EMG signal could be identified. The baseline noise may stem from factors such as the EMG amplifier, environmental conditions, and detection settings (Liu & Liu, 2016). The baseline quality for the EMG signal is an important checkpoint for the interpretation of the muscle activities. For optimal amplifier performance and appropriate skin preparation, the average baseline noise should ideally be below 3-5 microvolts.

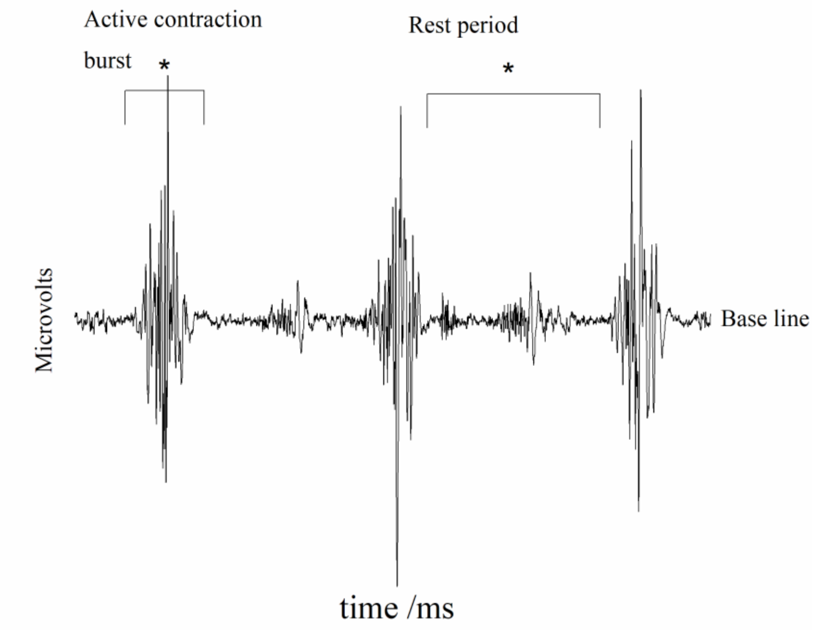


Figure 23 Raw EMG recording signals of 3 bursts of the Gmed

The EMG activity patterns present noise signals due to no action potentials for healthy relaxing muscles. Since the raw EMG spike exhibits a random shape, interpreting a single burst of raw recording may not yield an exact shape. A strong superposition spike would be produced if two or more motor units near the electrodes activate at the same time, which could be eliminated by applying a smooth algorithm (Kakehata et al., 2021). The raw EMG signals have a frequency range spanning from 6 Hz to 500 Hz, with the highest power concentrated between 20 Hz and 150 Hz.

The EMG signal might be affected by external factors, which change the shape and characteristics of the activities. Soft tissues could exhibit a direct influence on the EMG signal. The attachment location in the muscle belly of the sensors could also apply some external pressure during movement. The phenomenon ‘cross talk’ that indicates a significant amount of EMG signal from the neighbouring muscle should not exceed over 15% of the overall contents. It is important to note that removing the hair and cleaning the skin on the muscle before the EMG measuring are the essential procedure.

Anatomical landmarks for the sensor location

Surface EMG are usually used in kinesiological studies due to the non-invasive character (Besomi et al., 2020). The main deficit of the surface EMG is that deeper muscles below surface muscles or bones are not applicable. Electrodes of sensors should be applied along the muscle fibre direction and attach to the muscle belly for the best signals. The attachment site is based on the recommendations from the SENIAM (Hermens et al., 1999).

The DelSys Trigno wireless EMG system (DelSys, Boston, MA) was utilized to record the activities of the TFL, Gmax, Gmed, VL, and BF muscles at a sampling rate of 2000Hz. As stated by SENIAM guidelines, the electrode placement direction was parallel to the orientation of the muscle fibres (Hermens et al., 1999). The EMG electrodes for the TFL were positioned on the proximal 1/6 section of the line extending from the superior anterior iliac spine to the lateral femoral condyle. For the Gmax, the EMG electrodes were attached at the midpoint between the greater trochanter and the sacral vertebrae. The EMG electrodes for the Gmed were placed at the midpoint between the iliac crest and the trochanter. The VL EMG electrodes were positioned at the 2/3 point along the line from the superior anterior iliac spine to the lateral side of the patella. Finally, the EMG electrodes for the BF were positioned midway between the ischial tuberosity and the lateral epicondyle of the tibia. Detailed information about these muscles can be found in Table 6.

Table 6 The attaching position of the surface EMG

Muscle name	Attaching location	orientation	Muscle function
TFL	The electrodes should be placed at the proximal 1/6 of the line between the anterior spina iliac and the lateral femoral condyle	In the direction from the anterior spina iliac to the lateral femoral condyle	Adduction, internal rotation, and flexion of the hip. Tension of the fascia latae
Gmax	The electrodes should be placed at the middle line between the greater trochanter and the sacral vertebrae	In the direction from the posterior superior iliac spine to the middle of the thigh posterior	The Gmax assists in the adduction of the hip joint and stabilization of the knee joint
Gmed	The electrodes should be attached at the middle of the line between the greater trochanter and the crista iliac	In the direction from the crista iliac to the greater trochanter	Abduct the hip joint; The anterior medially rotate the hip joint and also might be involved in its flexion; the posterior laterally rotates and might be involved in its extension
BF	The electrodes should be placed in the middle between the ischial tuberosity and the lateral epicondyle of the tibia	In the direction from the ischial tuberosity to the lateral epicondyle of the tibia	External rotation and flexion of the knee joint. The long head involves in the extension and lateral rotation of the hip joint
VL	Electrodes should be placed at 2/3 of the line between anterior spina iliac superior and lateral side of the patella	In the direction of the muscle fibers	Extension of the knee joint

Before the formal running test, all the EMG were attached to the participants firmly to avoid dropping during running. Participants were required to familiar with the lab-provided running shoes before running. Dynamic trials were collected during the running. EMG data at 0 min, 10 min, 20 min. and 30 min were extracted to analyse the muscle activities. Ten intact strides for each time point would be used to calculate the average values.

EMG signal processing

Timing and amplitude characteristics could be derived from raw EMG signals that contained important for muscle activities, which would provide more reliability of findings (Distefano et al., 2009; Macadam & Feser, 2019). The initial processing step involves full-wave rectification, wherein all negative amplitudes are converted to positive values. The recruited motor units change within the diameter of the motor units and the action potentials superpose arbitrarily. Hence, the raw EMG spike cannot be replicated exactly in its shape a second time. To address the problem, a smoothing algorithm that outlines the mean signal development and cut off the steep spikes would be applied to the rectified EMG signals (Soljanik et al., 2012). A low pass second Butterworth filter with a cutoff frequency of 10 Hz would be performed to obtain linear envelopes (Elsais et al., 2020). Figure 24 shows the raw EMG signals with the envelope of one stride.

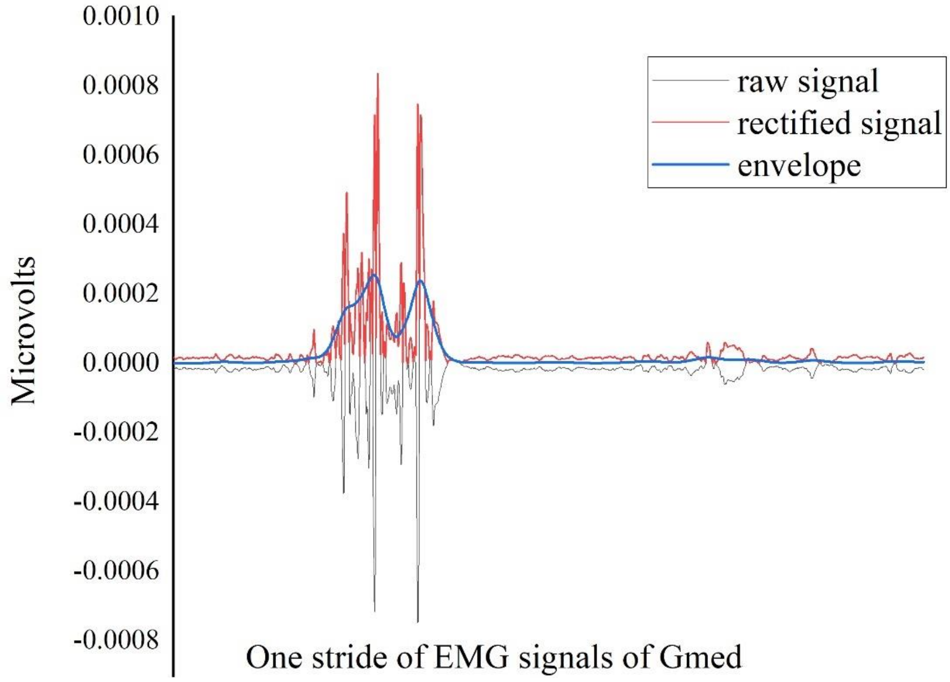


Figure 24 Illustration of the raw EMG signal processing

3.2.3 Musculoskeletal multibody model for ITB

OpenSim is a cutting-edge software platform that revolutionizes the study and analysis of human movement and biomechanics. It provides a comprehensive set of tools to create detailed musculoskeletal models and simulate the mechanics and dynamics of the human body. The construction of sophisticated musculoskeletal models includes the bones, joints, muscles, and other soft tissues of the human body. The models can be customized and scaled to fit specific research or clinical needs, enabling to delve into the intricacies of human movement. OpenSim offers a wide range of analysis tools that facilitate in-depth investigations of muscle forces, joint kinematics, and other biomechanical variables. By applying external forces and loads, researchers can replicate real-life scenarios and gain insights into joint loading, muscle activity, and overall movement patterns.

The workflow in OpenSim typically involves several steps, allowing to create musculoskeletal models, analyze movement data, and simulate biomechanical behaviour (Delp et al., 2007). The first step is to create a musculoskeletal model that accurately represents the anatomy and mechanics of the human body. This involves importing anatomical data, such as bone geometries. The model can be customized by adding joint constraints, muscle paths, and other relative details. The next step is scaling and inverse Kinematics: once the model is created, it needs to be scaled to match the size and proportions of the individual being studied. Scaling involves adjusting the model's dimensions to accurately represent the subject's body. Subsequently, inverse kinematics is employed to estimate joint angles and positions using motion capture data. Then, inverse dynamics: inverse dynamics analysis is performed to estimate the forces and torques acting on the joints of the musculoskeletal model. This is done by combining the motion capture data with the model's geometry, mass properties, and muscle representations. Inverse dynamics provides insights into joint loading, muscle forces, and overall biomechanical behaviour during movement. The final step would be dynamic simulations to analyse muscle function and performance. Muscle activation analysis provides information about muscle activation patterns throughout a movement. Muscle-tendon lengths, moment arms, and forces can also be computed. Dynamic simulations involve applying external forces and loads to the musculoskeletal model to predict how it will behave under different conditions. Dynamic simulations provide valuable insights into joint kinetics, muscle forces, and overall movement dynamics. Figure 25 provides the general workflow of the musculoskeletal model.

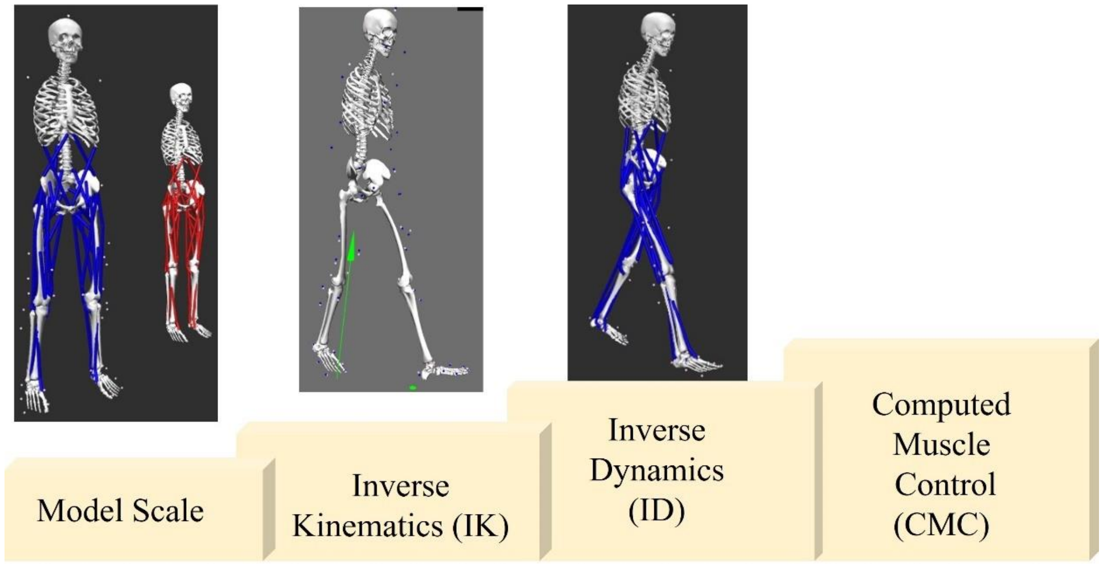


Figure 25 Workflow of musculoskeletal model in OpenSim

The ITB model was developed by adapting the model gait2392_simbody in OpenSim (Version 3.3). The model consists of major bones that are essential for running. These bones include the head, trunk, pelvis, femur, tibia, and foot segments. Each bone is represented as a rigid body with specific anatomical properties and segmental masses. The gait2392 model also incorporates a comprehensive set of muscles that actuate the joints of the lower limb. Each muscle is represented as a musculotendon unit, considering its optimal fibre length and maximum isometric force. The knee joint was modified to be a three-degree rotational freedom joint from its original single-degree joint (Day & Gillette, 2019). The ITB was assumed to follow the same anatomical pathway as the TFL (Hamill et al., 2008) from the iliac crest to Gerdy's tubercle. The ITB was defined to be passive elastic structure (Shinji Tomiyama et al., 2015) with five points to simulate its attachments including ilium, lateral intermuscular septum, distal femur, and anterolateral proximal tibia as shown in Figure 26.

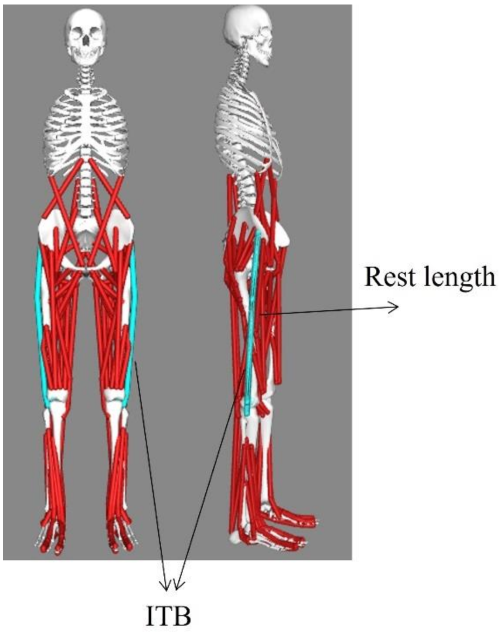


Figure 26 Musculoskeletal model with ITB

By incorporating ITB into the musculoskeletal model, its function, role in movement, and potential contributions to various pathologies or movement disorders such as ITBS can be discussed.

3.2.4 Finite element model

Geometry reconstruction

The MRI data of the right leg was taken using a whole-body 3.0T Siemens Prisma scanner (Seimens, Erlangen, Germany) with the configuration of T1 sequence, 1-mm slice interval, and a resolution of 0.625 mm pixel size in a neutral position. The ankle was fixed to a stable neutral position with a custom-made ankle-foot orthosis during the whole MRI taking. After collected the MRI data, geometry reconstruction could be performed from MRI images using software Mimics 15.0 (Materialise, Leuven, Belgium), which is a process that involves extracting and converting imaging data into three-dimensional models as shown in Figure 27. In Mimics, thresholding allows to segment regions based on intensity values. Thresholding for a specific mask sets a range of intensity values that correspond to the desired region, effectively separating it from the background and other structures. Based on the clear boundaries, region growing algorithms in Mimics that automatically segment regions based on seed points expands the region by iteratively including neighboring voxels, which is suitable for segmenting structures with homogeneous intensity characteristics, such as bones. As for low quality images with vague boundaries, manual segmentation was used to draw contours around the desired regions in each image slice. According to the complete mask boundaries, 3D surface models could be generated from the segmented regions. Surface extraction converts the segmented image data into a mesh representation, where each vertex represents a point in 3D space, and the connectivity between vertices forms triangular or polygonal faces. This mesh forms the basis for creating the 3D geometry. However, the generated surface model may undergo further refinement steps to improve its quality and accuracy due to the noise and artifacts. The steps might include smoothing the surface, removing any nonessential details or noise, and filling any holes or gaps in the model. The refinement of the 3D geometry was performed in 3-Matic 8.0 software (Materialise, Leuven, Belgium). Furtherly, the surface of the 3D geometry was refined in Geomagic Studio 12 (Geomagic, Morrisville, U.S.) to obtain solid geometry as shown in Figure 28. The capabilities of this software enable converting raw point cloud data into accurate, high-quality 3D models that is suitable for finite element analysis. Several processes in Geomagic Studio include point cloud cleanup, surface reconstruction, and mesh editing. Before proceeding with solid geometry and surface reconstruction, it's essential to clean up the point cloud data, which removes outliers, noise, and unwanted artifacts from the point cloud, resulting in a cleaner and more accurate dataset. Then advanced surface reconstruction algorithms were employed to convert the point cloud data into a continuous, smooth surface model. The solid geometries for bones and muscles were obtained finally.

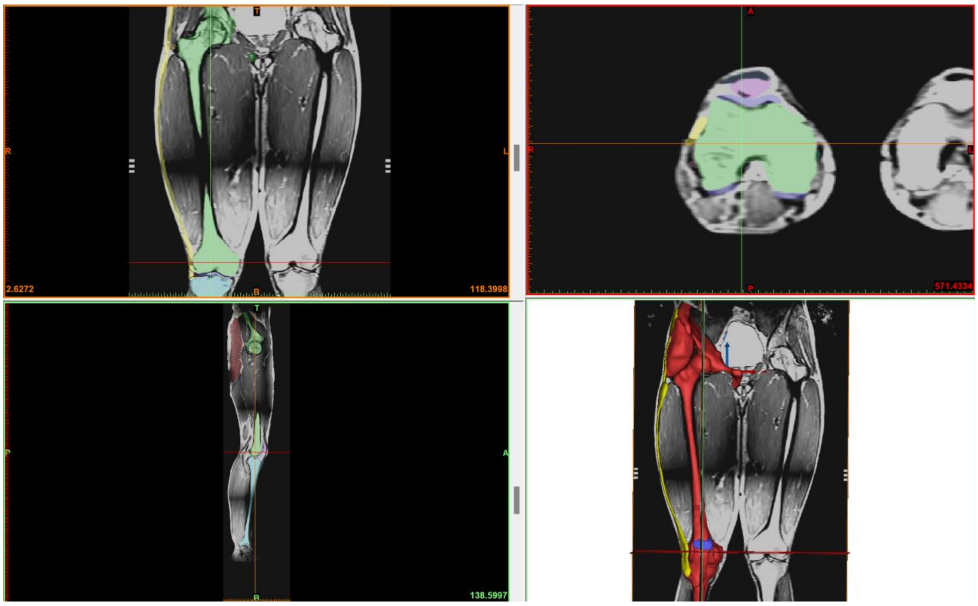


Figure 27 MRI data of the lower limb Mimics

Figure 28 The three-dimensional refined model in 3-Matic

After bone and soft tissues (meniscus, ligaments, and muscles) geometries were constructed, ligaments represented using trusses were constructed to connect bony structures. Muscles except for TFL and Gmax were represented using connects in the finite element model. The ankle joint forces were implemented to the tibial reference point (RP). In the model, constraints were applied to the cartilage-tibia bone interface, cartilage-femur bone interface, meniscus, and ligament insertion points in the femur and tibia for all six degrees of freedom (Bolcos et al., 2018).

Mesh and Materials

The finite element model of ITB was established in Abaqus 6.14 (Simulia, Dassault Systemes, Vélizy-Villacoublay, France). After the solid parts were imported and assembled in Abaqus, mesh generation was performed for all the parts. Bones were assigned to be linear tetrahedral element (C3D4) and ligaments were assigned to be two-node linear three-dimensional elements (T3D2).

CHAPTER 4 Effects of Alteration in Muscle Activation on the Iliotibial Band during an Exhaustive Run

4.1 Summary of the study

This study aimed to investigate the impact of in-series musculature on the behavior of ITB in healthy participants during an exhaustive run.

A total of 25 healthy participants (15 males, 10 females) engaged in a 30-minute exhaustive run at a self-selected speed while wearing laboratory-provided footwear. Surface EMG signals were obtained for the muscle activities of ITB-related muscles during an exhaustive run, including TFL, Gmax, Gmed, BF, and VL. The raw EMG signals were initially filtered to remove noise and unwanted artifacts followed by rectification. The EMG signals were subjected to full wave rectification, followed by the application of a second-order Butterworth bandpass filter with a frequency range of 10 Hz to 500 Hz. Subsequently, linear envelopes of the rectified EMG signals were attainable by applying a second-order Butterworth low-pass filter with a cutoff frequency of 6 Hz.

A comparison was made between the maximum amplitudes of muscle activity at three stages during the exhaustive run. Significant reductions were noted in the maximum amplitudes of the TFL, Gmax, Gmed, and BF muscles during the middle and final stages when compared to the initial stage. However, the onset and offset of muscle activation remained consistent throughout the running session.

The behaviour of ITB in healthy individuals may be influenced by the activities of in-series musculature during an exhaustive run. The cyclic and repetitive movements performed by runners can result in higher compression forces exerted by ITB on the lateral femoral epicondyle, contributing to knee joint stability. These findings enhance our understanding of the behavior of the healthy ITB and its role in knee joint stability during prolonged running.

4.2 Introduction

Among lower extremity injuries, ITBS is a commonly occurring overuse injury, accounting for a significant 22.2% incidence rate (Hamstra-Wright et al., 2020). It accounts for a significant proportion of running-related injuries (Kakouris et al., 2021; Taunton et al., 2002). Individuals experiencing ITBS commonly report pain that emerges after running a few kilometers. As running persists, the pain in the lateral femoral epicondyle becomes more intense (Noble, 1980). In severe cases, the pain often hampers runners from participating in sports activities either temporarily or permanently (Friede et al., 2021). Excessive compression forces on the lateral femoral epicondyle, exerted by ITB, are considered a primary cause of ITBS (John Fairclough et al., 2006). Running biomechanics have been suggested to directly impact knee compression forces through the tensioning of ITB (Hutchinson et al., 2022).

ITB is a segment of the superficial fascia that becomes thicker and primarily attaches to the lateral tibial condyle (Godin et al., 2017). In the proximal position, ITB receives tension from TFL, Gmax (Flato et al., 2017), and gluteal aponeurosis over the Gmed. The contraction of the VL and BF muscles causes distortion of ITB at the distal position (Besomi et al., 2021; Birnbaum et al., 2004). The coordination of ITB with the BF and VL is crucial for stabilizing the knee and hip joints (Foch et al., 2020). The ITB has a complete insertion into the TFL and partial attachment to the Gmax, allowing it to receive tension from these muscles. This tension plays a role in providing anterolateral rotational stability to the knee joint. Additionally, a previous study (John Fairclough et al., 2006) used magnetic resonance imaging, discovering that the tension in the ITB, induced by nearby muscles, can result in compression on the lateral femoral epicondyle. As a result, the compression forces exerted on the lateral femoral epicondyle by the tension of the ITB contribute to stabilizing knee stability (Hutchinson et al., 2022). The ITB has been suggested to be one of the most significant stabilizers of the knee joint, particularly during high knee flexion angles (Kaplan & Jazrawi, 2018).

Changes in the transmission of force among interconnected structures may potentially contribute to overuse injuries (Wilke et al., 2019). The force transmission in ITB is primarily influenced by the activation of the relevant muscles. The activation patterns of these muscles may undergo alterations during exhaustive running (Brown et al., 2016; Cowley & Gates, 2017). Moreover, there is evidence suggesting that passive loading in the tissues escalates during exhaustive running, thereby heightening the risk of injury (Debenham et al., 2016). Surface EMG Surface is commonly used to measure muscle activity (Gazendam & Hof, 2007; Schache et al., 2014), with average EMG amplitudes providing a non-invasive measure of muscle activation levels over a given period (Hsu et al., 2020). Previous studies have indirectly associated EMG amplitudes with injury risk, showing that increased amplitudes are related to increased musculotendon stiffness in major lower limb muscles (Hunter et al., 2014). Previous studies have indirectly associated EMG amplitudes with injury risk, showing that increased amplitudes are related to increased musculotendon stiffness in major lower limb muscles (Kyrolainen et al., 2005). Souza and Powers (2009) used surface EMG to explore the relationship between hip muscles and knee pain. Rabita et al. (2013) analyzed analyze the amplitudes of the BF during exhaustive running, finding a significant decrease in pre-activation phase. The activities of the Gmed and TFL have also been investigated during exhaustive running, with both muscles showing a significant decrease in amplitude (Brown et al., 2019). The timing of muscle activation (onset and offset) could verify the duration of muscle firing. However, accurately identifying the timing of muscle activation and deactivation can be challenging due to artifacts and noise in surface EMG signals during running. The Teager-Kaiser energy operator (TKEO) can aid in addressing this issue by offering more dependable results (Li et al., 2007; Solnik et al., 2010). This algorithm detects the first point in which the EMG signal exceeds a baseline as the onset of muscle activation, and the first point below a certain threshold as the offset. By using TKEO, the current study aims to obtain temporal parameters of muscle activation.

Runners engage in cyclic and repetitive movements (Gaudreault et al., 2018). Pain caused by ITBS typically arises after a period of running (Baker et al., 2011) rather than at the beginning. It is believed that the behavior of ITB may change, leading to excessive and abnormal compression forces between ITB and the lateral femoral epicondyle. Previous studies have explored the timing and amplitudes of ITB-related muscle activity in different phases of exhaustive running. Consequently, this study postulates that muscle activities will undergo alterations during exhaustive running, leading to changes in the behavior of the ITB. Furthermore, it is also conjectured that the heightened repetitive and cyclic running activities would result in increased compression forces exerted by the ITB on the lateral femoral epicondyle.

4.3 Methods

4.3.1 Participant information

The sample size for this study was determined using Gpower (G*power 3.0.10, Universität Düsseldorf, Germany) with a significance level of 0.05. Twenty-five healthy participants, comprising 15 males and 10 females, were enlisted from the Hong Kong Polytechnic University. All participants fell within the age range of 20 to 30 years old, with a BMI varying from 19 to 24. To ensure data integrity, individuals with musculoskeletal injuries within the past 12 months were excluded from the study. Furthermore, the recruited participants engaged in running activities of at least 10 miles per week. All participants provided informed written consent prior to the experiment. The Human Subjects Ethics Sub-Committee of the Hong Kong Polytechnic University (Number: HSEARS20150121003) has approved this study.

4.3.2 Experimental procedure

In this study, the DelSys Trigno wireless EMG system (DelSys, Boston, MA) was used to collect muscle activity data for the TFL, Gmax, Gmed, BF, and VL muscles. The sampling rate was set at 2000 Hz. Before placing the electrodes, the skin was cleaned and shaved, and bipolar differential surface electrodes (Ag/AgCL) were affixed to the muscle bellies. The placement of the electrodes followed the guidelines provided by the 'surface EMG for non-invasive assessment of muscles' (SENIAM), which is a European project that offers standard recommendations for electrode placement (Hermens et al., 1999). The specific electrode positions for each muscle were as follows: the TFL electrode was positioned on the proximal 1/6 of the line from the anterior superior iliac spine to the lateral femoral condyle. For Gmax, the electrode was attached at the 1/2 point between the great trochanter and the sacral vertebrae, while for Gmed, it was placed at the 1/2 point between the iliac crest and the trochanter. The VL electrode was positioned at the 2/3 point on the line from the anterior superior iliac spine to the lateral side of the patella. Similarly, the BF electrode was placed at the 1/2 point between the ischial tuberosity and the lateral epicondyle of the tibia. Additionally, it was assumed that the BF had similar EMG activation patterns in both the long and short heads based on previous research (Chen et al., 2018). To ensure consistency and accuracy in electrode placement, a single research assistant applied all the electrodes for all participants, minimizing potential variations in muscle activation recordings.

All participants were attired in neutral sports footwear (ARHQ025-4, Li-Ning Inc., Beijing, China) for the treadmill running test. Before the formal test, a 5-minute warm-up was conducted, followed by an additional 5 minutes for participants to familiarize themselves with the treadmill. Afterward, participants were instructed to run on the treadmill at their self-selected speeds and engage in a 30-minute exhaustive run. Throughout this period, they were required to maintain their chosen running speeds consistently (Besomi et al., 2020). During the exhaustive run, the running trial was continuous at the fixed speed chosen by each participant. Raw surface EMG signals of the five muscles (TFL, Gmax, Gmed, BF, and VL) were recorded for 1 minute at three different time points: the 1-minute (initial stage), the 15-minute (mid stage), and the 30-minute (end stage) of the 30-minute exhaustive running trial.

4.3.3 Data analysis

To process the raw EMG signals, a series of filtering techniques were implemented.

Initially, a fourth-order Butterworth high-pass filter with a cutoff frequency of 20 Hz was utilized to eliminate electrode artifacts (Baker et al., 2018; Gazendam & Hof, 2007). Subsequently, the TKEO algorithm was employed (Li et al., 2007; Solnik et al., 2010). To further enhance signal quality, an additional high-pass filter with cutoff frequencies ranging from 10 to 20 Hz was employed to eliminate any remaining artifacts. Following full wave rectification, the EMG signals underwent processing with a second-order bandpass Butterworth filter, with cutoff frequencies ranging from 10 Hz to 500 Hz. Linear envelopes were created by utilizing a low-pass second-order Butterworth filter with a cutoff frequency of 6 Hz. The EMG profile for each muscle was depicted by averaging the linear envelopes of ten successive cycles (Hug, 2011). For ease of comparison, the EMG data were normalized to the maximum amplitude recorded during the initial running phase.

The temporal parameters of surface EMG signals include determining the onset and offset of EMG bursts, which are important for detecting the timing of muscle activation. To determine these timing parameters, the detection threshold was set at 10% of the local peak of the maximum amplitude. The onset and offset points were identified at the intersection of the individual envelope and the established threshold (Hug, 2011). To derive the onset and offset timing, the average of ten consecutive cycles was calculated and then expressed as a percentage of the gait cycle.

We used IBM SPSS Version 19.0 (IBM Corp., Armonk, NY, USA) for statistical analysis. The Shapiro-Wilk test was employed to examine the normal distribution of variables, considering $p > 0.05$ as an indication of normality. However, since some parameters did not follow a normal distribution, we applied a non-parametric Friedman Two-Way Analysis of Variance (ANOVA) to assess differences between various conditions. If significant differences were present, post-hoc pairwise t-tests using the Bonferroni method were conducted. We set the significance level at $p < 0.05$ to determine any significant changes in muscle activations throughout the exhaustive running trial.

4.4 Results

Figure 29 depicts the variations in peak amplitudes among different muscles during the mid and end stages, compared to the initial stage. Statistical analysis was conducted using the peak amplitudes recorded at these stages. The results indicated notable reductions in the maximum amplitudes of TFL, Gmax, Gmed, and BF during the exhaustive run. Significantly, all four muscles displayed notable decreases in activation during both the mid and end stages when compared to the initial stage. Additionally, Figure 29 highlights a notable decline in Gmax activation during the mid stage, decreasing to approximately 72.09% of its initial activation. In contrast, TFL, Gmed, and BF maintained over 80% of their initial activation, suggesting a relatively rapid decrease in Gmax activation specifically during the mid stage of the exhaustive run.

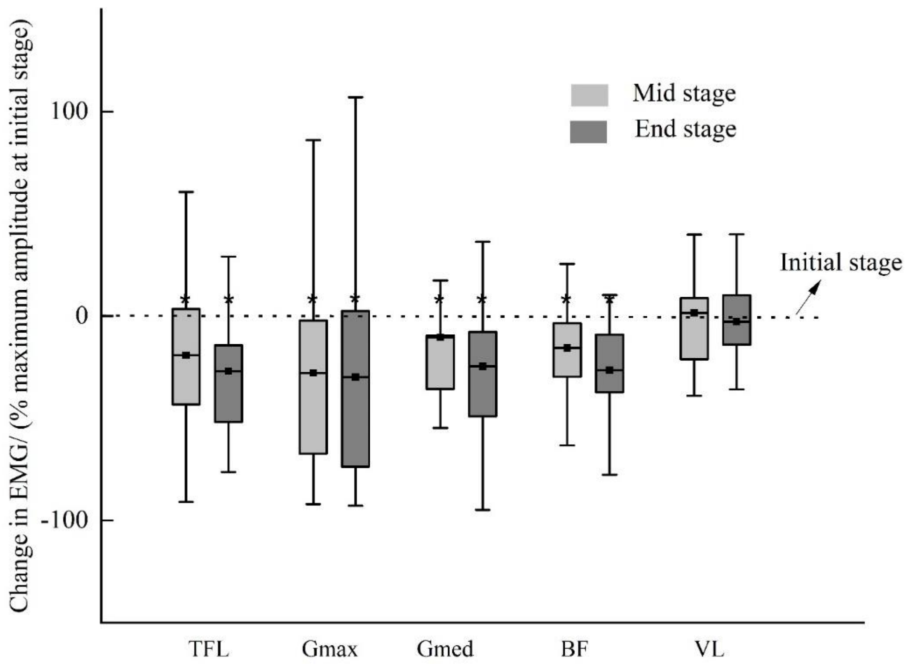


Figure 29 Change in EMG amplitude for TFL, Gmax, Gmed, BF, VL. The statistical results were performed for the maximal amplitudes. * Significant difference between different stages at p-value < 0.05.

Figure 30 illustrates the EMG activation patterns of the TFL, Gmax, Gmed, BF, and VL muscles at different stages. The timing of muscle activation and relaxation during these stages is summarized in Table 7. Specifically, we compared the onset of muscle activation during the late swing phase with the onset during the stance phase, excluding brief activations in the swing phase (e.g., Gmax). Importantly, there were no significant differences detected in the onset and offset timings between the initial, mid, and end stages. This suggests that the muscle firing timing remained consistent throughout the entire exhaustive running period. Figure 31 provides additional insight into the consistent activation patterns of the muscles during exhaustive running, presenting relative muscle timing based on a defined detection threshold.

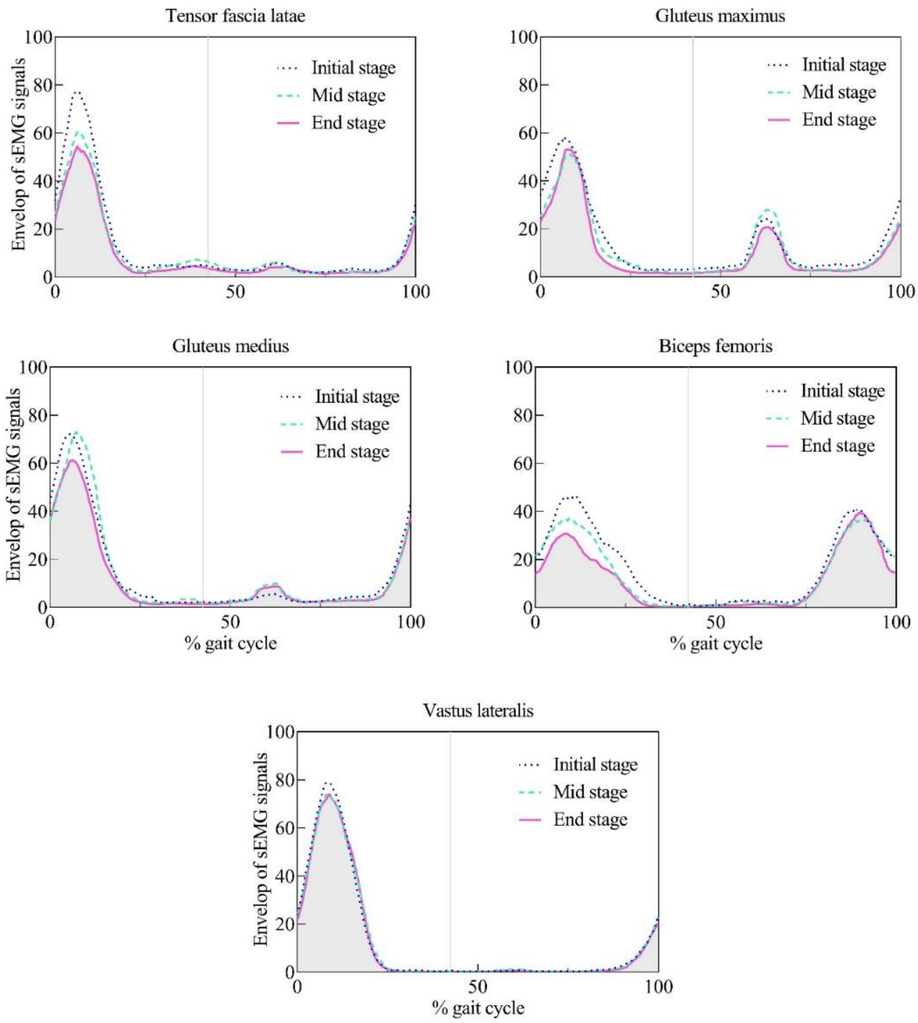


Figure 30 EMG activity patterns during an exhaustive run. Heel strike is at 0% and 100% of the stride.

Table 7 Muscle timing at different stages (mean ± SD)

Variable		Initial stage (% gait circle)	Mid stage (% gait circle)	End stage (% gait circle)	*P*
TFL	onset	96.35±3.75	96.37±3.24	96.65±3.73	0.529
	offset	16.55±3.97	16.92±3.81	16.39±3.66	0.424
Gmax	onset	94.29±5.81	93.43±6.01	94.46±6.48	0.435
	offset	17.81±5.19	18.24±4.49	16.52±4.01	0.061
Gmed	onset	94.01±3.57	93.41±4.28	93.56±3.27	0.157
	offset	17.68±4.21	17.71±3.91	17.39±4.06	0.237
BF	onset	79.62±8.41	79.88±8.35	79.91±8.26	0.876
	offset	20.62±9.49	19.95±9.75	19.08±8.64	0.585
VL	onset	96.55±4.00	95.76±3.95	95.73±3.44	0.131
	offset	18.45±3.34	19.15±3.33	19.16±2.75	0.056

* Significant difference between stages at P value < 0.05.

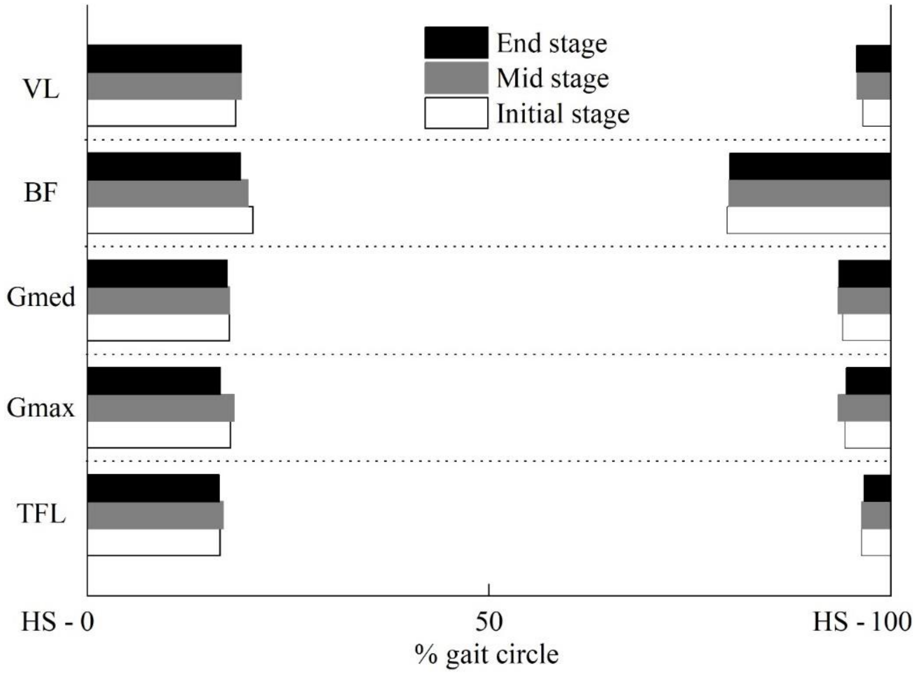


Figure 31 Relative muscle timing. HS: heel strike.

4.5 Discussion

The study aimed to investigate how changes in muscle activities during running affect the behavior of the iliotibial band (ITB). The study revealed significant decreases in the activation of the TFL, Gmax, Gmed, and BF muscles during the mid and end stages of an exhaustive run. Nevertheless, there were no significant alterations observed in the timing of muscle firing across the running stages. However, no significant changes in the timing of muscle firing were found throughout the running stages. These alterations in muscle activations could potentially influence the behavior of the ITB during an exhaustive run.

The alterations in the strategy of myofascial force transmission have been linked to the occurrence of overuse injuries in running, as the structural connection between the muscles and connective tissues play a role (Wilke et al., 2019). Additionally, prolonged and strenuous running can result in the development of an unfamiliar and compensatory running technique (Van Wilgen & Verhagen, 2012). Based on the hypothesis, it was suggested that the behavior of ITB could be modified during an exhaustive run. In the mid and end stages, particularly in the mid stage, there was a notable reduction in activation observed in the TFL and Gmax muscles. This decrease in TFL and Gmax activation could potentially contribute to alterations in the behavior of ITB. Hutchinson et al. (2022) created a simplified diagram to depict the correlation between tension in the ITB and compression forces on the knee. A previous study suggests that heightened tension in the ITB may result from underactivity of the gluteal muscles and overactivity of the TFL (Besomi et al., 2020). When examining the duration of running, both the TFL and Gmax showed weak activation as it progressed. Nevertheless, during the mid stage, Gmax exhibited a rapid amplitude decrease of 27.91%, surpassing the 19.17% decrease observed in TFL. This imbalance could result in overactivity in the TFL and underactivity in the Gmax compared to the initial stage. Furthermore, during the mid and end stages, reduced activity in Gmed may prompt compensatory activation of the TFL, resulting in heightened tension within the ITB

(Friede et al., 2021). The heightened tension within the ITB results in an increase in the compression force applied by the ITB on the lateral femoral epicondyle. To elucidate overuse injuries stemming from repetitive tissue loading, Bertelsen et al. (2017) proposed a conceptual model involving the accumulation of microdamage. Without proper adaptation, repetitive loading can lead to overuse injuries. Insufficient strength in the hip abductor muscles can lead to a significant inward movement of the hip joint, causing increased stress on ITB while running in the stance phase (Noehren et al., 2007). The modified activation patterns of the TFL and Gmax may contribute to an excessive strain on the ITB, functioning to knee joint stabilization.

The distal portion of the ITB comprises a deep capsule-osseous layer that originates close to the lateral gastrocnemius tubercle. Subsequently, this layer attaches to the anterolateral aspect of the proximal tibia at the lateral tibial tubercle (Godin et al., 2017; Runer et al., 2016). In addition to the ACL, which is less effective during high knee flexion angles, ITB has been shown to act as a secondary stabilizer for tibia rotation (Kaplan & Jazrawi, 2018). When the heel strikes the ground during running, the trunk tilts towards the same side of the strike to absorb shock. The contraction of the hip muscles opposes and stabilizes the lateral flexion of the trunk. To provide stability to the knee joint, ITB applies compression force against the lateral femur through tensioning, aiding in maintaining joint integrity (Hutchinson et al., 2022). Muscular activation during running was associated with a reduction in joint laxity, and this reduction was correlated with the duration of the run (Tsai et al., 2009). In the middle and late stages of exhaustive running, there was a significant decrease in the maximum activation of the TFL, Gmax, and Gmed. This reduction could suggest a limited capability to control the knee joints. As a compensatory running pattern, increased compression force on the lateral femoral epicondyle from the iliotibial band

(ITB) may be employed to enhance stability in the knee joint during exhaustive running.

The VL is an additional muscle involved in tensioning the iliotibial band (ITB), resulting in lateral stabilization of the pelvis. This action is facilitated by the "hydraulic amplifier" effect on the fascia latae. (Grimaldi, 2011). The contraction of the VL serves as an adjustable lever arm, capable of extending the distance between the ITB and the femoral shaft. This contraction results in heightened tension within the ITB. Remarkably, the maximum amplitudes of VL remained consistently high throughout the exhaustive running, which revealed the same results in the study of Rabita et al. (2013). The observed outcomes may be attributed to the compensatory approach adopted during running (Rabita et al., 2013). In contrast, the BF exhibited a notable reduction in maximum amplitude during the mid and end stages compared to the initial stage of exhaustive running. BF serves as a lateral stabilizing component for the knee joint, attaching to the posterior edge of ITB (Vieira et al., 2007; Whiteside & Roy, 2009). When the BF contracts, it may decrease the internal rotation of the tibia, resulting in reduced tension in ITB (Kwak et al., 2000). Additionally, significant decrease in BF activation is associated with decreased restriction on internal tibia rotation. As a result, during intense running activities, heightened traction in the ITB might potentially offer improved stability for the knee joint.

In this study, the amplitudes of muscles were utilized to characterize the overall muscle activation levels for the tasks. However, the timing parameters such as the onset and offset of specific muscles were analyzed to determine when these muscles were activated and deactivated. The results indicated that there were no significant differences in the onset and offset times of the TFL, Gmax, Gmed, BF, and VL muscles throughout the exhaustive running period. The results were consistent with the findings from Brown et al. (2019) that revealed no significant variances in the onset timing of TFL, Gmax, and Gmed during an exhaustive run. Changes in the timing of muscle activation have been linked to an increase in hip adduction angles, potentially resulting in compression at the femoral epicondyle (Fairclough et al., 2007). Nonetheless, the findings of the present study did not indicate any timing changes in the healthy ITB during a high-intensity run. To explore the activation timing of the muscles, future studies could consider extending both the duration and intensity of the running protocol.

The present study had certain limitations to consider. Potential variations may exist between running on a treadmill and running outdoors. Participants were unable to adjust their running pace, which could have influenced muscle activation. Despite efforts given to skin preparation and electrode placement, it is important to consider the presence of movement artifacts and noise in surface EMG measurements. While laboratory footwear was provided to minimize footwear bias, some participants may not have been accustomed to it. Moreover, the exhaustive treadmill running was conducted at a speed chosen by each participant, leading to variations in speed between male and female participants. Future research could explore the disparities in muscle activities between males and females.

4.6 Conclusion

The purpose of this study was to investigate how changes in the muscles connected in series with ITB affect its behavior during an exhaustive run, using physiological parameters. The researchers observed lower activation levels in the TFL, Gmax, Gmed, and BF muscles during the mid and end stages of the run. The reduced activation of TFL, Gmax, and Gmed muscles may result in increased compression force on the lateral femoral epicondyle from the tibial tuberosity to provide stability to the knee joint. Additionally, the maximum amplitudes of Gmax showed a rapid decline in the mid stage, potentially leading to relative underactivity of Gmax and overactivity of TFL, which can increase tension in ITB. Furthermore, the BF muscle's activation also significantly decreased, potentially contributing to increased ITB tension. The study also examined the timing of activation and deactivation of these ITB-related muscles during exhaustive running, but no significant differences were found. The findings of this study contribute to our understanding of the behavior of a healthy ITB and may offer insights into the development of ITB-related pathologies, such as ITBS.

CHAPTER 5 EFFECTS OF RUNNING SPEEDS AND EXHAUSTION ON ILIOTIBIAL BAND STRAIN DURING RUNNING

5.1 Summary of the study

ITBS is one of the most prevalent overuse injuries for runners. The rate of strain within the ITB has been proposed as the leading factor contributing to the onset of ITBS. Increased running speed and exhaustion have the potential to modify biomechanics, thus impacting the strain rate experienced by the iliotibial band. This study aimed to investigate the effects of exhaustion states and running speeds on ITB strain and strain rate.

We recruited a group of 26 healthy runners, including 16 males and 10 females, performed running trials at normal and fast speeds before and after an exhaustive run. Data of reflective markers and GRF were collected with the sample frequency of 250Hz and 1000 Hz respectively, simultaneously. Then the data were used as input for the musculoskeletal model with ITB in OpenSim. The kinematics of the knee and hip joint, and biomechanics of ITB were calculated to perform the analysis.

The results suggested that both fatigue and running velocities significantly influenced the strain rate of the ITB. Upon reaching exhaustion, ITB strain rate exhibited a notable rise of approximately 3%, irrespective of the running speed. Furthermore, a rapid escalation in running speed resulted in a higher ITB strain rate, both before and after reaching exhaustion. These findings highlight the importance of considering exhaustion and running speed as potential factors contributing to the development of ITBS.

The study suggests that an exhaustion state and abrupt increases in running speed may elevate the risk of ITBS due to higher ITB strain rates. Therefore, it is crucial to consider injury prevention measures and manage training loads carefully, particularly when experiencing fatigue or engaging in high-speed running.

5.2 Introduction

Iliotibial band syndrome is one of the most prevalent overuse injuries in runners, which leads to lateral knee pain (Aderem & Louw, 2015). The ITB originates from the iliac crest and insets into Gerdy's tubercle distally with the attachments at the lateral femoral epicondyle (J. Fairclough et al., 2006). The strain rate of ITB was proposed to be a major factor in the development of ITBS via a musculoskeletal model incorporated ITB in OpenSim (Hamill et al., 2008). Some previous studies have discovered the influential factors to ITB strain.

The noted elevations in hip adduction and knee internal rotation angles have been linked with the increased ITB strain rate. (Ferber et al., 2010; Hamill et al., 2008). Sinclair et al. (2020) examined different movements including running, cut, and hop, which indicated that running and cut revealed higher strain than hop. Additionally, a narrower step width was revealed to increase ITB stain rate significantly when compared to a preferred condition (Meardon et al., 2012). Female was proposed to exhibit a higher peak strain and strain rate due to an increase in the hip internal rotation angle compared to males during overground running (Day & Gillette, 2019). Due to the complex anatomy attachments and insertions, ITB strain and strain rate might be changed by the biomechanics of the knee and hip joints.

Running speed is one of the main influential factors in gait. Compensatory strategies might be adopted when runners increase their speeds. One dominant strategy is that runners shift to increase the stride cadence and push on the ground with a higher frequency, which consequently applies more force on the hip joint as the running speed increases (Schache et al., 2014). The increase in the cadence was indicated to lead to a decrease in the hip adduction angle and hip adductor moment (Hafer et al., 2015). Faster running speed was indicated to knee joint loads per stride during overground running (Petersen et al., 2015). Due to the influences of the running speed on the biomechanics of the lower limb, the potential contributions of the running speed to the strain and strain rate of ITB still remain unknown.

Prolonged exhaustive running presents significant effects on the biomechanics of the lower limb joints, which are essential intrinsic factors to running-related injuries(Jewell et al., 2019; Mei et al., 2019). A significant increase in segment coordination variability was found with the prolongation of the running mileage (Chen et al., 2022). The moments of the hip joints were observed to increase significantly at the initial contact after a 5 km treadmill running (Mei et al., 2019). Consequently, an exhaustive run may lead to alterations in the strain and strain rate of ITB. Although Miller et al. (2007) investigated the mechanics of ITB during an exhaustive, some limitations were observed. The running trials were performed on the treadmill, which might lead to different kinematics and kinetics of the lower limb joints compared with overground running. Additionally, participants wore their own shoes during running, while another study proposed that footwear affected ITB strain significantly (Sinclair et al., 2019).

Few studies to date have examined the influence of the running speed on the strain and strain rate of ITB, which has been proposed to be the essential factor in the development of ITBS. Both the running speed and the exhaustion states were examined to alter the kinematics and kinetics of the lower limb. The purpose of the current study was to identify the influences of the running speed before and after an exhausted running on the strain and strain rate of ITB. Our hypothesis posited that both states of exhaustion and an escalation in running speed could result in an increase in ITB strain rate.

5.3 Methods

5.3.1 Participant information

A total of twenty-five participants, comprising 15 males and 10 females, were recruited for this study. Detailed participant information is provided in Table 8. All participants were in good health and had no history of lower limb injuries in the preceding year. Their ages ranged between 20 and 30 years, and their BMI was between 20 and 24. All participants were regular runners. Prior to the experiment, informed written consent was obtained from all participants The Human Subjects Ethics Sub-Committee of the Hong Kong Polytechnic University (Number: HSEARS20150121003) has approved this study.

5.3.2 Experimental equipment and procedure

The motion data were collected using an eight-camera motion capture system (Vicon, Oxford Metrics Ltd., Oxford, England) with a sampling frequency of 250 Hz. Additionally, a force plate (OR6, AMTI, Watertown, United States) recorded the ground reaction force at a sampling frequency rate of 1000 Hz, synchronized with the Vicon system. Prior to the formal tests, a total of 34 reflective markers were attached to specific anatomical landmarks on the participants' foot, leg, and torso. As depicted in Figure 32, the markers were placed at key locations, including the forehead, acromioclavicular joints, posterior and anterior iliac spines, three on the thigh arranged in a triangle, lateral and medial femoral epicondyles, three on the shank arranged in a triangle, lateral and medial malleoli, the heads of the first and fifth metatarsals, and the tip of the toe. Each segment was equipped with three markers to track three-dimensional motion accurately.

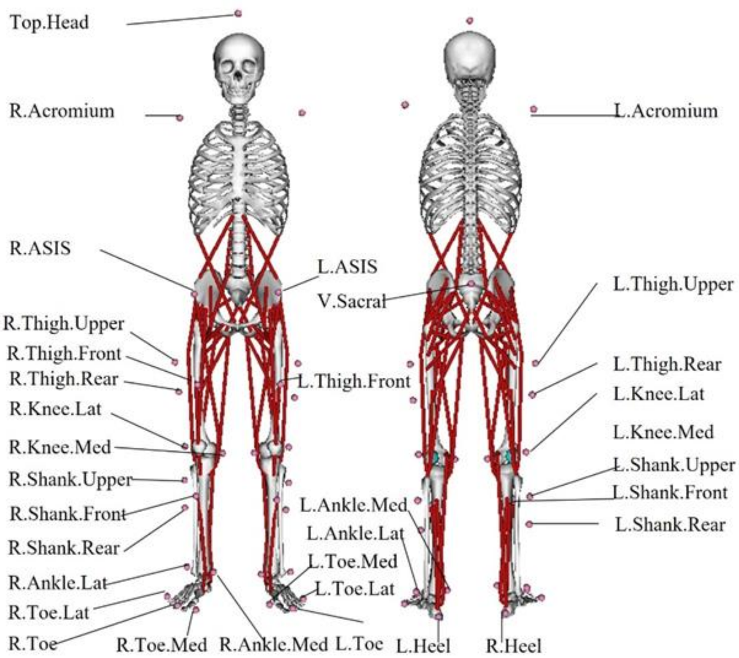


Figure 32 The musculoskeletal model with the marker set

Participants in the study were provided with laboratory-provided sports shoes (ARHQ025-4, Li-Ning Inc., Beijing, China) in their fitting size. They were encouraged to get familiar with the shoes and perform warm-up exercises. A static trial was conducted for each participant to scale the generic musculoskeletal model while standing on the force plate. Subsequently, dynamic trials were performed at the participants' normal running speed. Following this, participants were instructed to run at a speed exceeding 10% of their normal running pace, with successful trials confirmed by stepping on the force plate with their entire right foot. The speed tolerance for each trial at the same speed was maintained within ±5% to ensure accurate analysis. Five successful trials were collected for each condition. Afterwards, participants were asked to run on a treadmill at their self-selected speed for 30 minutes until they reached an exhaustive state. Following this, dynamic overground running trials were conducted. Prior to the formal tests, participants completed at least ten warm-up overground running trials to resume their natural gait. It's important to note that no rest was allowed for participants during the exhaustive state. Running speed was confirmed using the average speed of the anterior-posterior movement of the sacral marker, providing immediate feedback after each trial (Meardon et al., 2015). Further details of the running conditions are presented in Table 8. For all trials, marker trajectories and ground reaction forces were simultaneously recorded for analysis.

Table 8 Information of the participants and experiments

Participants characteristics		Value (mean ± SD)
Age / year		24.53 ± 3.24
Mass / kg		64.73 ± 11.65
Height / cm		170.90 ± 8.38
Pre-30 min-running	Normal speed / (m/s)	3.33 ± 0.36
	Fast speed / (m/s)	3.90 ± 0.41
Post-30 min-running	Normal speed / (m/s)	3.32 ± 0.36
	Fast speed / (m/s)	3.93 ± 0.38

5.3.3 Musculoskeletal multibody model

In this study, a musculoskeletal model of ITB was developed using OpenSim (Hamill et al., 2008). The model encompassed the foot, shank, patella, thigh, pelvis, torso, and head. The hip joint was designed with six degrees of freedom, enabling rotation and translation. The knee joint was modified to a three-degree-of-freedom joint from its original single-degree-of-freedom configuration. To simulate ITB, it was represented as an elastic structure extending from the iliac crest to Gerdy's tubercle, following the anatomical pathway of the TFL in the model (Hamill et al., 2008). To establish the rest length of ITB, the length of ITB in the static calibration model was scaled by the data obtained from the static trial. Subsequently, the strain and strain rate of ITB were calculated using the scaled musculoskeletal model driven by dynamic data for each participant. This approach allowed the investigation of the changes in ITB biomechanics during the dynamic running trials.

The length of ITB in the stance phase during running could be obtained from the model. The strain of ITB at each time step was calculated through the formula (Hamill et al., 2008):

$$Strain = \frac{L_i - L}{L}$$

where L_i is ITB length at time i during the stance phase and L is the rest length. The stain rate of ITB was calculated through the first central difference method for each time step, shown as follows:

$$Strain\ rate = \frac{Strain_{i+1} - Strain_{i-1}}{Time_{i+1} - Time_{i-1}}$$

5.3.4 Data analysis

In this study, all statistical analyses were conducted using SPSS software (Version 16.0; SPSS Inc., Chicago, IL, USA), with a significance level set at 0.05. The significant level for all tests was set at 0.05. The Shapiro-Wilk test was utilized to ascertain the adherence of variables to a normal distribution. Since all variables exhibited normal distribution, independent paired t-tests were used to assess group differences for different speed levels. To investigate the differences in kinematics and kinetics of the hip and knee joints, as well as the strain and strain rate of ITB between exhaustion (before and after) and different speed levels, a two-way repeated measures analysis of variance (ANOVA) was employed. To confirm the sphericity assumption of the repeated-measures ANOVA, Mauchly's test was utilized, and all variables met the assumption. For significant ANOVA interactions, post hoc tests with Bonferroni corrections were performed to reveal specific differences. The post hoc tests, with Bonferroni adjustment, confirmed significant differences between various conditions.

5.4 Results

All participants completed running trials at both their preferred speed and a faster speed, both before and after the treadmill running session lasting 30 minutes at their self-selected speed. The normal and fast running speeds were carefully controlled to remain at nearly the same level. To ensure consistency, real-time feedback on speed was provided to participants throughout the trials. The kinematics and kinetics of the knee and hip joints were recorded to facilitate the subsequent analysis. Additionally, the strain and strain rate of the relevant structures were calculated for each condition, enabling statistical comparisons among different conditions.

No significant interactions were observed between the exhaustion states and running speeds for all the variables of interest. However, both running speed and exhaustion states influenced the kinematic parameters of the knee and hip joints, as presented in Table 9. Specifically, the hip flexion angle showed a significant increase of approximately 5° with improved running speed and about 3° when comparing before and after the exhaustive running at the same speed. The hip adduction angle exhibited a slight increase ($p=0.002$) at the fast speed after the exhaustive run but showed no significant difference when the running speed changed. In contrast, the hip internal rotation angle significantly decreased ($p=0.001$) after an exhaustive run with increased running speed. Similarly, the knee flexion angle displayed a similar trend to that of the hip joint. It showed a significant increase with faster running speed and was also observed to increase after an exhaustive running session. The changes in the peak angle for the hip and knee joint are visually represented in Figures 33 and 34, respectively.

The kinetics of the knee and hip joints were also influenced by both exhaustion states and running speeds, as indicated in Table 10. The peak moment of hip extension tended to increase significantly with improved running speed under both pre- and post-exhaustion states. However, exhaustion states did not show significant effects on the hip extension moment. On the other hand, the peak moment of hip internal rotation exhibited a significant increase (p=0.007) during the exhaustive state when the running speed was increased. Furthermore, the peak moment of knee extension demonstrated a significant increase (p=0.002) at the normal running speed after exhaustive running. Nevertheless, running speed did not demonstrate significant effects on the knee extension moment.

Peak ITB strains showed similar values when the running speed increased, as presented in Table 11. Figure 35 displays ITB strain, while Figure 36 shows ITB strain rate. Exhaustion states showed no significant impact on ITB strain, yet both exhaustion states and running speeds demonstrated notable effects on ITB strain rate. The ITB strain rate exhibited an approximate 10% increase when subjects ran at a higher speed, both prior to and following exhaustion. Moreover, when the running speed was kept constant, the ITB strain rate displayed a significant increase of about 4% after an exhaustive run. The muscle forces of TFL and Gmax are depicted in Figure 37 and 38, respectively. It was observed that the muscle force of TFL increased with higher running speeds, whereas the muscle force of Gmax remained nearly constant before and after exhaustion.

Table 9 Peak joint angles for different speeds for pre - and post- conditions.

Joint angles /°	Before Exhaustion (mean ± SD)		After Exhaustion (mean ± SD)	
	Normal speed	Fast speed	Normal speed	Fast speed
Hip flexion	29.70 ± 4.00	34.96 ± 5.19[b]	32.40 ± 4.85[a]	37.86 ±4.55[a, b]
Hip adduction	13.89 ± 2.85	14.22 ± 3.21	15.15 ± 2.99	15.48 ± 2.82[a]
Hip internal rotation	7.90 ± 6.56	6.97 ± 5.92	8.01 ± 5.82	5.63 ± 6.22[b]
Knee flexion	44.59 ± 3.55	46.81 ± 3.86[b]	46.94 ± 3.69[a]	49.09 ± 3.43[a, b]
Knee adduction	5.92 ± 3.28	6.44 ± 3.46	6.26 ± 3.91	6.71 ± 4.11
Knee internal rotation	7.67 ± 5.29	7.99 ± 5.84	8.92 ± 6.22	8.722 ± 6.70

a: significant difference in exhaustion states from the matching speed condition ($p < 0.05$); b: significant difference in speed levels from the matching pre-exhaustion condition ($p < 0.05$)

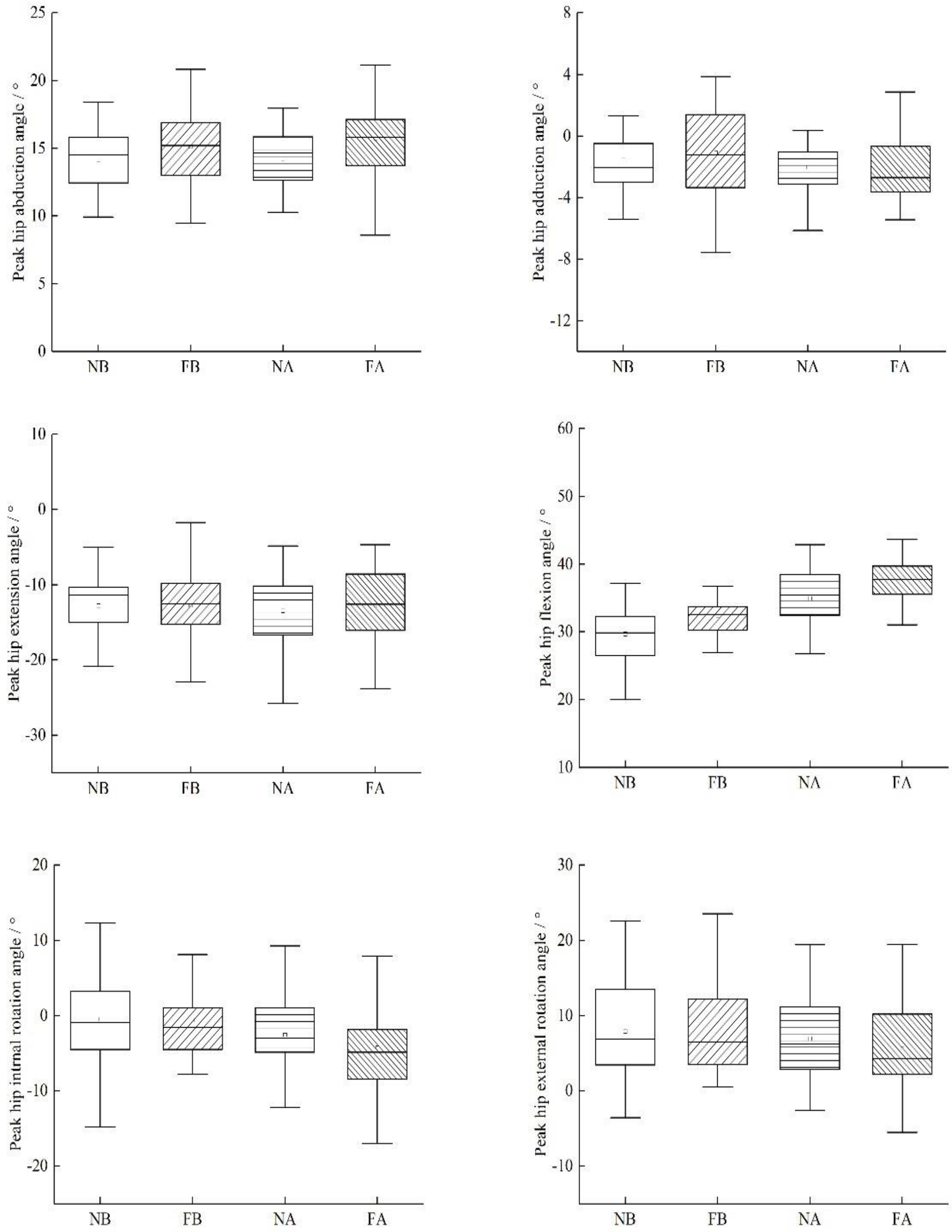


Figure 33 Changes in peak angles for the hip joint. NB: normal speed before exhaustion; FB: fast speed before exhaustion; NA: normal speed after exhaustion; FA: fast speed after exhaustion.

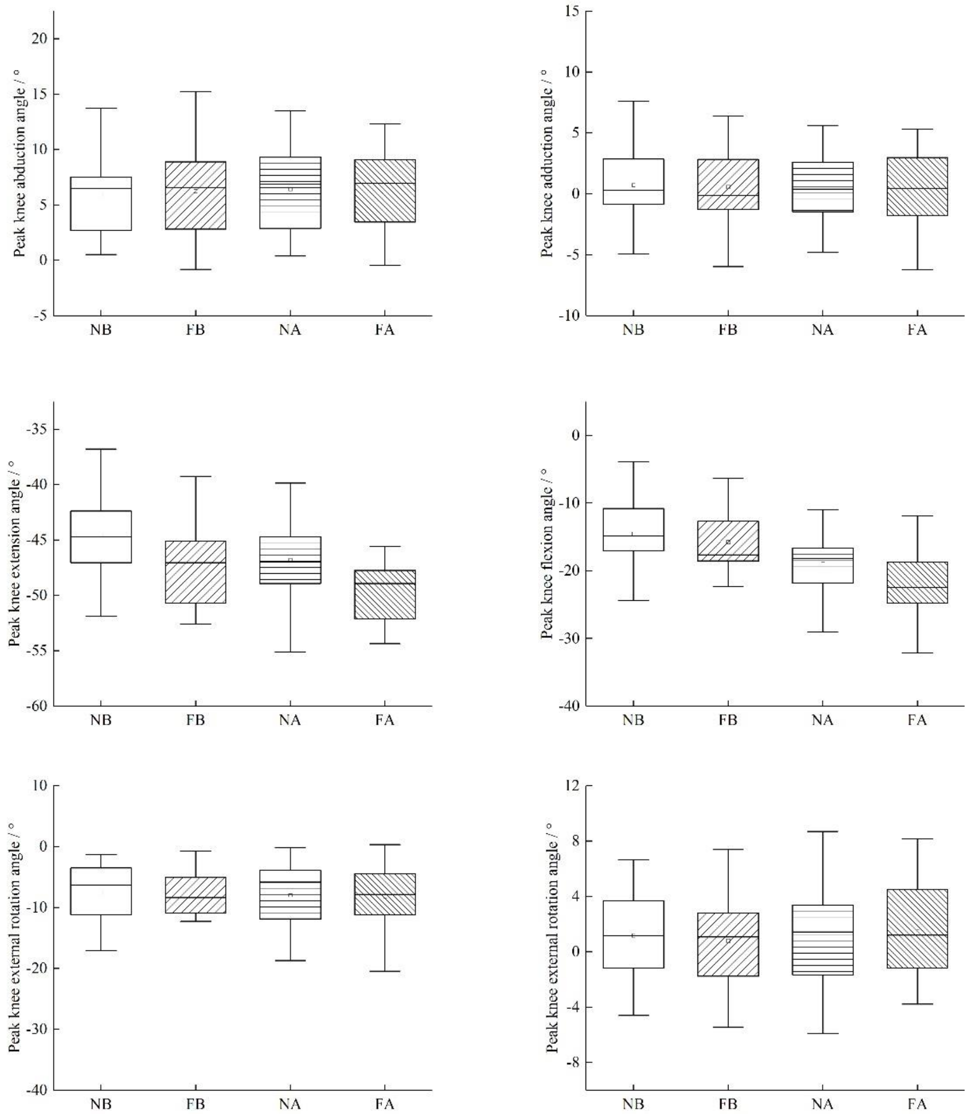


Figure 34 Changes in peak angles for the knee joint.

Table 10 Peak joint moments for different speed for pre- and post- conditions

Moment (Nm/kg)	Before Exhaustion (mean ± SD)		After Exhaustion (mean ± SD)	
	Normal speed	Fast speed	Normal speed	Fast speed
Hip extension	2.33 ± 0.62	2.77 ± 0.51[b]	2.42 ± 0.47	2.87 ± 0.60[b]
Hip abduction	1.18 ± 0.39	1.42 ± 0.72	1.32 ± 0.42	1.47 ± 0.47
Hip internal rotation	0.26 ± 0.11	0.32 ± 0.14	0.31 ± 0.10	0.36 ± 0.13[b]
Knee extension	1.73 ± 0.32	1.75 ± 0.30	1.87 ± 0.36[a]	1.86 ± 0.32
Knee adduction	0.66 ± 0.30	0.76 ± 0.48	0.73 ± 0.31	0.79 ± 0.38
Knee internal rotation	0.25 ± 0.15	0.31 ± 0.26	0.28 ± 0.19	0.32 ± 0.20

a: significant difference from the matching speed condition ($p < 0.05$)

b: significant difference from the matching pre-exhaustion condition ($p < 0.05$)

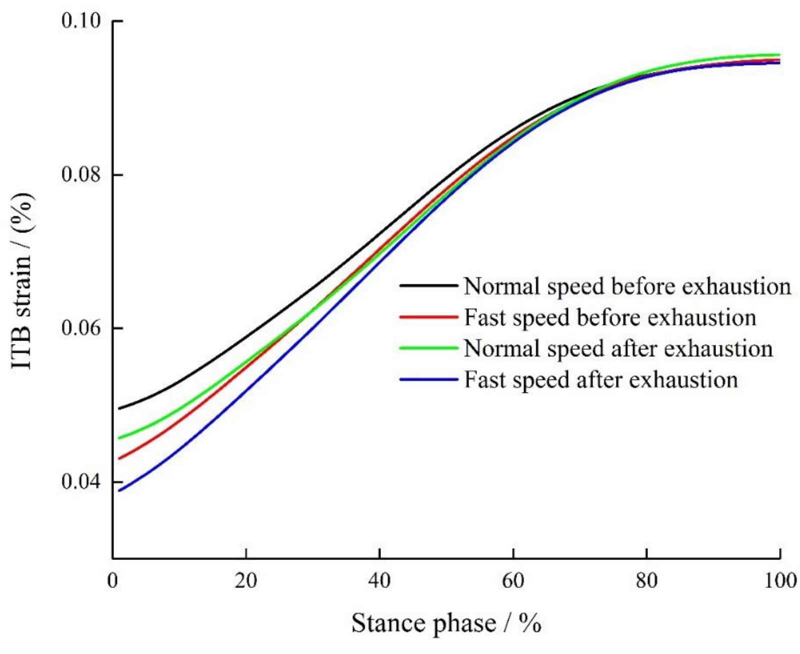


Figure 35 Changes in the ITB strain

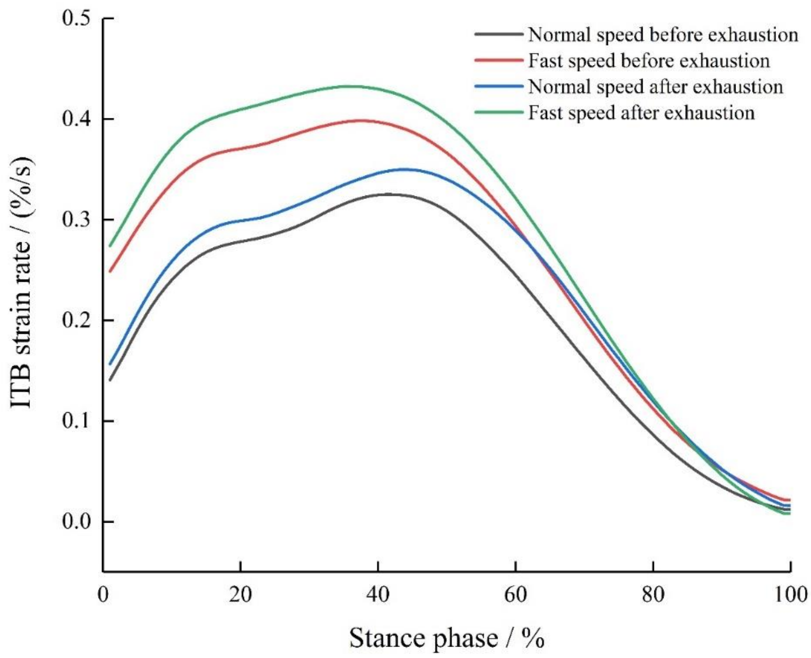


Figure 36 Changes in the ITB strain rate

Table 11 Peak joint moments for different speed for pre - and post conditions

Variables	Before Exhaustion (mean ± SD)		After Exhaustion (mean ± SD)	
	Normal speed	Fast speed	Normal speed	Fast speed
ITB strain / %	9.54 ± 0.59	9.56 ± 0.66	9.63 ± 0.70	9.53 ± 0.80
ITB strain rate / (%/s)	35.05 ± 7.79	44.76 ± 8.86[b]	38.57 ± 9.04[a]	48.44 ± 10.90[a, b]

a: significant difference in exhaustion states from the matching speed condition ($p < 0.05$); b: significant difference in speed levels from the matching pre-exhaustion condition ($p < 0.05$)

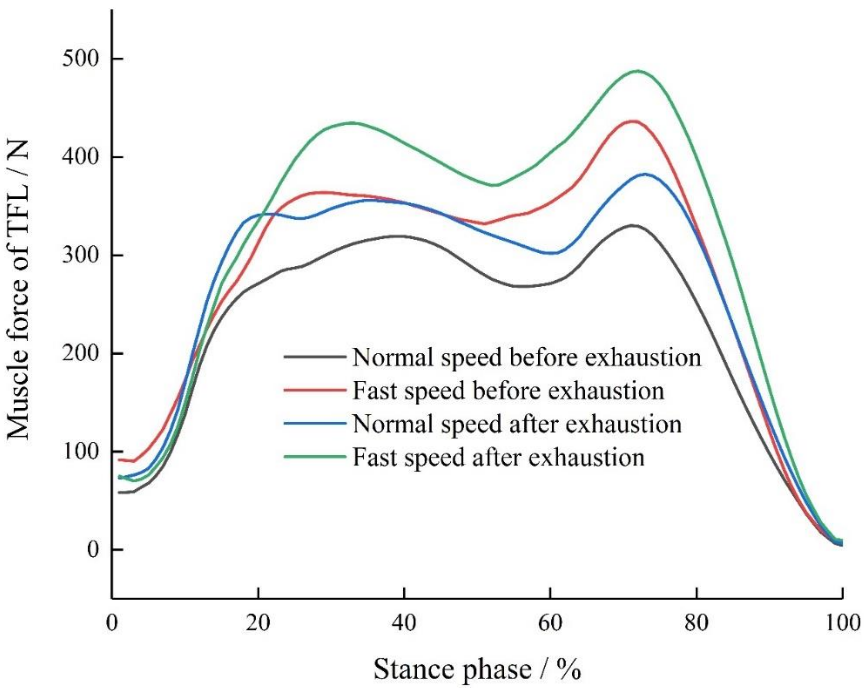


Figure 37 Muscle force of tensor fascia latae muscle for all the condition

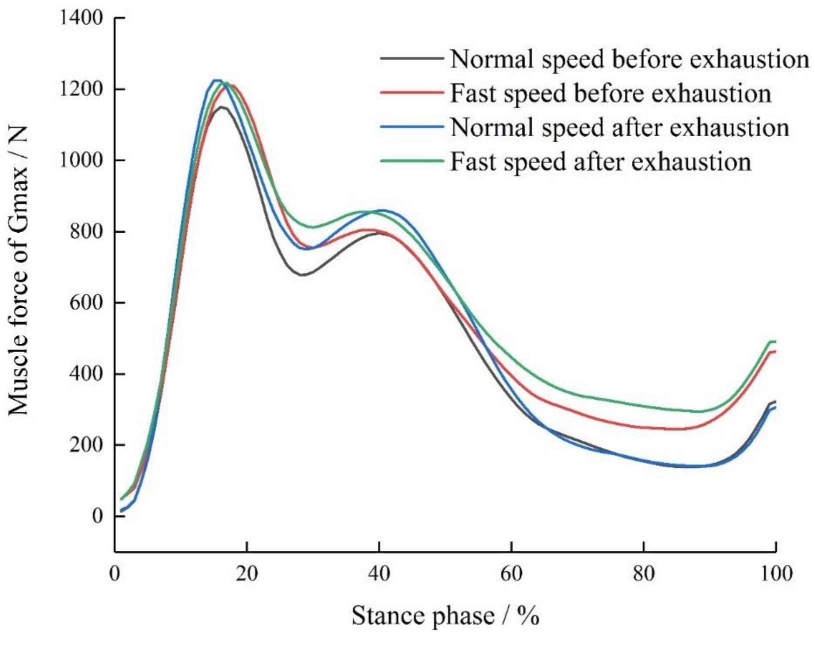


Figure 38 Muscle force of gluteus maximus muscle for all the condition

5.5 Discussion

The current study aimed to explore how exhaustion states (before and after an exhaustive run) and running speed altered the behaviors of ITB due to the changes of kinematics and kinetics of knee and hip joints. Both the flexion angles of the knee and hip joints revealed significant increases when the running speed improved. Also, the angles increased significantly when participants experienced an exhaustive run. The peak angle of the hip adduction also increased significantly when comparing pre- and post-exhaustion state at a fast speed. The running speed exhibited a negative relationship in the hip internal rotation. The peak angle of the hip internal rotation revealed a significant decrease when the running speed improved after an exhaustive run. The kinetics of the hip and knee joints also displayed alterations due to different exhaustion states and running speeds. The peak moment of the hip extension increased significantly due to the increase of the running speed. Additionally, the peak hip internal rotation moment revealed an increase when the running speed improved after an exhaustive run. However, no significant interactions were observed between these two factors.

The ITB strain remained similar when running at different speeds under different running states. Nevertheless, ITB strain rate possessed a consistent increasing trend along with running speed. Furthermore, the ITB strain rate exhibited a significant increase following an exhaustive run. The increase in the ITB strain rate has been proposed to be a major causative factor in the development of ITBS (Hamill et al., 2008). The ITB strain rate showed a similar increase for the identical running speed. Furthermore, the strain rate increased by about 10% when comparing the fast speed level to the normal speed level under the same exhaustion state. The improvement of the running speed was revealed to influence the ITB strain rate in a more severe way due to the bigger increase when the running speed was fast. The mechanical behaviors of ITB might be affected by the alteration in the proximal factors in the hip joint and the distal factors at the knee joint due to the multiple anatomical insertions (Day & Gillette, 2019). Thus, the changes in the joint biomechanics could make potential contributions to the ITB strain rate.

The current study found that the peak hip adduction angle increased when run at a fast speed, which is in agreement with previous findings (Radzak & Stickley, 2020). The peak hip adduction angle also showed an increase when ran at a normal speed after exhaustion, though not significantly. However, the increase in the angle was proposed to be related to the development of ITBS (Noehren et al., 2007). The ITB functions by counteracting hip adduction to fulfill the requirement of lateral hip stabilization in the proximal region (Fredericson et al., 2000). Consequently, it is possible for increased hip adduction angles to result in heightened tension within ITB. As the hip adduction increases in the exhaustion state, a higher moment of the hip adduction moment might occur due to the higher eccentric demand of gluteal musculature (Ferber et al., 2010). Though no significant effects were displayed in the peak moment of the hip abduction, an increase was still observed. The increase in the hip adduction was related to the weaker hip strength, which is an essential factor to limit and control the peak of the hip adduction angle during the stance phase of running (Brindle et al., 2020). The exhaustive run was proposed to induce weak hip strength that would lead to an increase of peak hip adduction angle. Following an exhaustive run, the peak ITB strain rate increased by around 4% for the corresponding running speed. This might be induced by the increase of the hip adduction angle. Consequently, exhaustion state might be an influential factor to the higher ITB stain rate.

With respect to the influences of the running speeds, the peak angle of the hip internal rotation tended to decrease as the running speed increased when participants were in post-exhaustion state. During the pre-exhaustion state, a small decrease also was observed in the hip internal rotation angle. There were no significant differences for the similar speeds between pre- and post-exhaustion states, which agreed with a previous study (Brown et al., 2016). The increase in the peak hip internal rotation angle was indicated to reduce the ITB strain rate (Day & Gillette, 2019). Instead, as the running speed increased, the peak angle of the hip internal rotation decreased, which might lead to an increase in the ITB strain rate. Repetitive loading to the lower limb joints might lead to overuse injuries (Willwacher et al., 2022). The decrease in the range of hip internal rotation posed a trend towards the increasing odds of ITBS. Running speeds also led to a noteworthy increase in the peak knee flexion angle. A greater knee flexion angle was proposed to increase the elongation of ITB, which might lead to a larger ITB strain rate (Day & Gillette, 2019). The results might imply that a rapid increase in training intensity, especially for a running speed increase, could cause an increasing risk for ITB-related injuries, especially for ITBS.

In this study, a few limitations should be considered. Participants in this study are healthy runners, of which the results might not be generalized to injured populations. Additionally, gender differences were not included in the investigation in the current study. The musculoskeletal model has the limitation to incorporate the complex anatomical structures of ITB. Validation of the ITB model would be very difficult with non-invasive measures. The ITB was simulated to be completely passive and its length was represented as the length between the connected markers (Day & Gillette, 2019). All the parameters analyzed in the study were calculated using a musculoskeletal model. However, the peak strains (9% − 10%) are within the normal in *vivo* limits of elastic structures (5% − 12%) (Monti et al., 2003), and lower than the failure strain of about 13% reported in a cadaver testing from a previous study (Birnbaum et al., 2004). As ITB is simulated as passive soft tissue in the model, influential factors from the in-series musculatures of ITB are not discussed in the current study. Although the running speeds are controlled to maintain the same before and after the exhaustion through real-time feedback during tests, small differences still exist for the same running speed level. Additionally, we discussed the effects of the normal speed condition and fast speed condition on the strain and strain rate of ITB before and after exhaustion. The model could not calculate the compression force between ITB and the femoral epicondyle, which is the pain area for ITBS, future investigations should incorporate the complex anatomy structures of ITB with in-series muscles to examine the impact factors from the muscles and the compression forces. Finally, all the participants recruited in the current study were healthy runners, so the results may not propagate to injured populations.

5.6 Conclusion

In this study, a musculoskeletal model was utilized to investigate the impact of exhaustion states and running speeds on the ITB strain and strain rate. The analysis revealed an increase in the hip adduction angle during the exhaustion state, while the peak hip internal rotation angle decreased with higher running speeds. Notably, both exhaustion states and running speeds exhibited a significant effect on the ITB strain rate. These findings suggest that a sudden escalation in training intensity may potentially increase the risk of higher ITB strain rates in healthy runners, leading to injuries like ITBS.

CHAPTER 6 INVESTIGATION OF INTERACTION BETWEEN THE ILIOTIBIAL BAND AND FEMUR DURING RUNNING USING A MUSCULOSKELETAL MULTIBODY DRIVEN FINITE ELEMENT MODEL

6.1 Summary of the study

Knee stability could be obtained through the compression force on the lateral femoral epicondyle applied by ITB. However, ITBS, as one of the most common overuse injuries for runners, was proposed to be caused by the excessive compressive force on the lateral femoral epicondyle. Thus, it is crucial to investigate the compression pressure between ITB and the lateral femoral epicondyle. So far, the compression pressure has not been examined.

In the current study, a subject specific ITB FE model was established to examine the changes of the compression pressure during the stance phase in running under different conditions. The FE model was reconstructed from a healthy subject MRI. The model consists of pelvis, femur, patella, tibia, TFL, Gmax, ITB, femoral/tibial cartilage, and meniscus. The whole FE model was established in software Abaqus. Mesh generation was confirmed and performed for all the parts after a mesh convergence test. The current model was validated through comparison of the results with findings from previous studies. We collected data of running trials that conclude four conditions (normal speed before exhaustion, fast speed before exhaustion, normal speed after exhaustion, and fast speed after exhaustion). The data of reflective markers and GRF were used as input for the musculoskeletal model. Through the musculoskeletal model, the ankle joint forces, muscle forces, and segment kinematics were obtained, which could provide the boundary and loading input for the ITB FE model. This musculoskeletal multibody driven finite element model could provide much more insights about the biomechanics of ITB using a dynamic solver.

The results of the model indicated a peak value of the compressive pressure on the lateral femoral epicondyle around knee flexion angle of 30° where it was proposed to be the pain zone for ITBS runners. Additionally, the stress in the ACL also indicated a peak value between 20° and 30° knee flexion angle. The ACL plays as the first stabilizer for the knee joint when the knee flexion angle is lower than 30°. Following those movements, ITB functions to maintain stability in the knee joint. Running speed was proposed to have a significant effect on the compressive pressure on the lateral femoral epicondyle. Faster speeds would cause in higher compressive force. Furthermore, when experiencing exhaustion, there can be changes in muscle activation, resulting in an increase in compressive force on the lateral femoral epicondyle.

The current study quantified the compressive pressure on the lateral femoral epicondyle, which was proposed to be the primary factor to ITBS. Influential factors such as running speed and exhaustive running state were discussed to indicate the potential consequences. our study's subject-specific approach and consideration of dynamic loading conditions provide valuable insights into the biomechanics of ITB during running. These findings have important implications for injury prevention, rehabilitation strategies, and the development of personalized interventions for individuals with ITB-related issues.

6.2 Introduction

The ITB is the thickening portion of the fascia latae that originates from the iliac crest proximately (Hutchinson et al., 2022). With the connection to the TFL, ITB passes over the greater trochanter without a fixation, but fixes to the distal femur and finally inserts into the Gerdy's tubercle (Birnbaum et al., 2004) and the patella via the lateral retinaculum (Godin et al., 2017). Additionally, the Gmax partially inserts into ITB, which could be substantially regarded as a tendon of Gmax (Birnbaum et al., 2004), and partially attaches to the femur (Eng et al., 2015). Due to the complex anatomical structures, ITB were considered to play important roles in hip and knee joints. Birnbaum et al. (2004) constructed a finite element model to reveal that the hip centralizing forces provided by ITB contribute to the stability of the hip joint and prevent hip luxation. For the preservation of knee joint stability, the femur experiences a compressive force exerted by the ITB (Hutchinson et al., 2022). During high flexion angles over 30° of the knee joint, ITB could decrease tibial translation in the anterior direction for the ACL – deficient knee based on the basis of cadaveric studies (Yamamoto et al., 2006). Godin et al. (2017) reanalyzed the distal anatomy of ITB, thereby drawing the conclusion that ITB contributes to the rotational knee stability. Most studies examined functions of ITB via cadaveric experiments in a passive way without the consideration of the force transmission from the TFL and Gmax. The force transmission from the in-series muscles determines the mechanical functions of ITB, which warrants further exploration.

ITBS is a common overuse injury recognized for inducing pain at the lateral femoral epicondyle (Falvey et al., 2010). ITBS was indicated to account for 12% of running-related injuries (Taunton et al., 2002). Runners with severe pain caused by ITBS at the lateral knee might affect their physical activities. The aetiology of ITBS was proposed to be the friction between ITB and the lateral femoral epicondyle in previous studies (Ellis et al., 2007; Fredericson & Wolf, 2005). This proposed mechanism of friction was questioned as ITB attaches to the lateral femoral epicondyle firmly without relevant movements (Hamstra-Wright et al., 2020). The ITBS is caused by the increase of the compression force between ITB and the lateral femoral epicondyle, which might be related to the dysfunctions of the in-series muscles of ITB (Fairclough et al., 2007). Consequently, a more specific understanding of the relationship between ITB and the lateral femoral epicondyle might provide more evidence for the aetiology and treatments for ITBS.

Previous studies adopted many approaches to examine the biomechanics of ITB. Miller et al. (2007) established a musculoskeletal model of the lower extremity. The model of ITB was simple and lacked essential anatomical features, such as the in-series musculatures that influence the biomechanics of ITB largely. Ultrasound elastography units also were used to measure ITB strain that is proposed to be the key indicators for ITBS (Hamill et al., 2008) during wight-bearing (Kim et al., 2020). However, more details of the internal biomechanical information might not be obtained via experimental approaches, such as stress distribution in soft tissues. Additionally, the mechanics of the joints are affected by the muscle forces that in turn could be altered through the kinematics of the joints, of which the interaction was proposed to have a significant effects on the biomechanics of the knee joint (Hume et al., 2019; Li, 2021). A number of studies examined activities of TFL and Gmax using surface electromyography to explore their potential contributions to ITB during physical activities (Baker et al., 2018; Watcharakhueankhan et al., 2022). However, the fidelity of this approach is questioned due to the substantial artifacts and the uncertainties during the acquisition (Beretta-Piccoli et al., 2019). Another popular approach is multibody musculoskeletal model (MSK) that assumed muscles as on-dimensional muscle model (Chen et al., 2018; Peng, Wang, et al., 2021), which thus lacked the essential parameters, such as strains of muscles.

The current study aims to (1) develop a subject-specific FE lower limb model to simulate running; (2) simulate the biomechanics of ITB during running; (3) simulate the compression force between ITB and the lateral femoral epicondyle.

6.3 Materials and Methods

6.3.1 General information

A healthy young male adult (age 30, height 173 cm, and 70 kg) with running habit was recruited in this study. The participant was free of any neuromuscular disorders or surgery histories. He was informed of the research contents and signed the consent form before the formal experiments. The Human Subjects Ethics Sub-Committee of the Hong Kong Polytechnic University (Number: HSEARS20150121003) has approved this study. We developed a MSK-driven lower limb model with 3-dimensional iliotibial band geometry and 3D geometries of TFL and Gmax based MRI data, which could simulate the internal biomechanics of ITB and in-series muscles. The workflow of the whole study included data collection, geometries reconstruction, boundary condition calculation, FE model configuration and calculation, FE model validation, seen in Fig. 1.

6.3.2 Experimental procedures

A motion capture system with eight cameras (Vicon, Oxford Metrics Ltd., Oxford, UK) was used to collect the marker trajectories at a sampling rate of 250 Hz and four force plates (AMTI, Watertown, USA) were used to record the ground reaction force (GRF) at a sampling rate of 1000 Hz for the participant. Reflective markers were attached to the subject. The marker set was listed in Table 12. Marker trajectories and GRF were collected synchronously for running trials under each condition for the participant following the protocols as mentioned before. In addition, the activities of TFL and Gmax were recorded using surface EMG (Delsys Inc., Boston, MA) at the same time.

Table 12 Marker set protocol.

Segment	Marker names	Landmark locations
Trunk	Top.Head R.Acromium L.Acromium	Top of the head Right acromion Left acromion
Pelvis	R.ASIS	Right anterior superior iliac crest
	L.ASIS	Left posterior superior iliac crest
	V.Sacral	Right posterior superior iliac crest
Right thigh	R.Thigh.Upper R.Thigh.Lat R.Thigh.Rear	Triangle-positioned in the right thigh
Right knee	R.Knee.Lat	Right lateral femoral epicondyle
	R.Knee.Med	Right medial femoral epicondyle
Right shank	R.Shank.Upper R.Shank.Lat R.Shank.Med	Triangle-positioned in the right shank
Right ankle	R.Ankle.Med R.Ankle.Lat	Right medial malleolus Right lateral malleolus
Right foot	R.Heel R.Toe.Med R.Toe.Lat R.Toe	Right heel Right medial toe Right lateral toe Right toe
Left thigh	L.Thigh.Upper L.Thigh.Lat L.Thigh.Rear	Triangle-positioned in the left thigh
Left knee	L.Knee.Lat	Left lateral femoral epicondyle
	L.Knee.Med	Left medial femoral epicondyle
Left shank	L.Shank.Upper L.Shank.Lat L.Shank.Med	Triangle-positioned in the left shank
Left ankle	L.Ankle.Med L.Ankle.Lat	Left medial malleolus Left lateral malleolus
Left foot	L.Heel L.Toe.Med L.Toe.Lat L.Toe	Left heel Left medial toe Left lateral toe Left toe

6.3.3 Musculoskeletal multibody model

The joint and muscle forces were calculated in OpenSim. The generic Gait2392 model was scaled to the subject-specific model using the static marker data and anthropometrics of the subject. Inverse kinematics (IK) was performed to get the joint kinematics using the dynamic marker data. The muscle and joints forces were estimated using the Analyze toolkit in the OpenSim and then output as the boundaries and loading conditions in the FE simulation. The general information about the experiments was displayed in Table 13.

Table 13 General gait parameters for all the experimental conditions

	Normal speed pre-exhaustion	Fast speed pre-exhaustion	Normal speed pos-exhaustion	Fast speed pos-exhaustion
Measured speed (m/s)	3.45	4.10	3.48	4.13
Duration of the stance phase (s)	0.24	0.24	0.27	0.21
Peak vertical ground reaction force (BW)	2.41	2.84	2.51	2.96

6.3.4 Finite element model

Geometry and mesh

The MRI data were taken from a healthy male subject using a whole-body 3.0T Siemens Prisma scanner (Seimens, Erlangen, Germany). The configuration of the scanner was T1 sequence, 1-mm slice interval, and a resolution of 0.625 mm pixel size in a neutral position of the subject. Then the segmentation process was performed in Mimics (Materialise, Leuven, Belgium) and further constructed the solid geometries in Geomagic Studio 12 (Geomagic, Morrisville, U.S.) for all the parts (pelvis, femur, tibia, patella, Gmax, ITB, TFL, femoral cartilage, tibial cartilage, meniscus, ACL, PCL, MCL, patellar ligament, and quadriceps tendon). The whole FE model was established in Abaqus 6.14 (Simulia, Dassault Systemes, France). In the model, other necessary ligaments were modeled as truss to connect the bone structures. Major muscles except for Gmax and TFL were modeled as connectors in the model. The bone contact was applied with a frictionless contact with a non-linear contact stiffness (Peng, Niu, et al., 2021). The FE model was displayed in Figure 39. All the bones (pelvis, femur, patella, and tibia) and soft tissues (ligaments, ITB and muscles) were meshed using 4-node linear tetrahedral elements (C3D4) in Abaqus. The mesh type C3D4 could provide details of the geometry and specific insertion sites of the soft tissues. Furthermore, it can also reflect the internal stress and strain. The element size was set as 3mm for bones, 1mm for soft tissues, respectively. Figure 43 also showed the details of the mesh. The materials of ligaments including ACL, PCL, MCL, LCL, quadriceps tendon, and patella ligament were set as isotropic and hyperplastic, which were modelled as Neo-Hookean material (Shao et al., 2022). The details about the material properties were displayed in Table 14 and 15. In addition, three dimensional muscles (Gmax and TFL) in the current study were simplified as a quasi-incompressible, isotropic, hyperplastic material using Neo-Hookean material (C10=0.003 MPa, D=0.14 MPa^{-1}) in the ITB FE model (Lu et al., 2019) under the consideration of efficient computation. The material properties of ITB were set as Young's modulus of 0.4 GPa and Poisson ratio of 0.4 (Hutchinson et al., 2022).

The mesh convergence test was conducted for the ITB FE model with a reduction of element size of 10%. The comparison of peak contact pressure on the medial tibial cartilage was conducted under a load of 1000N (Shriram et al., 2021). The results showed that the mesh size in the current model was acceptable under the criteria of less than 5% deviation (Peng, Niu, et al., 2021).

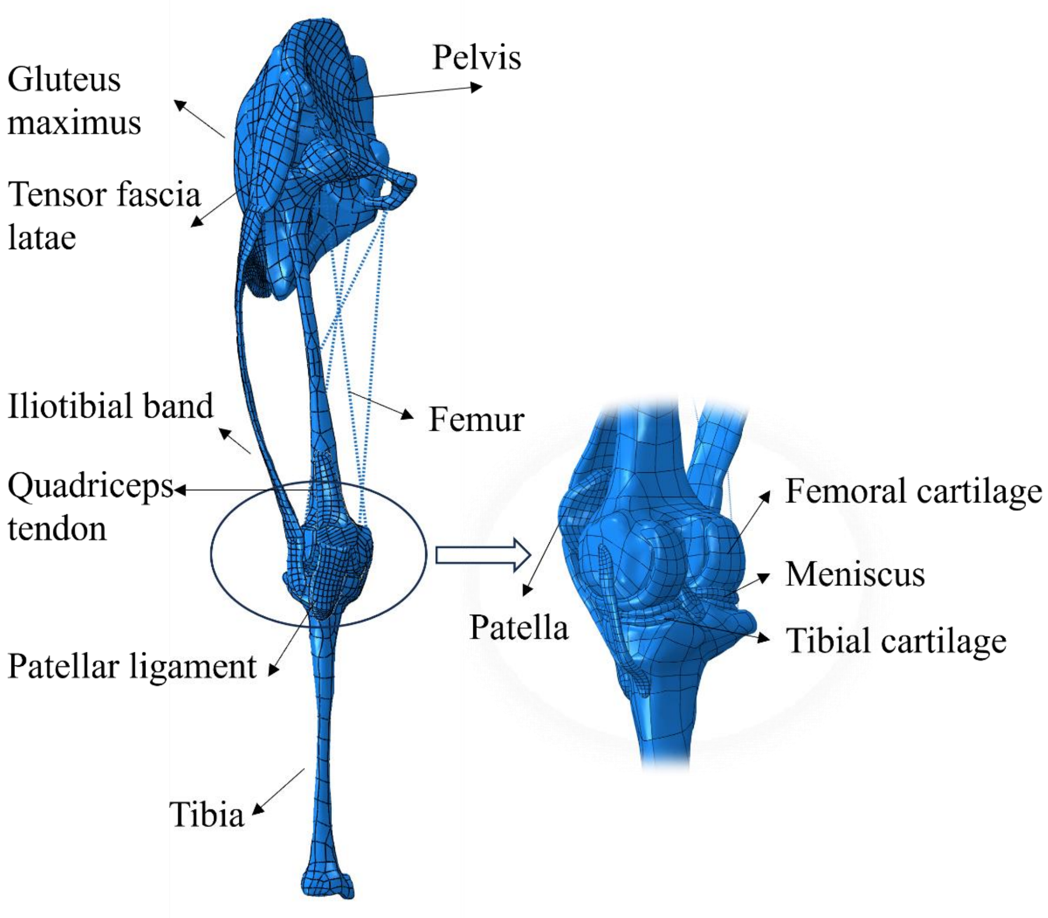


Figure 39 The finite element model of ITB

Table 14 Material parameters of bone, menisci, and cartilage

Items	Elastic modulus (MPa)	Poisson ratio	References
Bone	17,000	0.3	(Bayraktar et al., 2004)
Menisci	E_1=20, E_2=120, E_3=20	ν_{12}=0.3 ν_{13}=0.45 ν_{23}=0.3	(Kiapour et al., 2014)
Cartilage	20	0.45	(Zhu et al., 2019)

Table 15 Material parameters of ligaments in the FE model (Shao et al., 2022)

	C1 (MPa)	D
ACL	1.95	0.00683
PCL	3.25	0.0041
MCL	1.44	0.00126
LCL	1.44	0.00683
Quadriceps tendon	2.75	0.00484
Patella ligament	3.25	0.0041

In the ITB FE model, the contact between femoral cartilage and tibial cartilage, femoral cartilage and menisci, menisci and tibial cartilage geometries were set as friction-free contact with minor sliding. The attachments of ligament-bone and cartilage-bone were confirmed using MRI data and bonded the nodes on the relative surfaces. The bone-to bone interaction was assumed to be frictionless with non-linear contact.

Boundary and loading conditions

The pelvis was fixed in all the translational and rotational degrees of freedom. A reference point located in the central position of distal tibia was coupled to the tibial surface using constraint method in Abaqus. The global coordinate system was set up in accordance with OpenSim. The major lower limb muscles were applied using slipring connectors in the ITB FE model. Figure 40 shows the simulated muscle forces. The joint forces of the ankle joint calculated from OpenSim was applied to the reference point, which related to the stance phase load for all the experimental conditions. All the input data was imported in tabulated time-series matrix from the musculoskeletal model.

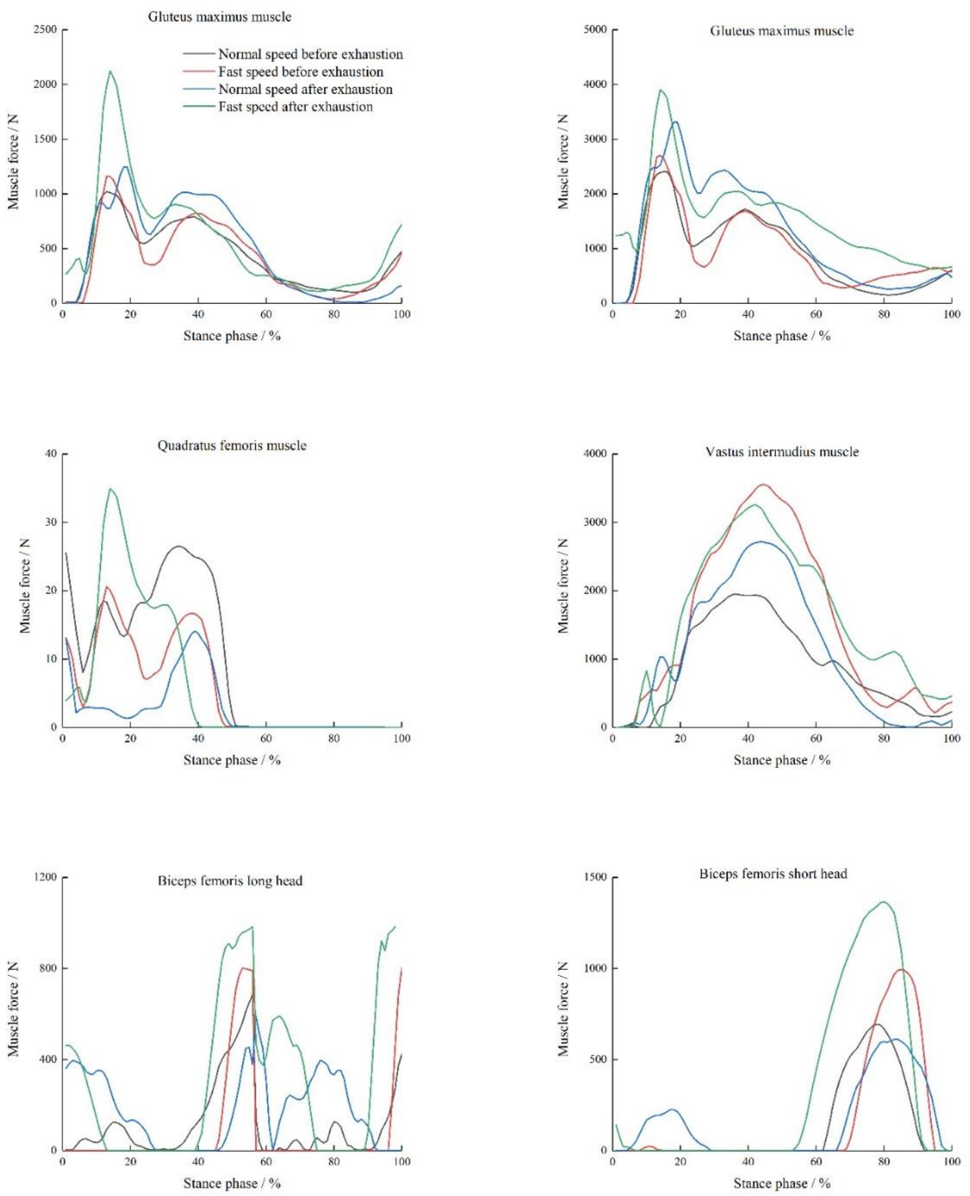


Figure 40 The simulated muscle forces from OpenSim

6.4 Results

The ITB FE model simulated the contact pressures between the lateral femoral epicondyle and the ITB. High contact pressure might be applied to the contact area during dynamic activities, which could cause pain. The contact pressures would be proposed to play an important role in the development of ITBS that is a common overuse injury for runners. The running speed conditions before and after exhaustion remained similar. When the running speed increased, the GRF increased. Additionally, the muscles force calculated in the OpenSim also increased due to the increase in running speed. The GRF increased by about 17% when the running speed increased by about 18%. Figure 41 A-D showed the contact pressure distribution at the lateral femoral epicondyle for all the conditions (normal speed pre-exhaustion, fast speed pre-exhaustion, normal speed pos-exhaustion, and fast speed pos-exhaustion). Three different conditions with different knee flexion angles were selected to discuss the results. The ‘pain area’ when the knee flexion angle is at $20 - 30°$ was proposed in previous studies. The peak pressure values obtained from Figure 41 were plotted in Figure 42.

The principal stresses on the ACL were extracted for all the same conditions as shown in Figure 43. Additionally, the peak stress in the ACL was plotted in Figure 44. When the running speed increases, the peak stress showed an increase for both pre-exhaustion and post-exhaustion. When the running speed remains similar, the exhaustion state also contributes to the increase of the stress in the ACL.

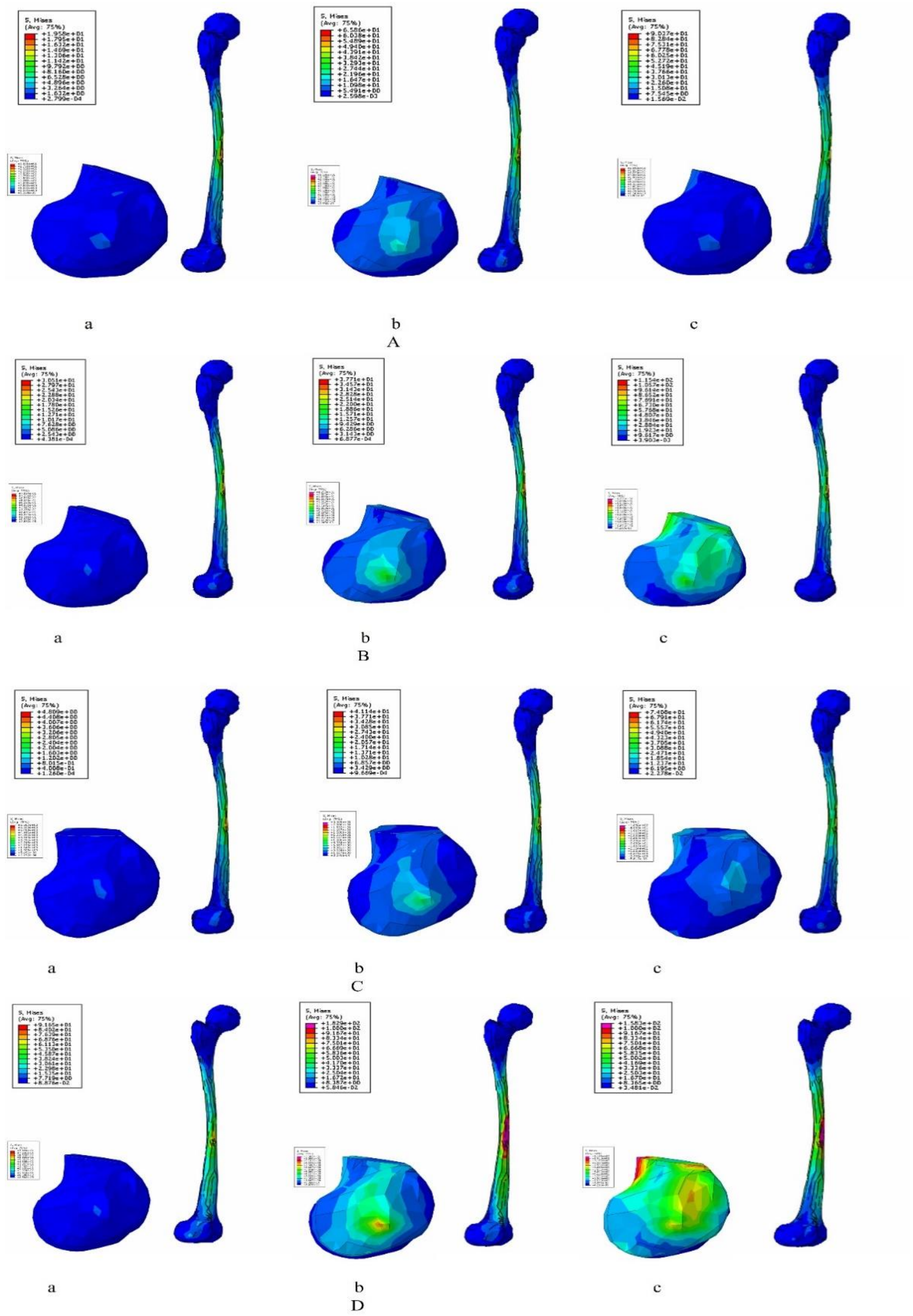


Figure 41 Von Mises stress in the lateral femoral epicondyle. A: normal speed before exhaustion; B: fast speed before exhaustion; C: normal speed after exhaustion; D: fast speed after exhaustion; a: at 10% of the stance phase; b: maximum compression stress; c: at 90% of the stan phase.

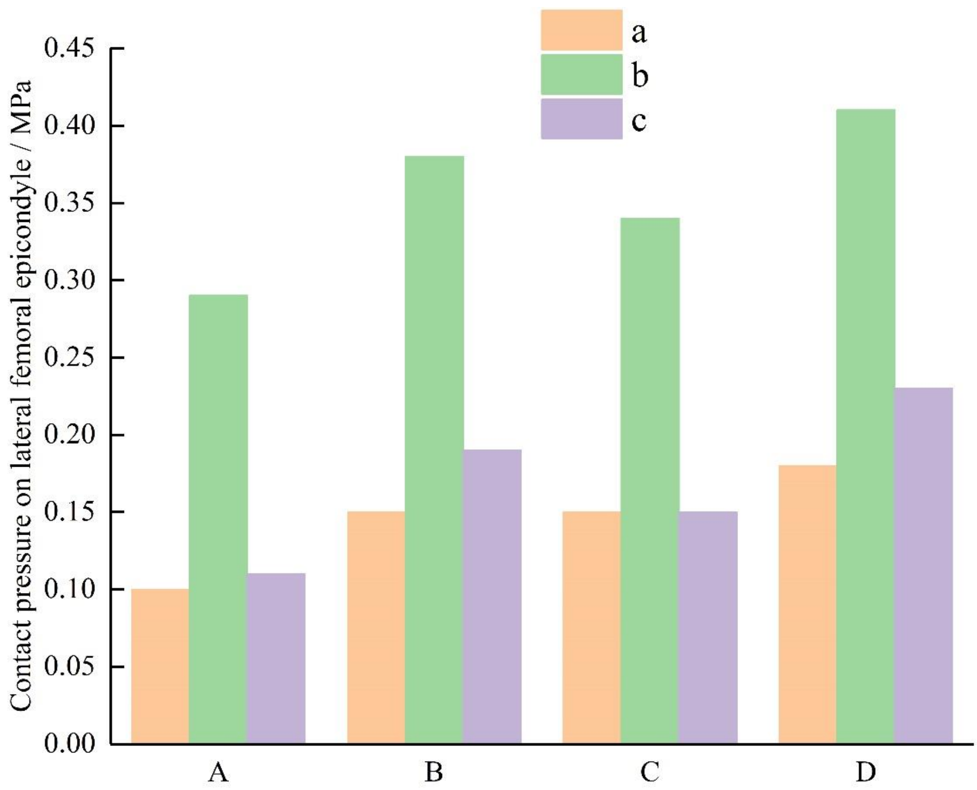


Figure 42 The peak compressive pressure in the lateral femoral epicondyle applied by ITB. A: normal speed before exhaustion; B: fast speed before exhaustion; C: normal speed after exhaustion; D: fast speed after exhaustion; a: at 10% of the stance phase; b: maximum compression stress; c: at 90% of the stan phase.

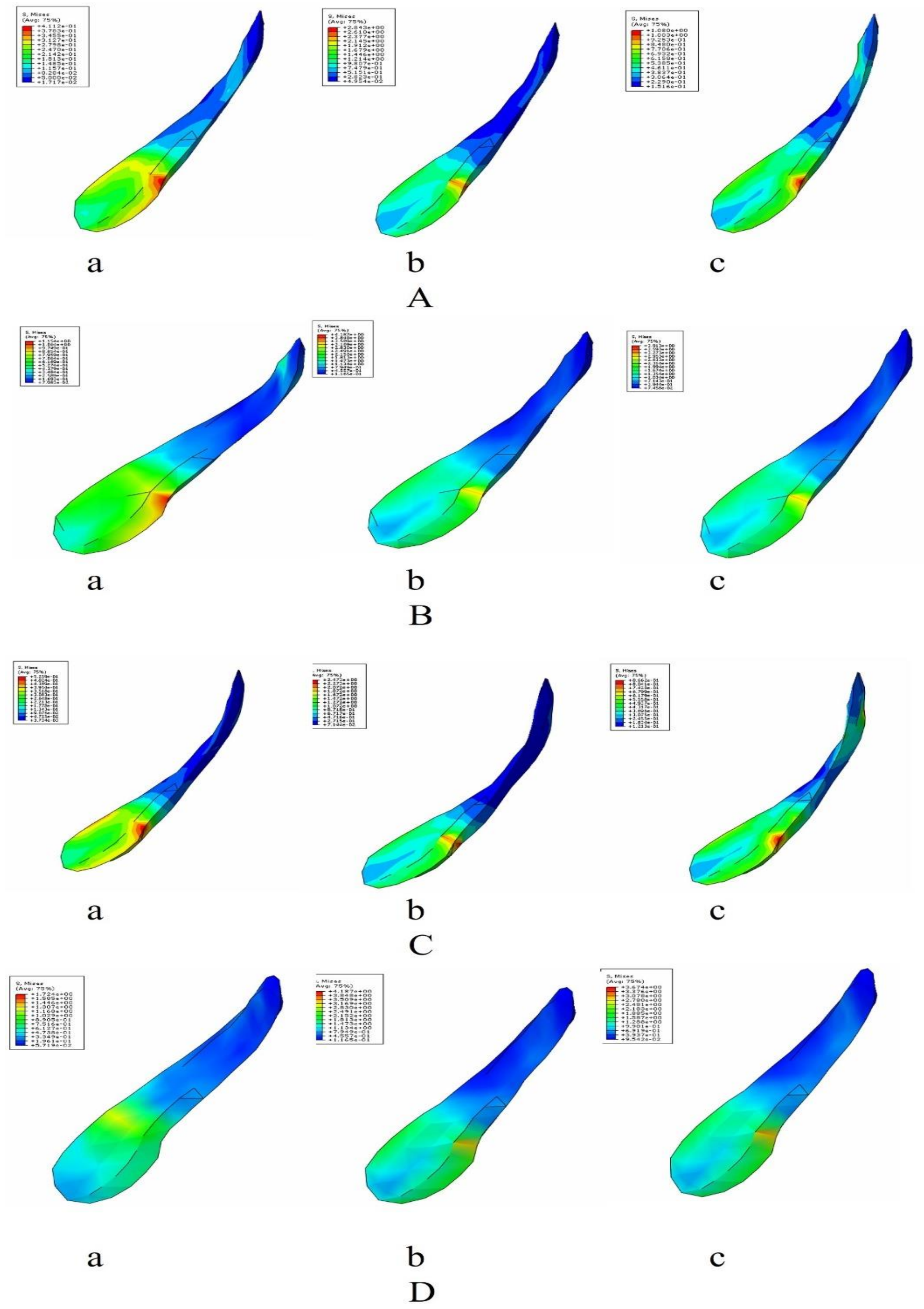


Figure 43 Von Mises stress in ACL during running. A: normal speed before exhaustion; B: fast speed before exhaustion; C: normal speed after exhaustion; D: fast speed after exhaustion; a: at 10% of the stance phase; b: maximum compression stress; c: at 90% of the stan phase.

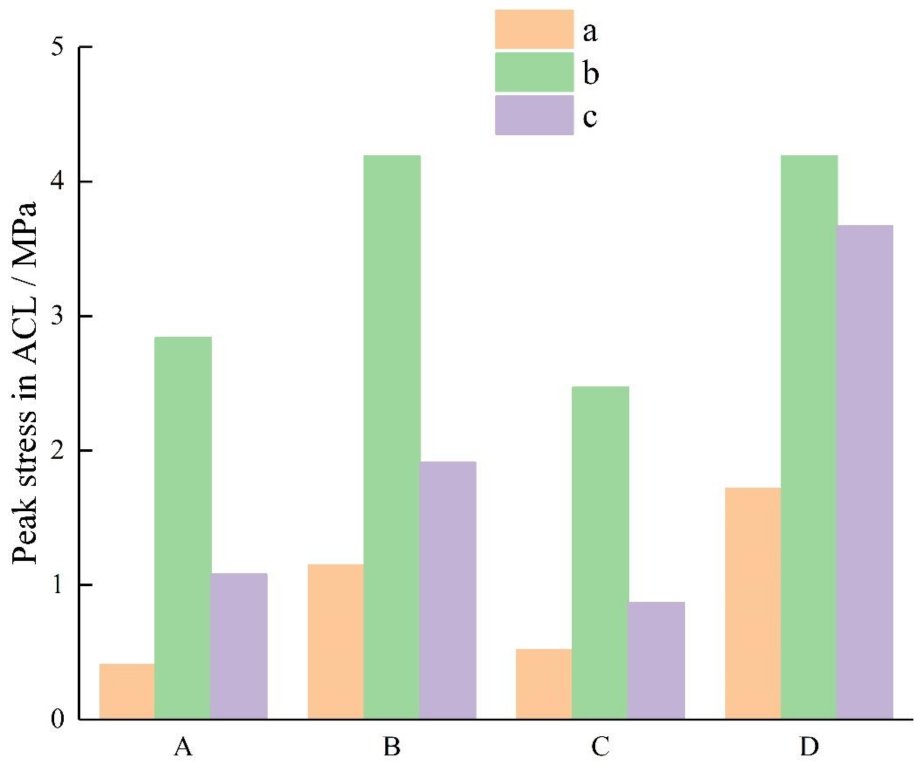


Figure 44 Peak stress in ACL. A: normal speed before exhaustion; B: fast speed before exhaustion; C: normal speed after exhaustion; D: fast speed after exhaustion; a: at 10% of the stance phase; b: maximum compression stress; c: at 90% of the stan phase.

6.5 Discussion

In the ITB FE model, the compressive force and ACL stress have been examined for the four running states including normal speed pre-exhaustion, fast speed pre-exhaustion, normal speed post-exhaustion, and fast speed post-exhaustion. We compared the compressive force that is between ITB and lateral femoral epicondyle at 10%, 90% of the stance phase, and also the peak values. The peak values of the compressive force usually happened at the $20-30°$ knee flexion angle. We also examined the peak stress in the ACL for the four different running conditions. The peak stress in the ACL also showed a higher value when the knee flexion angle is at

$20 - 30°$ than those at 10% and 90% of the stance phase.

There is an impingement zone that is defined to be $20 - 30°$ of knee flexion angle for ITBS patients, which occurs when the distal ITB compresses against the lateral femoral epicondyle (Hamill et al., 2008). The ITB stretches from the hip down to the knee along the thigh. Its position in relation to the femur varies as the knee joint moves through flexion. In the fully extended knee position, the ITB is positioned anteriorly to the lateral femoral epicondyle. However, as the knee flexes, this position rapidly changes. As a matter of fact, ITB lies behind or on the lateral femoral epicondyle when the knee flexion angle is near 30°. This flexion angle approximates the knee joint angle at the early stance of running. Runners usually feel pain sharply following the heel strike. In the current, we explored the peak compressive pressures during running trials. The highest compressive pressures were observed when the knee flexion angle was approximately 30°. The compressive pressure applied to lateral femoral epicondyle from ITB went increase sharply after the heel strike to the apex near 30° knee flexion angle. After that, the compressive pressure reduced along with the knee flexion angle. The highest compressive pressure occurred when the runners were at exhaustion state with a fast speed. With the self-comfortable speed, a higher compressive pressure was also found for runners at post-exhaustion state. The excessive compressive pressure between ITB and the lateral femoral epicondyle was proposed to be one main factor for the common overuse injury ITBS (Louw & Deary, 2014). When the runners experienced high intensity training such as fast speed or exhaustion state, the compressive pressure between ITB and lateral femoral epicondyle would increase suddenly than the runners were at ‘normal’ state. This excessive compressive pressure might be one compensatory force for the knee stability through tensioning of ITB

(Hutchinson et al., 2022).

The ITB functions to reduce the anterior subluxation of tibia when the knee joint is flexed. With the absent of ITB, the lateral tibial plateau might be anteriorly subluxated, which is so called pivot shift subluxation (Yamamoto et al., 2006). The ITB may limit the internal rotation of the tibia. Excessive internal rotation of the tibia might result in ACL injuries during stance phase. ACL functions as the primary stabilizer for the knee joint when the knee flexion angle is between 0° and 30°. Afterward, ITB contributes to the stabilization of the knee joint as the secondary stabilizer when the knee flexion angle is over 30° (Kaplan & Jazrawi, 2018). The compressive force on the lateral femoral epicondyle applied by the IBT reach to the apex to provide restraint for the knee joint around knee flexion angle of 30° when the ACL has also passed over its primary restraint stage as the knee joint stabilizer. As the running speed increased, the peak compressive force on the lateral femoral epicondyle also rose. A previous study concluded that ITB stores much more elastic energy during fast running (Eng et al., 2015). An argument can be made that ITB applies increased compressive force on the lateral femoral epicondyle to contribute to internal rotational stabilization of the knee joint. ITBS as a prevalent overuse injury usually caused by repetitive flexion and extension movements of the knee (Ellis et al., 2007). As the runners stayed in an exhaustive state, the compressive force also tended to increase which might predispose a runner to ITBS. A previous study illustrated that the knee flexion angle at the initial contact would increase as the running time prolonged (Tian et al., 2020). The higher knee flexion angle at initial contact might indicate that runners experiencing higher compressive force more likely occur when they are under exhaustive state. Previous also indicated that runners with ITBS have larger minimum knee flexion angle, which suggest a closer knee position to the impingement zone for them (Miller et al., 2007). Knee joints under exhaustive state tended to be more flexed, predisposing to ITBS. Except for lateral hamstrings that can restrict the internal tibial rotation to protect the ACL from excessive strain, the tensor fascia latae muscle and gluteus maximus muscle through also acts an external rotator of tibia through the insertions into ITB (Cibulka & Bennett, 2020). Thus, the weakness of these muscles might have limited ability to restrict internal rotation of tibia, thereby increasing the compression force in the lateral femoral epicondyle applied by ITB to provide more stability for the knee joint. The biomechanics of ITB was discussed during running to provide more insights into the ITB-related injuries.

This study has some limitations. First, the material properties of the bone and soft tissues were referred to in previous literature, which might have some negative influence on the simulation accuracy. Second, the ITB finite element model was established based on a single subject, which impairs the generalization of the results. Additionally, we simplified the muscles to achieve more efficient computation in the model. The assumption might underestimate the knee joint stiffness in the model although such simplification was common for finite element model of the knee joint.

6.6 Conclusion

In this study, we aimed to investigate the biomechanics of ITB during running by developing a subject-specific musculoskeletal-driven FE model. The subject-specific model allowed us to capture individual variations in anatomy and mechanics, providing more accurate and personalized results. To accurately simulate the loading conditions on ITB, we utilized a musculoskeletal model to obtain the dynamic muscle forces and joint moments during running. These loading boundaries were then applied to drive the FE model of ITB. By coupling the musculoskeletal and FE models, we were able to simulate the biomechanics of ITB. We found that the compressive pressure on the lateral epicondyle reach to the peak value when the knee flexion angle is around 30°. The running speed is one main factor that could increase the pressure. Additionally, the exhaustion state would also increase the compressive pressure.

By employing this subject-specific approach and considering the dynamic loading conditions, our study contributes to a better understanding of the biomechanics of ITB during running. These findings can have implications for injury prevention, rehabilitation strategies, and the development of personalized interventions for individuals with ITB-related issues.

CHAPTER 7 CONCLUSION AND SUGGESTIONS FOR FUTURE RESEARCH

7.1 Significance of the study

The ITB plays a crucial role in the biomechanics of running, contributing to joint stability, load transmission, and energy efficiency. Investigating the biomechanics of ITB during running can provide insights into its function and contribution to overall running performance. This knowledge can be used to develop training strategies that optimize ITB function and improve running performance. Dysfunction of ITB can contribute to movement disorders and altered lower limb kinematics. By investigating the biomechanics of ITB, researchers can gain insights into its role in movement disorders and develop targeted interventions to address these issues. Gaining insights into the biomechanics of ITB during running can aid in identifying risk factors associated with ITB-related injuries, such as ITBS. By identifying these risk factors, clinicians and coaches can develop targeted prevention strategies and interventions to reduce the incidence of ITB injuries among runners. For individuals with ITBS, knowledge of the biomechanics of ITB can guide the development of effective rehabilitation protocols. By addressing the underlying mechanical factors contributing to ITBS, rehabilitation programs can be tailored to target specific deficiencies and promote optimal recovery.

Studying the biomechanics of ITB during running is vital for injury prevention, rehabilitation, performance optimization, and understanding movement disorders. It provides valuable insights into the mechanical behaviour of ITB and its contribution to lower limb function, leading to improved interventions and better outcomes for individuals involved in running activities.

7.2 Conclusions

The current study aimed to investigate the biomechanics of ITB during running and its implications for ITBS. To understand the biomechanics of ITB, the study employed a subject-specific approach and utilized musculoskeletal and finite element modelling techniques. The research focused on three main aspects: the effects of in-series musculature on the behaviour of ITB during an exhaustive run, the influence of exhaustion state and running speed on ITB strain and strain rate, and the interaction between ITB and the lateral femoral epicondyle using a finite element model.

The findings of the study revealed several important insights. Firstly, the behavior of the healthy ITB was found to be influenced by the activities of in-series musculature, leading to altered compression forces applied to the lateral femoral epicondyle during exhaustive running. Secondly, exhaustion state and running speed were identified as significant factors affecting ITB strain rate, emphasizing the importance of considering these factors in the prevention and treatment of ITBS. Finally, the study highlighted the impingement zone between ITB and lateral femoral epicondyle, with peak compressive force and ACL stress occurring at a knee flexion angle of $20 - 30$ degrees.

These findings have significant implications for injury prevention, rehabilitation strategies, and the development of personalized interventions for individuals with ITB-related issues. By understanding the biomechanics of ITB and its interaction with other factors such as muscle activation, exhaustion state, and running speed, tailored approaches can be developed to optimize performance and reduce the risk of ITB-related injuries. The subject-specific modelling approach employed in this study enhances the understanding of individual variability and provides valuable insights that can guide the customization of interventions for better outcomes.

In conclusion, the study sheds light on the biomechanics of ITB during running, offering important insights into ITB-related injuries and their prevention. By considering the complex interplay between various factors, this research contributes to a better understanding of ITB's role in knee joint stability and provides a foundation for developing targeted strategies to improve performance and reduce the risk of ITBS.

7.3 Future Direction

The current study has discussed the biomechanics of ITB during different running conditions. In the future, incorporating advanced imaging techniques such as ultrasound can provide more detailed information on ITB structure, morphology, and function. This can enhance our understanding of ITB mechanics and aid in the development of more accurate computational models. Studying the dynamic behaviour of ITB during various functional tasks, such as cycling, jumping, and cutting, can provide insights into how it responds to different loading conditions and movements. Dynamic analysis can help identify specific phases of movement where ITB is subjected to higher loads or exhibits altered mechanics. Since ITB in the current study simplified ITB in the FE model, future studies could integrate multi-scale modelling approaches that consider the hierarchical structure of ITB, which can provide a comprehensive understanding of its mechanical properties and behaviours. Overuse injuries such as ITBS is a long-term sport injury. Conducting longitudinal studies that follow individuals over an extended period can provide insights into the changes in ITB biomechanics over time. It can help identify risk factors for ITB-related injuries, track the effectiveness of interventions, and understand the natural progression of ITB biomechanics with aging or training. Additionally, future studies could also focus on investigating the effectiveness of various intervention strategies, such as strength training, stretching, footwear modifications, or gait retraining, can contribute to evidence-based recommendations for managing and preventing ITB-related injuries. By focusing on these directions, researchers can deepen their understanding of ITB biomechanics, leading to improved injury prevention strategies, more effective rehabilitation protocols, and enhanced athletic performance for individuals involved in running activities.

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