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GAIT ANALYSIS RELATED TO RUNNING INJURIES

ANALYSE QUANTIFIEE DE LA FOULEE EN RAPPORT AVEC LES PATHOLOGIES DU SPORT

Thèse dirigée par **PR REDHA TAIAR**

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Analyse quantifiée de la foulée en rapport avec les pathologies du sport

L’objectif de cette thèse était de déterminer l'effet de la fatigue sur l'atténuation des ondes de choc d'impact et d'évaluer la relation entre la biomécanique humaine et l'atténuation des chocs pendant la course. Dans cette thèse, nous proposons une nouvelle méthodologie pour l'analyse des événements de choc survenant au cours de la procédure expérimentale proposée. Notre approche est basée sur le spectre de réponse aux chocs (SRS), qui est une fonction basée sur la fréquence utilisée pour indiquer l'amplitude des vibrations dues à un choc ou à un événement transitoire Nous avons utilisé des technologies embarquées tel que les centrales inertielles (IMU) (RunScribe®, San Francisco, CA, USA) pour notre expérimentation.

Les blessures de surmenage en course à pied sont souvent provoquées par la fatigue ou une mauvaise technique, qui se reflètent toutes deux dans la cinématique du coureur. La recherche de pointe sur la cinétique et la cinématique dans le sport utilise des systèmes d'analyse de mouvement qui sont inaccessibles à la plupart des athlètes. Le potentiel des capteurs embarqués pour l'analyse cinétique et cinématique des coureurs est extrêmement pertinent et rentable. Tout au long de nos recherches, nous avons démontré le potentiel des capteurs portables pour l'analyse cinétique et cinématique des coureurs. Nous présentons plusieurs études utilisant des centrales inertielles (IMU) pour l'évaluation du niveau de performance et surveillance de la fatigue. Nous avons extrait de nombreux paramètres de foulée pour les évaluations de performance et de santé. Les capteurs embarqués constituent un outil précieux pour les coureurs, des débutants aux experts, pour l'évaluation de la technique de course.

Notre hypothèse est que la fatigue entraîne une diminution de la capacité d'atténuation des chocs du système musculosquelettique, impliquant ainsi potentiellement un risque plus élevé de blessure due au surmenage.

Mots-clés : Analyse quantifiée de la foulée, centrale inertielle (IMU), fatigue, spectre de réponse au choc, blessure

Gait analysis related to running injuries

The objective of this thesis was to determine the effect of fatigue on impact shock wave attenuation and assess how human biomechanics relate to shock attenuation during running. In this paper, we propose a new methodology for the analysis of shock events occurring during the proposed experimental procedure. Our approach is based on the Shock Response Spectrum (SRS), which is a frequency-based function that is used to indicate the magnitude of vibration due to a shock or a transient event. Five high level CrossFit athletes who ran at least three times per week and who were free from musculoskeletal injury volunteered to take part in this study. Two Micromachined Microelectromechanical Systems (MEMS) accelerometers (RunScribe®, San Francisco, CA, USA) were used for this experiment.

Injuries in running are often provoked by fatigue or improper technique, which are both reflected in the runner’s kinematics. State of the art research on kinetics and kinematics in sports is using motion analysis systems that are inaccessible to most athletes. The potential of wearable sensors for runners’ kinetic and kinematics analysis is extremely relevant and cost effective. Throughout our research we demonstrate the potential of wearable sensors for runners’ kinetic and kinematics analysis. We present several studies using inertial measurement units (IMU) for performance level assessment, training assistance, and fatigue monitoring. We extracted many gait parameters for performance and health assessments. Wearable sensors provide a valuable tool for runners, from beginners to experts, for running technique assessment.

Our hypothesis is that fatigue leads to a decrease in the shock attenuation capacity of the musculoskeletal system, thus potentially implying a higher risk of overuse injury.

Keywords: Gait analysis, micro-electro-mechanical systems (MEMS), fatigue, shock response spectrum, injury

Discipline : Biomécanique

Spécialité : Biomécanique humaine

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First and foremost, I wish to sincerely thank Professor TAIAR Redha who gave me this great opportunity throughout my PhD to combine podiatry and human biomechanics, this will hopefully lead to a whole new level of understanding of the human posture and gait. His valuable guidance and advice have inspired me greatly to work on this project. His willingness and motivation have pushed me to fully understand and master the field of running biomechanics.

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A very special thought goes to Mr GUAY Laurent, Mr BOYER Cyril and Mr CHISTIAEN Benoit for their help in recruiting patients and collecting data. Your encouragements on my research as well as on my career have been priceless.

I also send my kind regards to “RunScribe®” and “HIKOB®” for kindly lending us the microelectromechanical systems (MEMS). No financial nor any kind off compensation were involved and the MEMS were returned after the experimentation.

I'd also like to thank all of my friends who encouraged me to keep writing and pushed me to achieve my goal. A kind thought goes out to my dear wife Clothilde, who has spent many restless nights with me and has always been my pillar of strength when no one else could answer my questions.

LIST OF PUBLICATIONS AND COMMUNICATIONS

This thesis is based on 3 conference abstracts that have been published in indexed scientific journals:

Appendix 1
Preliminary comparison between angular velocity and force measuring treadmill for running foot strike patterns classification
October 2019
Computer Methods in Biomechanics and Biomedical Engineering, Vol. 22(sup1): S484-S486
DOI: 10.1080/10255842.2020.1714990

Appendix 2
Harmonic decomposition and analysis of running gait
October 2019
Computer Methods in Biomechanics and Biomedical Engineering, Vol. 22(sup1): S343-S344
DOI: https://doi.org/10.1080/10255842.2020.1714937

Appendix 3
Shock response spectrum analysis in running performance
November 2020
Computer Methods in Biomechanics and Biomedical Engineering, Vol. 23(sup1): S28-S30
DOI: https://doi.org/10.1080/10255842.2020.1811500

And on one original paper published in an indexed scientific journal:

Appendix 4
Shock Response Spectrum Analysis of Fatigued Runners
March 2022
Sensors, Vol. 22, Issue 6: 2350.
DOI: https://doi.org/10.3390/s22062350

LIST OF FIGURES

CHAPTER 1

General introduction

In this chapter we will cover my personal motivation and dedication to this PhD and the research that came with it. We will go over the challenges tackled throughout our research and the motivation behind it. We will also talk about the previous contribution podiatrists have made throughout time to the world of biomechanics. I will also walk you through the history of gait analysis and athletic footwear. In the last section we will detail my thesis' structure.

1. Motivation

Running is by far one of the most popular physical activities in the world. It is probably the most universal and easy sport to practice at all ages and pretty much anywhere. Unfortunately, running can come with a wealth of problems such as musculoskeletal injuries. Therefore, the need to study injuries and properly understand how they occur becomes very important.

I have been practicing as a sport Musculoskeletal (MSK) podiatrist for over a decade in Paris providing clinical assessments, gait analysis and custom-made orthotics for my patients. A very large part of my professional activity is dedicated to taking care of injured runners. I can honestly say that I have seen a very broad spectrum of runners from the inconsistent recreational runner to the elite ultra-runner, the have come to me in all shapes, sizes and ages.

From the beginning of my career, I was lucky enough to be able to invest in quality gait analysis tools. Our gait analysis tools range from gold standard instrumented treadmills, to light-optical scanning systems based on Video-Raster-Stereography (VRS), to optometric systems, to video analysis systems and all the way to connected wearables for outdoor measurements such as inertial measurements (IMU) microelectromechanical systems (MEMs). Having access to such technologies has enabled me to collect a lot of data on a wide spectrum of runners throughout my career. Very quickly I came to realize that my sole podiatry training would not be enough for me to be able to decrypt and analyze these data clusters properly and put them to good use. I have also been using the latest CAD/CAM technologies to manufacture orthotics such as 3D milling and 3D printing to offer the best quality treatment for patients.

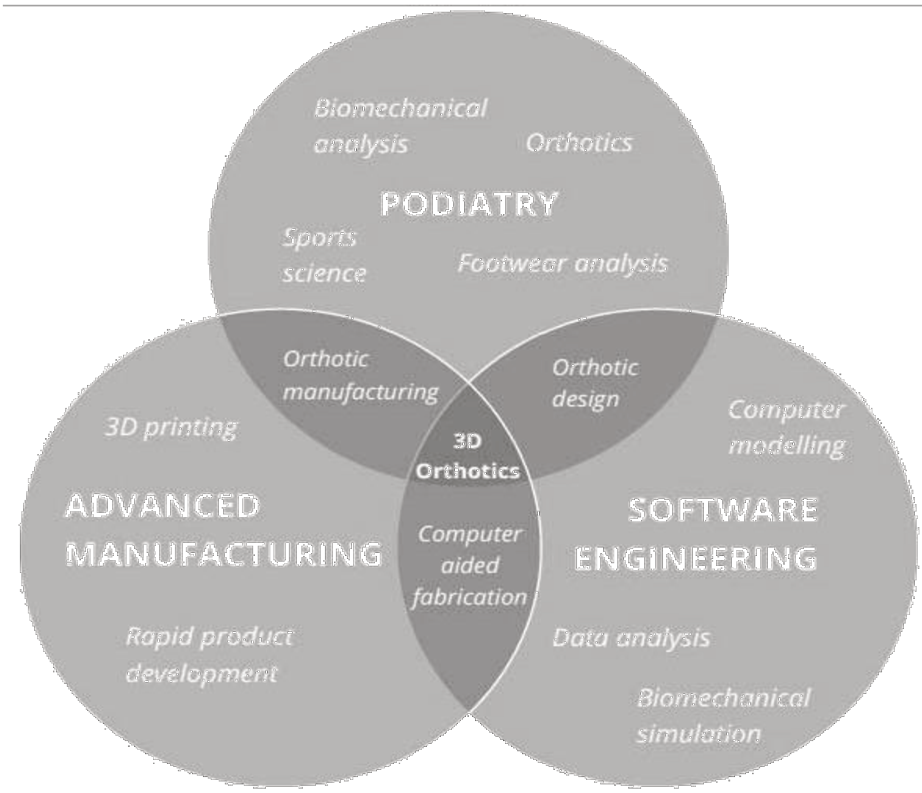


Figure 1: Modern sports podiatry skillset

The CAD/CAM manufacturing method enables reproducibility and repeatability of the foot orthoses treatment. A modern musculoskeletal sports podiatrist needs nowadays to master a larger number of skillsets as shown in figure 1.

This is when I met Pr Redha Taiar who was and still remains a reference in the world of biomechanics. I first met him through a university training for podiatrists that he was in charge of, this is also where I met Dr François Fourchet who gave us a class on running biomechanics and athlete physical rehabilitation. Meeting these two specialists has been the cornerstone of my motivation to pursue my research in the field of biomechanics and podiatry. Pr Taiar has then opened the doors for me to pursue my work throughout a master's degree at the university of Reims Champagne-Ardennes that led to my PhD.

During my PhD I was then incredibly lucky to meet Mr Serge Odoff and Mr Boussad Abbes who have mentored me in the field of physical mechanics, their perception and contribution to my research are invaluable and has pushed this thesis to a whole new level.

Their different knowledge and background have opened my eyes on a wealth of issues and solutions that I never thought of. Thanks to the hard work and dedication of this team we have managed to come up with novel and elegant approach to the field of running biomechanics. Their guidance has enabled me to rise up to the task and tackle the challenge of analyzing and interpreting my very large running data bank.

The pinnacle of my PhD has been to offer a new methodology for the analysis of shock events occurring during the proposed experimental procedure. Our approach is based on the Shock Response Spectrum (SRS), which is a frequency-based function that is used to indicate the magnitude of vibration due to a shock or a transient event.

The primary objective is to examine the capacity of the human musculoskeletal system to attenuate the mechanical stresses caused by the fatigue effect, as measured by the Shock Responses Spectrum (SRS) of the foot strike-generated shock waves, while jogging. The use of SRS as a measurement in running gait analysis has not yet been investigated. This novel methodology could open the way for an entirely new method of analyzing the gait pattern of a runner utilizing smart connected shoes.

2. Challenges

The advent of connected wearables and, more specifically, "Smart Wearables" (SW) results in a quantitative explosion of data that introduces new information for analysis and interpretation [1]. Numerous experts/large institutions (such as MIT in the United States), administrations, and specialists in the subject of biomechanics have considered addressing the scientific and technology problems provided by the phenomena of "Big Data" one of their top goals for the decade. In podiatry, connected wearables for outdoor measurements [2] such as inertial measurements (IMU), microelectromechanical systems (MEMs), or plantar pressure measurement are utilized in a variety of clinical settings, including post-operative follow-up, design of orthotics, gait rehabilitation, assistance with preoperative decisions, follow-up of diabetic patients, and evaluation of foot surgery. Simultaneously, it is utilized in the laboratory to study the phenomena that regulate walking, running, and human posture.

Numerous instruments, such as connected wearables for outdoor measurements such as inertial measurements (IMU), microelectromechanical systems (MEMs), or insoles connected to plantar pressure analysis, simplify data gathering and real-world analysis (figure 2). This sort of gadget enables the monitoring of patients or athletes as well as the monitoring of racing parameters, which opens up a huge array of options for the avoidance of diseases, the enhancement of performance, the rehabilitation of patients, and the implementation of orthotics. The quantification of abnormal running gait patterns is a significant obstacle for the adoption of preventative measures and predictive sports medicine. Recent study on various styles of running has focused on the distinctions between heel and mid/forefoot foot strikes in terms of potential injury risks and shock absorption capacities.

Figure 2: IMU and wireless connected pressure insoles

According to contentious and arbitrary estimates, heel striking is frequently associated with higher injury rates due to damage [3] and impact because of insufficient shock absorption and ineffective biomechanical compensations for these forces. This is owing to the fact that blows from heel strikes are absorbed by bones rather than muscles, which are responsible for shock absorption. Because bones cannot easily distribute stresses, the forces are transmitted to other areas of the body, including ligaments, joints, and bones in the remainder of the lower limb up to the lower back. To prevent severe bone damage, the body employs abnormal compensatory mechanisms.

Examples of compensations include the internal rotation of the tibia, knee, and hip joints. Over time, excessive compensation has been linked to an increased risk of damage in the joints and muscles involved in these actions. The triceps surae is used as a lever system to absorb stresses eccentrically rather than through the bone with a mid/forefoot stroke [4]. Landing with a mid/forefoot strike not only properly attenuates shock after extending to absorb ground contact loads, but also permits the triceps surae to contribute to propulsion via reflexive plantarflexion. Therefore, a mid/forefoot hit may aid in propulsion [5].

There are, however, variances in self-selected footstrike techniques even among elite athletes. This is especially true for longer distance races, in which heel striking is prevalent. In competitive fields, however, there is a greater proportion of mid/forefoot strikers, particularly among quicker runners and winning individuals or teams. While physiological differences may account for the faster speeds of elite runners compared to those of recreational runners with comparable footstrikes, the hip and joints have been left out of the equation for proper propulsion. This begs the question of how elite distance runners with possibly inefficient and harmful foot strike techniques are able to maintain such high speeds.

Estimating activity-related traits is a common topic of contemporary research. Particular attention has been paid to the evaluation of temporal and spatial gait parameters [6], the identification of gait phases, and the classification of different gait activities (such as walking, jogging, and stair climbing). Running is characterized by frequent foot contact with the ground.

These collisions are distinguished by a transient peak in the ground response force (impact force), rapid deceleration of the lower extremities (impact shock), and the initiation of a propagating wave of acceleration and deceleration (impact shock wave). Repetitive impact loads have been linked to degenerative joint diseases and overuse injuries in athletes, including stress fractures, shin splints, osteoarthritis, and lower back discomfort [7]. The connection between impact and injury is well-documented, despite the fact that the particular causes of impact-related harm are mostly unknown and debatable.

3. Related work

During the past two decades, gait analysis has been transformed from a purely academic discipline to a useful tool in the hands of physicians and therapists.

People have been fascinated by the movements involved in walking throughout history. The classical Greek and Roman painters and sculptors possessed a grasp of the shape and alignment of the limbs during various occupations, as seen by their paintings and sculptures. During the Renaissance, this understanding was enhanced by human dissection and attempts to comprehend the fundamentals of biomechanics, notably by Leonardo da Vinci, Galileo, Newton, and particularly Borelli according to Whittle (1996) [8]. The Weber brothers in Germany conducted the first serious biomechanical experiments during the eighteenth century. Since then, advancements in four distinct scientific disciplines have contributed to the development of gait analysis. Kinematics, kinetics, electromyography, and engineering mathematics are these four. Garrison (1929), Bresler and Frankel (1950), and Steindler (1970) provided insightful accounts of the early history of gait analysis (1953).

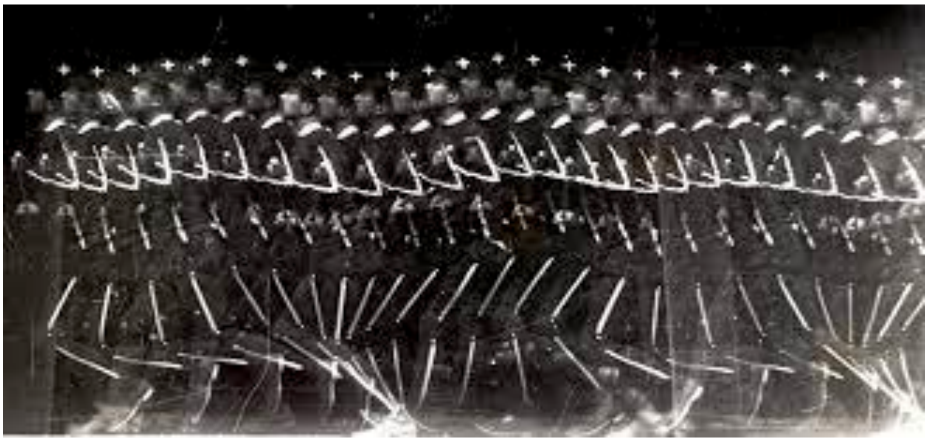


Figure 3: Etienne Jules Marey running motion capture 1886 (La mémoire de l'oeil, Mannoni L, 1999)

The study of how motion is measured is called kinematics. According to Baker (2007) in the 1870s, kinematic research on human walking was first conducted by Marey in Paris (Figure 3), and by Muybridge in California. Both of these researchers were based in the United States. These early researches were carried out with the use of still cameras; the advent of tine photography, which became the primary way for making kinematic measurements until relatively recently, led to significant improvements in precision. A very active group based in California carried out the first significant investigation of the kinematics of gait during the 1940s and 1950s. This research was conducted in California [9].

For the purpose of measuring the positions of the limbs, it was necessary to manually digitize hundreds of photos from a single film, which was then followed by a number of complex calculations. The latter half of the 1970s and the beginning of the 1980s saw the introduction of measurement systems that were based on television cameras. These systems were directly linked into computers, which made the entire process significantly quicker and more convenient. Several countries, like the United States of America, Canada, Italy, and the Netherlands, as well as Scotland and Sweden, have evolved their own system independently. Several of these early systems eventually developed into the machinery that is available for purchase today.

Kinetic measurements have mostly focused on the forces that are acting between the foot and the ground. These forces are monitored by an instrumented area of floor that is known as a force platform. After the first designs were created that were solely mechanical and one-dimensional, subsequent improvements by the late 1970s led to the creation of precise three-dimensional instruments that had an electrical output and a high frequency response. A certain amount of study on the kinetics of gait has also been done on limb accelerations, despite the fact that the majority of research on the kinetics of gait has focused on ground response forces. Based on kinematic and force platform data, as well as the use of engineering mathematics, modern gait analysis systems are able to deliver extra kinetic information to the user in the form of joint moments and joint forces [9].

During the first half of the 20th century, electromyography (EMG), which measures the electrical activity of muscles, came into existence. The Californian group that was described before conducted the first extensive study of the electromyographic activity that occurred during walking during the 1940s and 1950s. Since then, there have been significant advancements in both processes and equipment, and the monitoring of electromyographic activity (EMG) while walking is now a standard component of clinical gait analysis (Perry, 1992) [10].

In the 1890s, mathematics was used for the first time in a significant way to better understand the mechanics of walking. This strategy saw some additional development in Moscow in the 1930s, and a group in California in the 1950s, respectively. Beginning in the 1960s and continuing onward, numerous significant research on the transmission of forces and moments at various joints have been published in academic journals. The majority of modern gait analyses make use of a technique known as "inverse dynamics" to determine joint moments and powers. This technique takes as input data the motion of the limbs (as determined by a kinematic system) and the ground response force (from a force platform). The biomechanical analysis of an individual's gait was possible from the late 1940s until the late 1970s, but it was not very practical because of the complexity of the calculations involved, which were initially performed on slide rules. This period of time spans from the late 1940s until the late 1970s. With the introduction of electronic calculators, calculating became somewhat less difficult, and with the arrival of mainframe computers, it got far less difficult [9].

The routine use of gait analysis in the clinic did not start until the cost, size, and complexity of computers had decreased to the point where it was conceivable to have a dedicated machine within the laboratory. Before this point, gait analysis was not used in the clinic. Minicomputers were employed in the first generation of clinical gait analysis systems, whereas microcomputers are used in the current generation [8].

From the 1940s until the 1970s, there were sporadic reports of the application of gait analysis in planning treatment for individual patients. However, these reports were primarily examples of how the technology might be used, and they came from centers whose primary interest and skill lay in the scientific aspects of the subject matter. Gait analysis had to be put on hold in the clinic until sufficient systems could be made available for normal usage before it could be used routinely there. After this took place in the late 1970s, four orthopedic physicians bore a significant amount of responsibility for the incorporation of gait analysis into ordinary patient care. In the United States, they were Jaquelin Perry [10], David Sutherland, and Jim Gage, while in the United Kingdom, it was Gordon Rose.

Clinicians in the fields of orthopedics, physiotherapy, and podiatry, in particular, have been trying to use biomechanical approaches, such as shoe inserts, to alleviate functional foot and lower limb disorders for more than a century. According to Kirby (2010) [11] in the 1960s and 1970s, Merton Root and his podiatric colleagues were instrumental in providing foot health practitioners, and podiatrists in particular, with a consistent basis for the evaluation and biomechanical treatment of foot and lower extremity pathologies. This was accomplished during this time period. Researchers from the international community of biomechanics started taking a greater interest in the clinical outcome of their investigations as the podiatric community grew more active in biomechanics. In terms of research on the foot and lower extremity biomechanics, it wasn't until the first half of the twentieth century that the medical literature began to suggest that there was a connection between foot mechanics and foot pathology and that it was possible that the mechanics of the human foot were significantly more complicated than what had been thought in the past. This research was conducted in relation to the lower extremity biomechanics research. Merton Root [13] was one of the first researchers to discover the triplanar nature of the subtalar joint axis. He was also one of the first researchers to emphasize how significant the motions of this joint were to the function and dysfunction of the foot and the lower extremities [11].

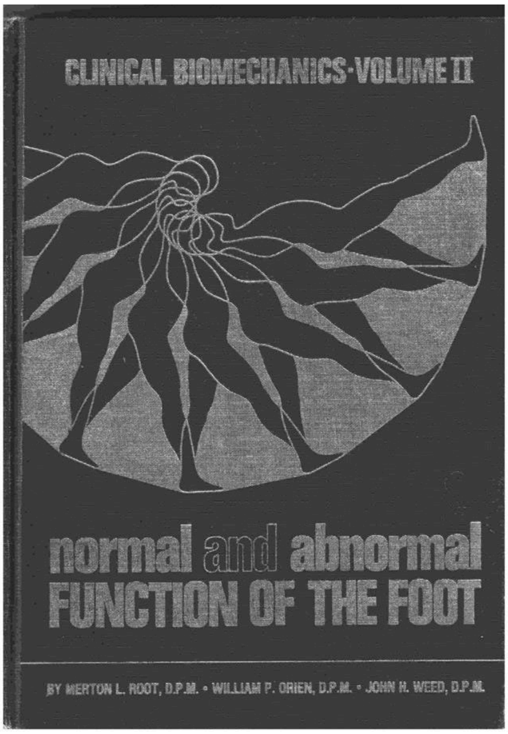


Figure 4: Merton Root's book considered as the father of modern podiatry biomechanics

John Hicks, a prominent foot biomechanics researcher in the 1950s, published a series of papers examining the function of foot joints, the significance of foot muscles, and the mechanisms of the foot in balancing the body's center of mass while standing [12]. However, Hicks' concept of the windlass effect of the plantar fascia at the first metatarsophalangeal joint became his most well-known contribution to foot biomechanics, as it was instrumental in the introduction of the concept of functional hallux limitus as formulated by Dananberg [147]. Although the "Root paradigm" continued to dominate the podiatric scene for the remainder of the twentieth century (Figure 4), new theoretical developments in podiatric biomechanics have supplemented or partially replaced the theories of Root and co-workers [13]. Notable theories include Howard Dananberg's [147] "Sagittal Plane Facilitation Theory," in which the sagittal plane component of foot and lower joint biomechanics, and particularly of the first metatarsophalangeal joint, was emphasized as being crucial for the function of the foot, lower extremity, and lower back during walking.

In 1987, Kirby began to theorize that the spatial location of the subtalar joint axis relative to the plantar foot significantly affected foot function. Kirby's 2001 "Subtalar Joint Axis Location and Rotational Equilibrium Theory" of foot function provides a coherent explanation for the kinetic effects that abnormal subtalar joint spatial location may have on the foot and lower extremity (Kirby 2001). In 1995, Thomas McPoil and Gordon Hunt introduced the concept of designing orthoses using "Tissue Stress Theory" [15], which posits that overuse injuries and biomechanical dysfunction of the foot and lower extremity can be explained and best treated by designing foot orthoses to reduce stress in the injured bone, ligament, tendon, cartilage, or muscle.

In "Tissue Stress Theory," injury is not believed to be caused by excessive pronation or supination motion away from the neutral position of the subtalar joint, as proposed by Root and colleagues, but rather by biomechanical overload of certain structural components of the foot and lower extremity (McPoil and Hunt 1995, p. 15). This approach suggests that any therapy (e.g., orthoses, shoe modifications, and physiotherapy) should aim to modify ground reaction forces and muscle forces in order to maintain tolerable tissue stress levels.

In addition to podiatrists, renowned biomechanical researchers have proposed hypotheses concerning the possible function of the foot and lower extremities. Benno Nigg's "Preferred Movement Pathway Theory" (1999, 2001) [16,17] proposes that foot orthoses do not function by realigning the skeleton, but rather by modifying the input signals into the plantar foot, which in turn alters the "muscle tuning" of the lower extremity foot during athletic activities. Joseph Hamill's "Dynamical systems approach" and his work on lower limbs biomechanics (Hamill et al. 1988, 1999, 2008) [18] have been cited frequently in podiatric discussions, as has Peter Cavanagh's work on the diabetic foot (Cavanagh 2000, 2006) [19]. In addition to the examples listed above, there are additional instances of biomechanics researchers and clinicians combining their individual skills to conduct research that contributed to the foot and lower extremity biomechanics knowledge base.

Lastly, it is impossible to ignore the enormous contribution sports brands have made to the world of biomechanics through their constant innovation in running footwear, thanks to Amber J. Keyser's contribution who wrote a book about the history of athletic footwear [20]. One of the earliest running shoes was created in 1865. It is believed to have belonged to a Lord Spencer and was discovered in a museum in Northampton, England, a city known for its shoemaking at the time. The shoe was likely used for cross-country running because it was lightweight, made of leather, and featured a band that provided enhanced lateral support. J. W. Foster and Sons, now known as Reebok, was another early manufacturer of running shoes. The company was founded in 1890 by Joseph William Foster, a skilled runner who desired to design shoes that would enable him to run faster. British athletes, including 1924 Olympic 100-meter champion Harold Abrahams, wore shoes with leather spikes. In the middle of the nineteenth century, the vulcanization process was developed. In the 1920s, Adi and Rudolf Dassler founded a sports shoe company specializing in track and field footwear in the small German town of Herzogenaurach. Politics, the war, and the incompatibility of their wives all contributed to a rift between the two brothers. In the 1940s, they established competing businesses on opposite sides of the river in the same city. Rudolf Puma founded the company in 1948. Adi was the founder of Adidas (Both companies are still based in Herzogenaurach).

In Phil Knight's biography (Nike's founder) [21] in the 1950s, Bill Bowerman, the head track coach at the University of Oregon, sought to build lighter, quicker running shoes for his runners. The majority of runners used leather spikes identical to those worn by Roger Bannister in 1954 when he broke the four-minute mile mark. Bowerman presented his ideas to many firms, but none of them were interested in participating. In the 1960s, as the popularity of long-distance running increased, Bowerman and one of his former student athletes, Phil Knight, formed Blue Ribbon Sports, a company that imported Japanese athletic footwear known as Onitsuka, which is now known as ASICS. They purchased shoes inspired by Bowerman from Japanese running shoe manufacturers and sold them from the back of vans at races. The Cortez was their best-selling shoe. Due to a spongy rubber midsole, it was one of the first shoes to give cushioning against the stress of the road. In May of 1971, when Bowerman and Knight started their own manufacturing company, the Cortez became the brand's signature shoe. The name of the corporation was Nike [21].

In the 1970s, running shoes were influenced by sports science. Podiatrists, who are becoming increasingly involved in research and design, have found a variety of running gaits and the optimal footwear for each. Ethylene vinyl acetate, or EVA, which is an air-infused foam that provides cushioning and absorbs shock and is still used in the majority of shoes today, was one of the most significant technological advancements in shoe history. In 1975, Brooks introduced EVA for the first time with the Villanova shoe. During the same decade, Nike's Bowerman produced a new sort of lighter traction sole for track shoes. While experimenting with rubber and a waffle iron in his kitchen, he devised the waffle sole, which is currently used on some Nike shoes (although the waffle iron has been retired). Nike was not the first corporation to develop innovative concepts. In 1978, the first shoes designed specifically for women were developed using smaller lasts.

In the late 1970s and early 1980s, running grew in popularity. Between 1971 and 1981, the number of marathon finishers increased by 1,800 percent, and running shoes, along with shoulder pads and mobile phones, became increasingly fashionable. Brooks introduced the first shoe that sought to regulate pronation in 1976 (foot rotation). The Brooks Vantage contained a built-in wedge to encourage the runner's foot to lean slightly outwards, which was thought to cause injuries. In the 1980s, shoes with a combination of sturdy support and enough cushioning were the height of fashion. Among them were the Brooks Chariot (now known as the Beast) and the X-Caliber GT (later known as the Asics Kayano). By incorporating its heel-cushioning bubble innovation, introduced in 1979, into the Nike Air Max in 1987, Nike created an iconic sneaker. In the 1990s and early 2000s, if your shoes did not have a bubble on them, you were absolutely out of style.

Beginning in the year 2000, new studies, including an article published in Nature by Harvard professor Daniel E. Lieberman [4], proved that barefoot jogging can be comfortable for certain individuals, even on the roughest surfaces. This notion became a popular topic of discussion in the running community. Nike created the Nike Free in 2001, a shoe designed to simulate barefoot running by reducing shoe weight and integrating a sole that brought runners closer to the ground. In 2006, Vibram's FiveFingers — glove-like minimal shoes with separate compartments for each toe and ultra-thin rubber soles — were introduced, and minimalism, the idea that fewer shoes are better, became the new vogue. Christopher McDougall published "Born to Run" [22] shortly after staying with the Tarahumara, a Native Mexican band of prolific ultra-runners who wore just narrow sandals to accomplish extraordinary feats of endurance, such as running hundreds of kilometers. This barefoot idea, that footwear should work with the foot's mobility rather than against it, was solidified. Although barefoot running was not the sole solution to injury prevention, it did leave a lasting impression. In recent years, footwear has become lighter, more comfortable, and more fashionable.

Nike launched Flyknit in 2012. Flyknit is a lighter yarn fabric that can be micro-engineered to provide extra support when necessary; after 10 years of development, uppers may be cut from a single piece of fabric. Compared to a conventional cut-and-sew shoe, it saves a few ounces and reduces waste by approximately 60 percent. In 2013, Adidas introduced Boost, a cushioning system developed in partnership with BASF, a famous chemical company. The objective was to create a material that was "better than EVA." The thermoplastic polyurethane midsole contains thousands of energy-returning capsules, which arguably meets the goal. With each step, it contracts under pressure to absorb stress and rapidly rebounds to restore energy. The race is on to create the first pair of shoes that can break the two-hour marathon barrier. The Vaporfly is a two-hour goal-specific shoe featuring a carbon-fiber midsole plate that increases energy return and propels runners ahead [23].

4. Thesis structure

This thesis explores the topic of recognizing and classifying running gait patterns that could lead to overuse injuries. We have designed an experimental protocol and a novel measuring tool that we have applied to running biomechanics. In addition, we have developed an inhouse algorithm than enables us to measure several types of gait sequence features that can be used to extract high level characterizations, such as a biometric signature of the running subjects.

These time and acceleration data integration methods, spanning a range of coarseness of aggregation, are designed to answer the following question, "Why have injured myself today?" These gait features are tested on acceleration/gyroscope data we collected to simulate realistic outdoor running scenarios.

The approach dealing with the running-related injury risk factors aimed to determine:

- How running-related fatigue may affect the ability of the human musculoskeletal system to attenuate the mechanical stresses resulting from running
- How to find a novel approach and methodology to properly measure the risk of impacts while running
- How running shock waves leads to injury and how do we measure one runner's threshold to injury and impact loading.

In chapter 2 we will discuss and describe the functional anatomy of the human locomotor system. To better understand how this locomotor system functions we will classify the motions the joints, the muscles, the levers and in the basic functions. We will then talk about the foot and ankle anatomy then go on to different scientific foot classifications This chapter will give us a better understanding of all the different structures involved in human locomotion. We will also focus on the biomechanical aspects of the human locomotor system. We will lay emphasis on how the anatomy described previously plays a role in walking and running. We will then take a deep dive into the scientific description of walking and running known as gait cycles, phase of gait, kinetics, kinematics and footstrike patterns.

In Chapter 3 we will lay emphasis on how to measure properly walking and running also known as gait analysis. We will take a look into the diverse methods and material that can be used to quantify gait, their relevance and their scientific interest. We will also discuss how these measurements are related to fatigue and which biometric gait parameters can be linked to overuse injuries. We will also describe our protocols, calculations and scientific reasoning behind our hypothesis.

In chapter 4 we will talk about running overuse injuries and how gait analysis enables to predict, prevent and treat biomechanical overuse injuries. We will then discuss the novelty of our experimental approach using the Shock Response Spectrum as a measuring tool to help determine a runner's injury threshold. Finally, we will talk about our general conclusion and the perspective of using our approach for running connected wearables as a mean to perhaps predict overuse injuries before they occur.

Chapter 5 will summarise this thesis in French.

In conclusion the main objective of this thesis is to enable me as a clinician to better understand and correlate data acquisition for my clinic's setup and my patients' pathologies.

CHAPTER 2
Functional anatomy and biomechanics

In this chapter we will cover the human locomotor system from an anatomical and functional perspective. We will go over the basic scientific current concepts of walking and running know as gait analysis in the scientific world. We will obviously lay emphasis on the foot, especially on how the field of podiatry and biomechanics are intertwined when it comes to foot classification and function.

1. Functional anatomy

The human locomotor system (HLS) consists of the muscular and skeletal systems that allow humans to move. The musculoskeletal system supplies the body with structure, support, stability, and mobility. This entire system consists of muscles, cartilage, tendons, ligaments, joints, and other connective tissue that supports and binds tissues and organs. The basic tasks of the musculoskeletal system include supporting the body, permitting movement, and safeguarding essential organs [24,61].

This system illustrates how bones are joined via joints and how muscle fibers connect to these bones and joints through tendons and ligaments. Bones provide support and stability to the body, and joint mobility allows for bone motion and movement. Muscles maintain the position of bones and play a part in bone and joint mobility; when a muscle contracts, the bone linked to the joint is moved. Joints are mostly composed of cartilage to prevent bone ends from rubbing against one another and causing arthritis.

An adult human skeleton is comprised of 22 skull and face bones, 6 ear bones, 1 throat bone, 4 shoulder bones, 25 chest bones, 26 spinal bones, 6 arm and forearm bones, 54 hand bones, 2 pelvic bones, 8 leg bones, and 52 foot bones (Figure 5). There are three primary roles of the skeleton:

The skeleton supports and molds the body; without the spine, for example, upright posture would be impossible.

Our skeleton protects our internal organs, including the brain and heart within the skull and the lungs and heart within the ribcage.

The majority of skeletal muscles, such as the biceps and triceps of the upper arm, are connected to opposing working groups of bones.

Muscle fibers are joined together to produce skeletal muscles. Each fiber can constrict or relax on desire. To shorten a muscle, its fibers must all contract simultaneously. The signal to contract or relax a muscle originates in the brain and is transmitted to the muscle via nerves.

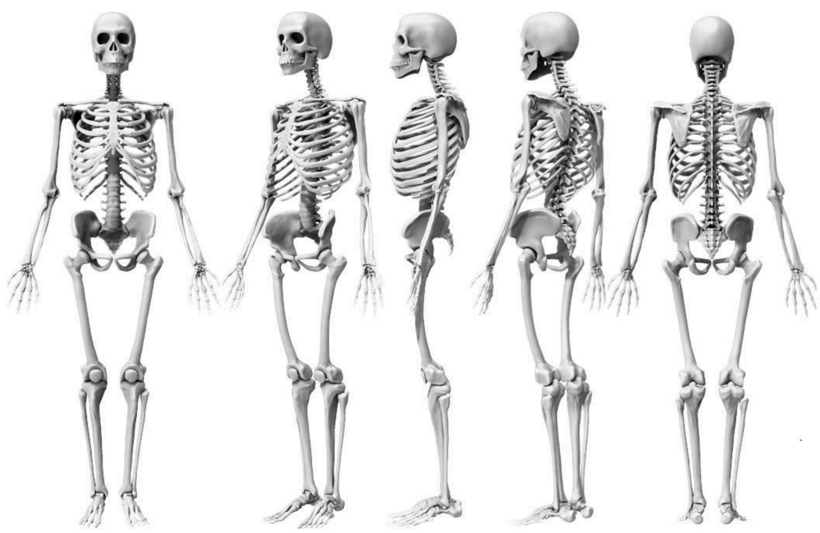


Figure 5: Human skeleton (3D anatomy)

There are three distinct planes of motion: the sagittal plane, the coronal plane, and the transverse plane. As depicted in figure 6, motions are classified according to the anatomical planes in which they occur, but movement is typically a mix of many motions occurring simultaneously in multiple planes [25].

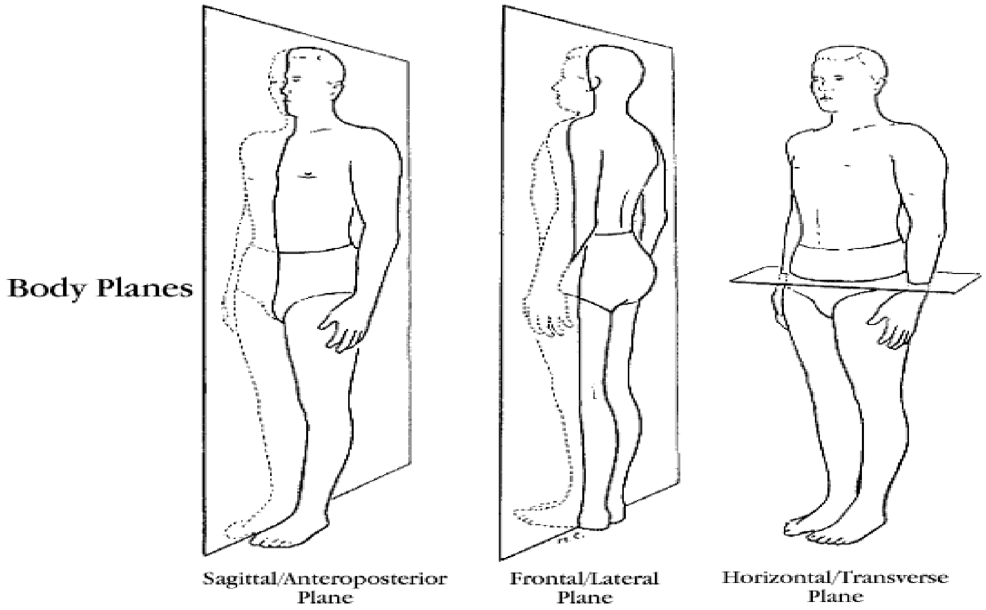


Figure 6: Human body planes (From Human Walking (p. 34) by V. T. Inman, H. J. Ralston, and F. Todd, 1981 Baltimore).

The flexion/extension movement happens in the sagittal plane and about the frontal axis. Flexion is a bending movement that reduces the angle between two parts and brings the distal bone portion closer to the proximal bone portion. The reverse of flexion, extension is a movement that increases the angle between body components and separates the distal end of the bone from its proximal end. Dorsiflexion and plantarflexion are the terms for foot flexion and extension, respectively (figure 7).

Figure 7: Sagittal plane (Kanthi Hegde, 2013)

The abduction/adduction movement occurs in the frontal plane and around the sagittal axis. Abduction is a movement that pulls a structure or part away from the body's midline; doing a side split is an example of hip abduction. Bringing the knees together is a wonderful example of hip adduction, which is a motion that pushes a structure or portion toward the body's midline (figure 8).

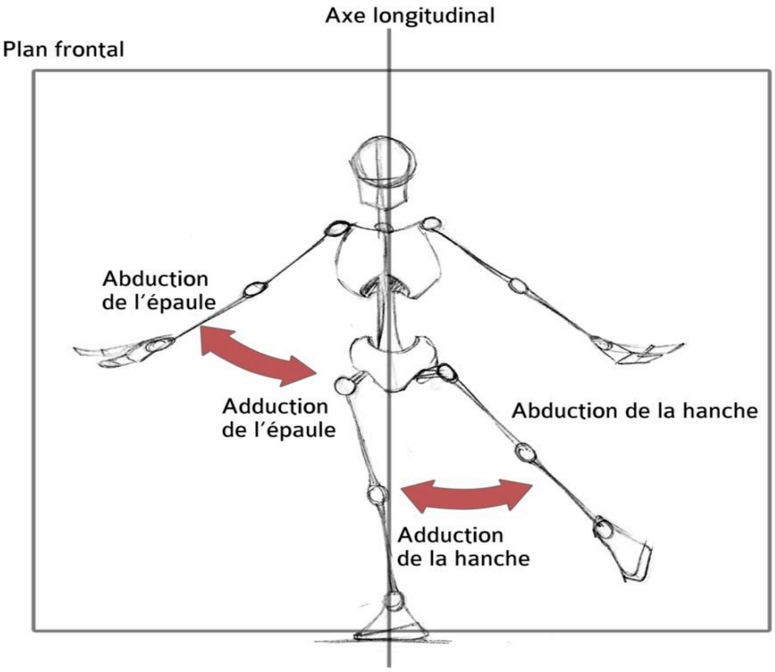


Figure 8: Frontal plane (www.sci-sport.com, 2012)

The motion of medial/lateral rotation takes place in the transversal plane and around the longitudinal axis of the body. A rotation that is toward the axis of the body is referred to as a medial rotation, whereas a rotation that is away from the center of the body is referred to as a lateral rotation (figure 9).

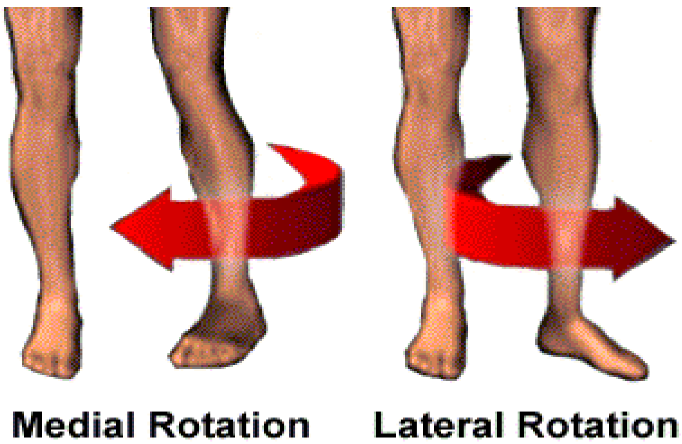


Figure 9: Transversal plane (www.basicmedicalkey.com)

When someone says they are pronating their foot, they are referring to the action of moving their foot outwards such that their weight is supported by the middle part of their foot. Supination of the foot is when the sole of the foot is turned inwards and the weight is borne on the lateral area of the foot. This movement is referred to as the supination of the foot. Eversion is a movement that incorporates flexion, abduction, and pronation into a single motion. As illustrated in figure 10, the inversion action is a mixture of three separate movements: extension, adduction, and supination.

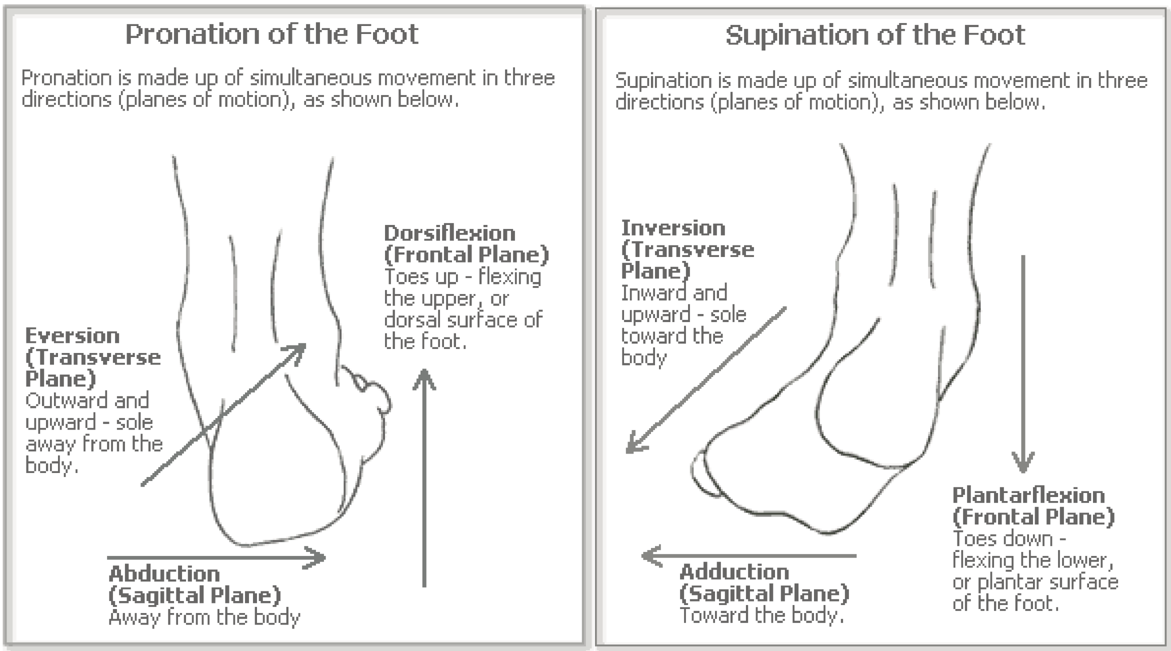


Figure 10: Pronation/Supination (www.boneandspine.com)

The circumduction motion produces a conical movement of the limb. Only the ball-and-socket joints (hip) are truly capable of circumduction or 360 degrees of movement (figure 11).

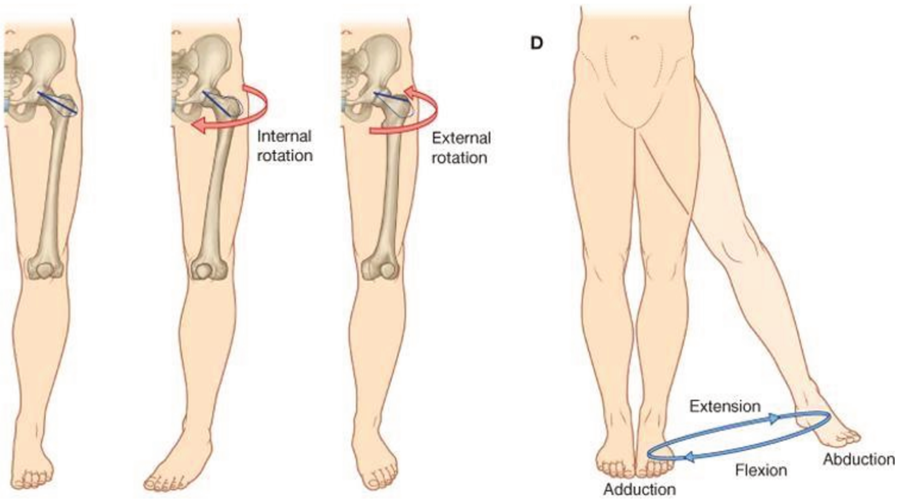


Figure 11: Circumduction (www.basicmedicalkey.com)

The joints can be broken down into three categories: synarthroses, amphiarthroses, and diarthroses [27].

A synarthrose is an immovable joint between bones, such as the sutures of the skull.

An amphiarthrose is a slightly movable joint composed of dense fibrous connective tissue, such as the syndesmosis between the distal ends of the tibia and fibula.

A diarthrose is a freely movable joint between bones covered.

Types of diarthrotic joints:

- A translational joint known as a gliding joint or arthrodia is one in which the articular cartilage surfaces of the surrounding bones are planar and the joint capsule and ligaments allow for only very small sliding movements. There are seven distinct intertarsal articulations in the human foot. These include the calcaneocuboid articulation, the intercuneiform articulation, the cuneonavicular articulation, the cuboideonavicular articulation, and the talocalcaneonavicular articulation (figure 12).

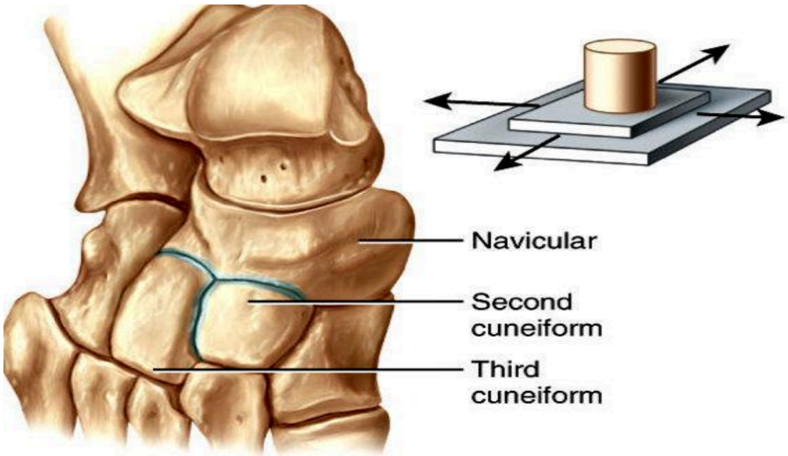


Figure 12: Gliding joint (https://anatomy.co.uk)

- Ginglymus or hinge joint is a type of uniaxial synovial joint in which the articular surfaces are moulded to each other in such a manner as to permit motion only in one plane, and the articular surfaces are connected together by strong collateral ligaments. Ginglymus joints are found in birds, reptiles, and amphibians. The ginglymus is the joint that is typically found between the humerus and the ulna (figure 13). Even though the ankle mortise allows for a very tiny amount of rotation or side-to-side movement in certain places of the limb, it is still considered a hinge joint.

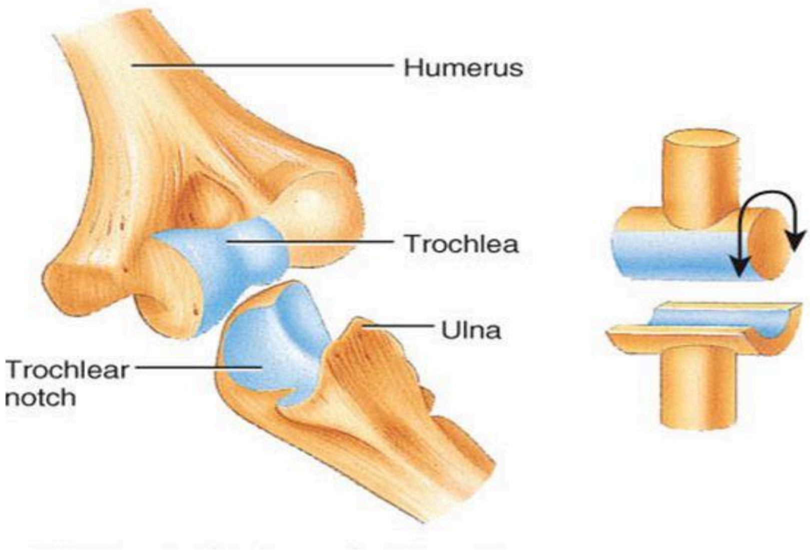


Figure 13: Hinge joint (https://radiopaedia.org)

- A pivot joint, also known as a trochoid, is a type of uniaxial synovial joint in which the range of motion is restricted to rotation. A pivot joint is created by a pivot-like process revolving within a ring, or by a ring resting on a pivot; the ring is composed of bone and ligament to a certain extent. A nice illustration of this can be seen in figure 14, which depicts the rotation that occurs between the end of the axis (C2) and the atlas (C1).

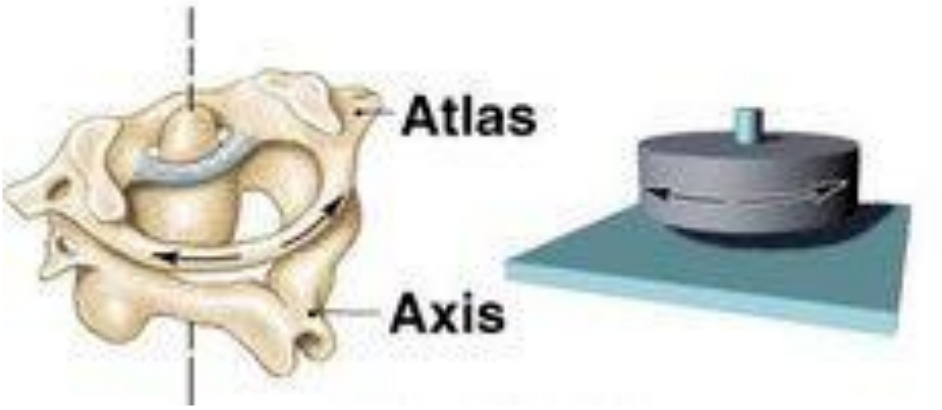


Figure 14: Pivot joint (https://musculoskeletalkey.com)

- One type of synovial joint is called a condyloid joint. In this type of joint, the articular cartilage surface of one bone is rounded and convex, and it slides against the rounded, concave articular cartilage surface of the other bone. Biaxial angular movements such as flexion, extension, abduction, adduction, and circumduction are made possible by this design. In the flexed position, the tibiofemoral component of the knee joint (figure 15) allows for a very modest rotation in either the medial or lateral direction about the axis of the lower leg. At the same time, the femur is able to roll and glide across both menisci during the extension-flexion motion. It is referred to as a condyloid joint because of this particular function of the knee articulation.

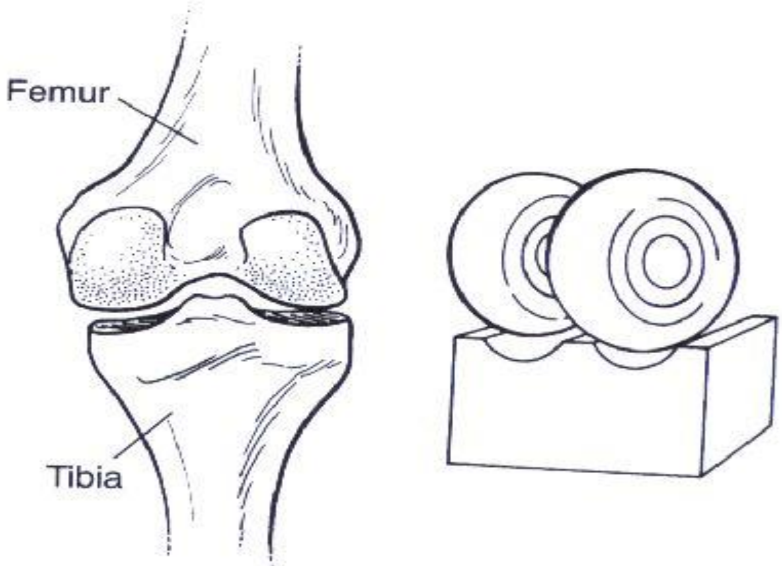


Figure 15: Condyloid joint (https://medical-dictionary.thefreedictionary.com)

- A type of biaxial synovial joint known as a saddle joint or an articulation by reciprocal reception is a joint in which the articular cartilage surfaces of both bones are rounded and include both convex and concave regions on each articular surface. Another name for this type of joint is an articulation. Condyloid joints are replaced with this design, which provides for greater freedom of movement. The carpometacarpal joints of the thumbs, which are seen in figure 16, are an example of a classic saddle joint.

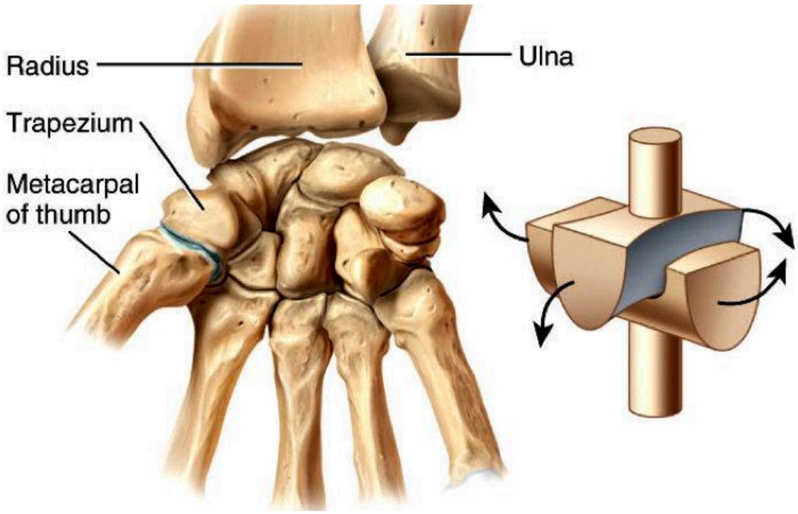


Figure 16: Saddle joint (https://anatomy.co.uk)

- Enarthrosis is a type of synovial multiaxial joint in which the distal bone is capable of mobility around an indeterminate number of axes that have one common center. These joints are also known as ball-and-socket joints. The articular cartilage surface of one bone is convex and hemispherical, and it slides against an articular cartilage surface on the other bone that is rounded and concave, like a cup. This design makes it possible for the user to perform multiaxial angular and rotational motions, including as flexion and extension, abduction and adduction, rotation around the long axis of the first bone, and circumduction. The hip (figure 17) and the shoulder are two places where you can see this type of articulation in action.

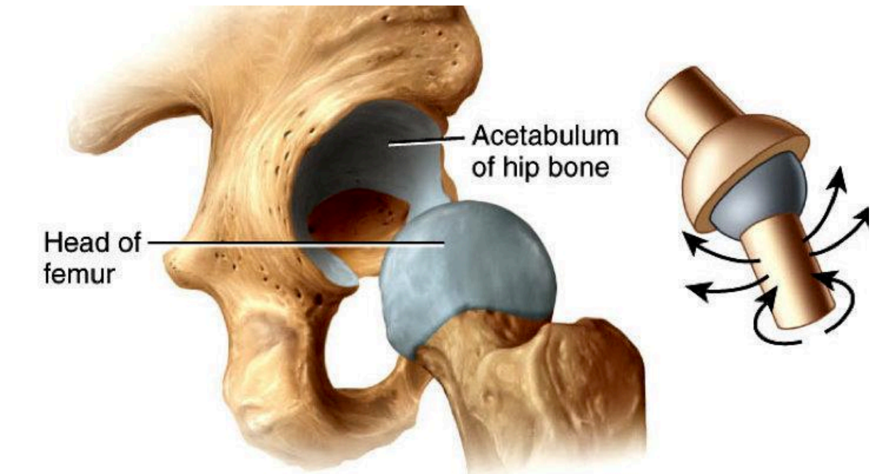


Figure 17: Ball and socket joint (https://anatomy.co.uk)

The muscular portion of the HLS is formed of cardiac, skeletal, and smooth muscles. Smooth muscles are used to govern the flow of substances inside the body and are not consciously controlled, whereas cardiac muscles are entirely responsible for blood circulation and are located in the heart.

Only the skeletal muscles (figure 18) are capable of inducing motion; they are linked to bones and organised in opposing groups around joints, using a variety of contraction types to produce forces that enable gait and posture. Controlling voluntary muscle contraction is the central nervous system.

Classification of voluntary muscle contractions [28]:

- Concentric contraction: the muscle's force is adequate to overcome the external force and generate motion; the muscle shortens as it contracts.The concentric contraction of the triceps surae muscle, for example, causes the toe-off phase of stride.

- Eccentric contraction: the force created is insufficient to counteract the external strain on the muscle, and as the muscle contracts, the muscle fibers lengthen. A body portion is decelerated through the application of an eccentric contraction. The heel contact phase of gait, for instance, produces an eccentric contraction of the triceps surae muscle in order to attenuate the loading phase.

- Isometric contraction: the muscle maintains its length and there are no resulting movements. Pauwels' theory of hip biomechanics, for instance, enables unipodal stance and balance maintenance.

- Isotonic contraction: despite a change in muscle length, the tension in the muscle remains constant. These contractions may be either eccentric or concentric.

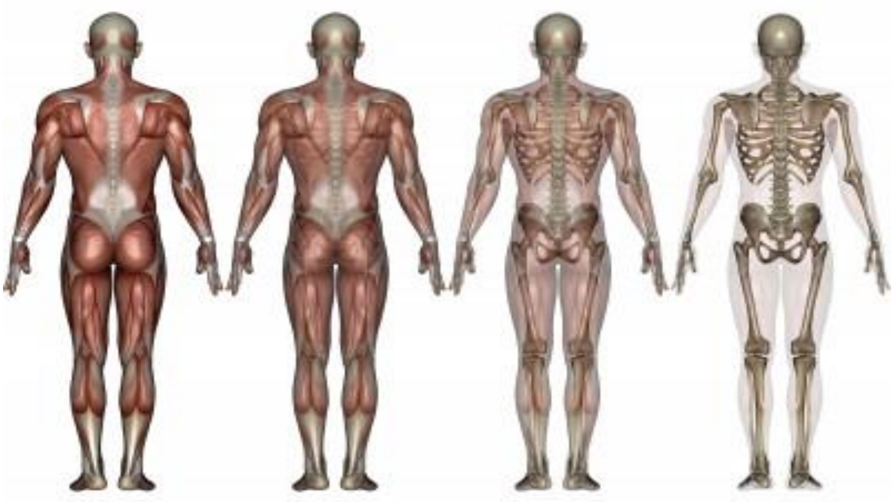


Figure 18: Body muscles (Netter, F. H. 2010. Atlas of Human Anatomy)

Functional classification:

Muscles can be categorized based on the functions they perform, primarily as prime movers, antagonists, fixation muscles, and synergists. Those with a paradoxical or eccentric action, in which muscles extend while contracting, fall into a distinct category. In doing so, they engage in negative activity. A muscle may function as the prime mover in one pattern, the antagonist in another, or the synergist in a third [29].

- Primary Motors A prime mover is a muscle or set of muscles that directly causes the movement that is desired (e.g., flexion of the fingers). Gravity may also serve as a driving force. For instance, if a person holds an object and lowers it to a table, gravity is responsible for the lowering. The sole muscular motion required is to control the pace of descent, a paradoxical activity.

- Antagonists. Antagonistic muscles are those that immediately resist the movement being considered. Thus, the antagonist of the elbow flexors is the triceps brachii, which acts as the elbow's extensor while operating as a prime mover. Depending on the speed and force of the movement, antagonists can be relaxed or, by lengthening while contracting, they can control the action and make it smooth, jerk-free, and precise. The word antagonist is inappropriate, as these muscles cooperate rather than compete. Gravity can also function as an antagonist, such as when the elbow of an anatomically positioned forearm is flexed.

- Fixation Muscles. Generally, fixation muscles stabilize joints or body components and maintain posture or position while the primary movers are at work.

- Synergists. Synergists are a unique category of fixation muscles. Synergists prevent unintended motions at intermediate joints when a primary mover crosses two or more joints. Therefore, the lengthy muscles that flex the fingers would also flex the wrist if the wrist were not anchored by the extensors of the wrist, which act as synergists in this movement.

Biomechanics applies the laws of physics to the study of human movement. Some joints function as levers, others as pulleys, and others as a wheel-axle mechanism. A lever is a rigid device used in conjunction with an axis to double the mechanical force (effort) or resistive force (load) that is applied to it. Mechanical advantage is the effectiveness of the lever (MA). Less work is necessary, the higher the mechanical advantage. To determine mechanical advantage, divide the effort arm by the resistance arm (MA = EA/RA).

In order to analyze the effectiveness of the lever (mechanical advantage), it is necessary to define and quantify the system's constituent parts. This is possible by determining the distance between the axis and the location where the force exerts its influence. Levers are utilized in nearly every movement. A lever is composed of a stiff "bar" that pivots about a fixed fulcrum:

- Fulcrum or axis: the point about which the lever rotate
- Load or resistance : the force applied by the lever system (in biomechanics it is most of the time the body weight)
- Effort or force : the force applied by the user of the lever system

In the human body, the fulcrum is the joint axis, bones are the levers, skeletal muscles usually create the motion, and resistance can be the weight of a body part, the weight of an object one is acting upon, the tension of an antagonistic muscle. Levers are classified by first, second, and third class, depending upon the relations among the fulcrum, the effort, and the resistance [30].

The axis (fulcrum) of a first-class lever (figure 19) is positioned in between weight (resistance) and the force. In the human body, first-class levers are rare. The junction between the head and the first vertebra is an example. The head provides the weight (resistance), the axis is the joint, and the muscular action (force) is provided by any of the posterior muscles attached to the skull, such as the trapezius.

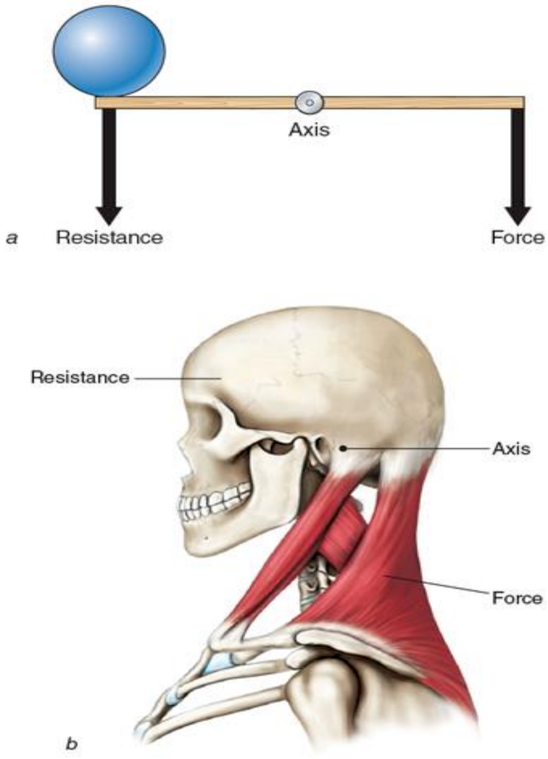


Figure 19: First class lever (https://us.humankinetics.com)

The weight (resistance) is situated between the axis (fulcrum) and the force in a second-class lever (figure 20). The lower leg, when standing on tiptoes, is an example of a second-class lever in the human body. The axis is provided by the metatarsophalangeal joints, the resistance is the body's weight, and the force is applied to the calcaneus bone through the Achilles tendon by the gastrocnemius and soleus muscles.

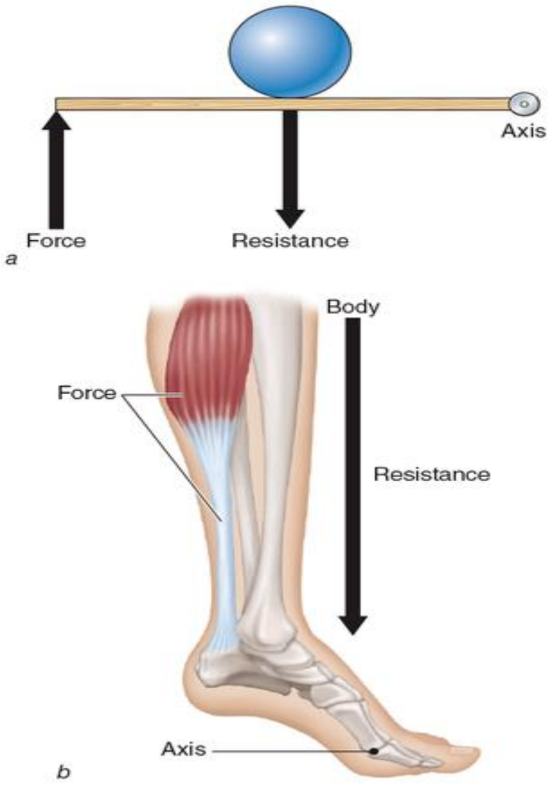


Figure 20: Second class lever (https://us.humankinetics.com)

Force is exerted between the resistance (weight) and the axis of a third-class lever (figure 21), the most prevalent in the human body (fulcrum). The elbow joint is an example of an axis joint (fulcrum). The forearm, wrist, and hand provide resistance (weight). When the elbow is flexed, the force is generated by the biceps muscle.

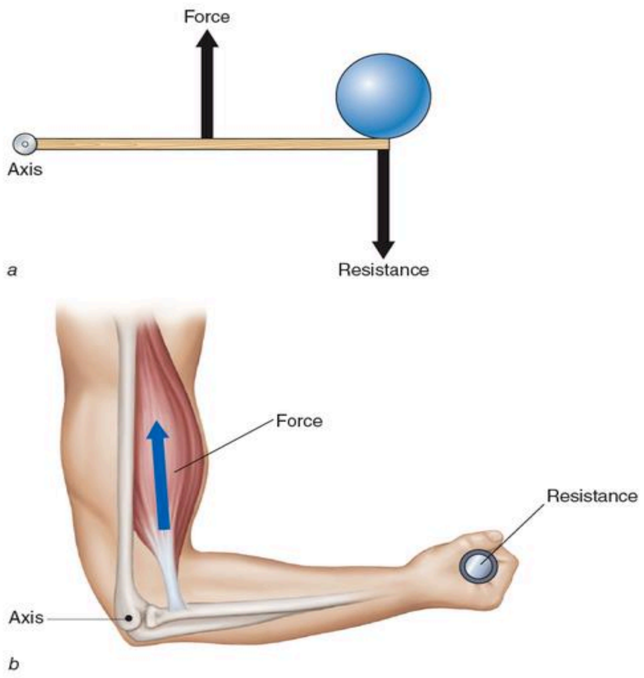


Figure 21: Third class lever (https://us.humankinetics.com)

2. Foot function

During walking the body functionally divides itself in 2 units: passenger and locomotor (figure 22). The passenger unit is responsible only for its own postural integrity, normal gait mechanics are so efficient that the demands on the passenger unit are reduced to a bare minimum making it virtually a passive entity that is carried by the locomotor system. Alignment of the passenger unit over the limbs is a major determinant of muscle action within the locomotor system. The passenger unit is the head, arms, and trunk (HAT), because muscular movement in the neck and trunk solely serves to maintain neutral spinal posture. During normal gait, there is very little postural alteration.

The arm swing is mainly a passive response to the momentum that has been generated. As indicated by the ease with which one may carry packages, a little amount of active control has not proven to be necessary. Furthermore, experimental arm swing restriction had no effect on the energy expenditure of walking. The center of gravity is just above the hip joints, just anterior to the tenth thoracic vertebra. Because of the length of the lever (33 cm in an average adult), the passenger unit's balance is extremely sensitive to changes in the supporting limbs' alignment. The locomotor unit is comprised of two limbs interconnected by a pelvis. With two highly movable connection locations, the lumbar spine and hip joints, the pelvis becomes a component of both the passenger and locomotor units. The human locomotor system's unique structure allows for bipedal movement as well as upright stability and gait [10].

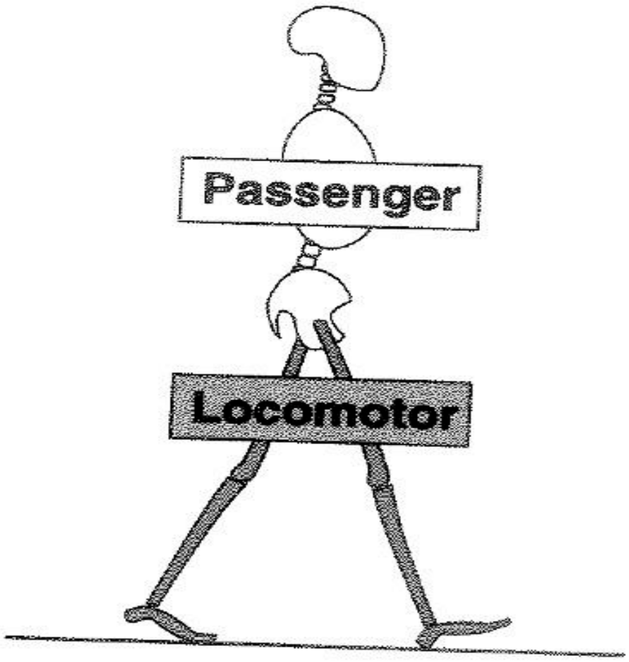


Figure 22: Human locomotor system units (Perry, J., 1992. Gait analysis: normal and pathological function)

The foot contains 26 bones, 2 sesamoid bones, 33 joints, 19 muscles and 107 ligaments as shown in figure 23. The foot is commonly divided into three segments [31]:
The rearfoot consists of two bones: The talus and calcaneus. The tibia articulates with the talus dome, allowing leg forces to be transferred to the ankle. The talus also articulates with the fibula. The talus articulates with the calcaneus, the main weight-bearing (and the largest) bone of the foot by way of the subtalar joint which has three surfaces of articulation with three separate facet joints. A great deal of the movement in the ankle occurs in this joint - the rest of the movement occurs at the tibial-talar joint.
The navicular bone, the cuboid, and three cuneiform bones: medial, middle, and lateral, make up the midfoot and have multiple articular surfaces. The fourth and fifth metatarsals articulate with the cuboid

bone on the outside. Each of the cuneiform bones articulates with the first, second, and third metatarsals. Each of them has its own joint capsule, but they are all wrapped together to form the tarso-metatarsal joint (also known as the "Lisfranc joint"). The talonavicular and calcaneocuboid joints form the combined articulation of the midtarsal joint ("Chopart joint") when they are close together.

The forefoot is comprised of five metatarsals, numbered one through five, and five toes, each of which is made up of three bones (except for the big toe which consists of two). The proximal phalanx, middle phalanx, and distal phalanx are the bones of each toe (except the big toe which has only proximal and distal). A joint connects each of these bones, allowing for the movement required by each portion of the foot.

- The MTP joint (metatarsal phalangeal joint) connects the metatarsal and the proximal phalanx of the neighboring toe in the forefoot.
- Between the proximal phalanx and the middle phalanx of each toe is the PIP joint, or proximal interphalangeal joint.
- DIP joint - distal interphalangeal joint - connects each toe's middle and distal phalanx.

- Because the large ("great") toe has just one joint between its two phalanges, the great (or "big") toe interphalangeal joint is named after it.
- - The metatarsal head is the end of the metatarsals that articulates with the neighboring toe's proximal phalanx.

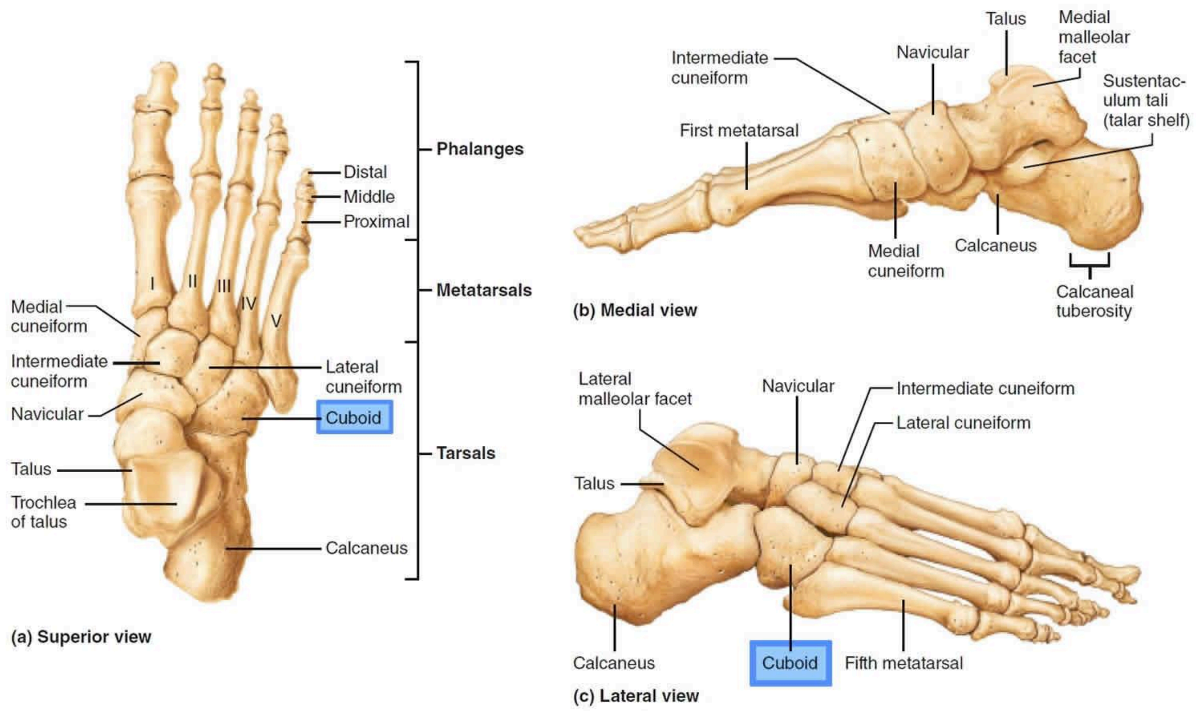


Figure 23: Foot anatomy (Netter, F. H. 2010. Atlas of Human Anatomy)

The ankle joint consists of a bony fit between the talus and the tibia proximally and medially and the talus and the fibula laterally (figure 24). The dorsal and the medial surface of the talus contact reciprocally shaped areas of the tibia. The lateral aspect of the talus articulates with the articular surface of the distal fibula. This joint adds critical stability to the ankle joint. Functionally, the ankle joint also includes the proximal tibiofibular joint, located at the posterior inferior and lateral region of the knee.

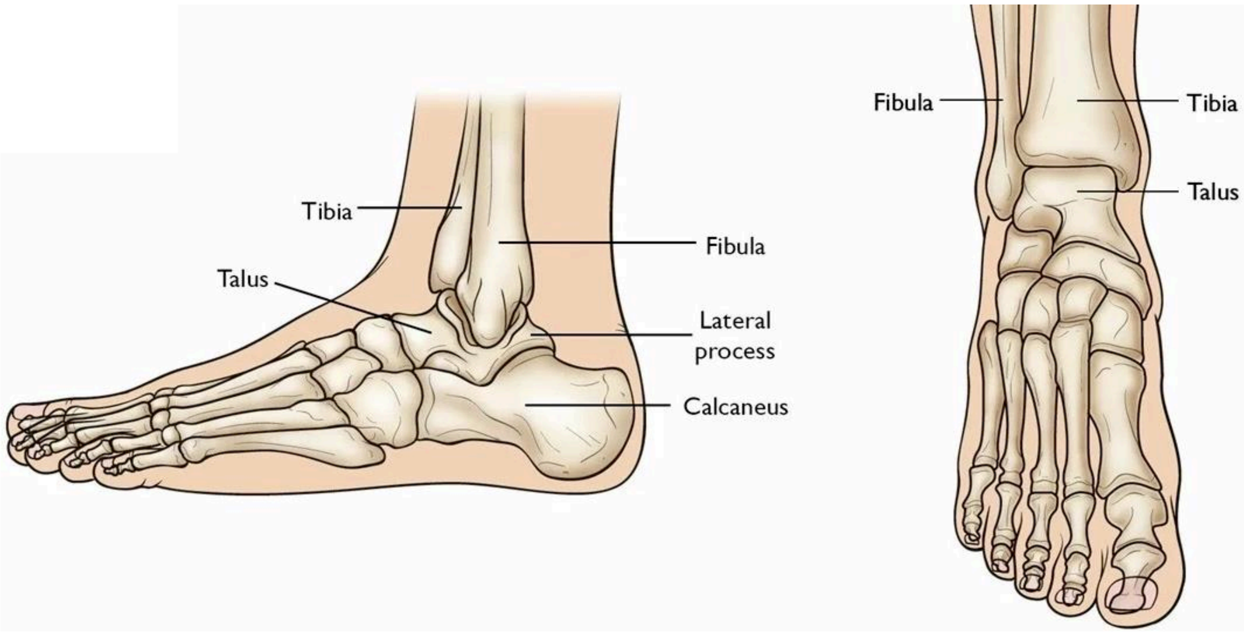


Figure 24 Ankle anatomy (https://orthofixar.com)

As indicated in figure 25, Tong et al established three foot type classifications [32]:

- Pes cavus, cavus feet, varus foot, supinated, under-pronating, non-pronating - High arch, pes cavus, cavus feet, varus foot, supinated, under-pronating, non-pronating
- Normal, middle, average arch, rectus, normal foot - Neutral, normal, middle, average arch, rectus, normal foot
- Flat Foot - Flatfoot, pes planus, flat arch, planus feet, low arch, valgus foot are all terms for the same thing.foot

Other classifications may include:

- Neutral Foot - Shock absorption, adaption, stance, and propulsion all occur at the appropriate times.
- Forefoot Varus - This foot spends too much time in the shock-absorbing phase and does not switch to propulsion until it is too late. Surface knee discomfort, shin pains, Achilles tendonitis, iliotibial (IT) band pain, plantar fasciitis, and low back pain are some of the symptoms.
- Rearfoot Varus - When found with a Forefoot Varus, this foot works similarly to a Forefoot Varus. When found with a Valgus Foot, however, it acts similarly as a Valgus Foot.
- Rigid Forefoot Valgus - This foot turns to propulsion prematurely at a time when it should be absorbing shock. Ankle sprains, a shaky walk, every type of foot discomfort, leg muscle difficulties, and stress fractures are all symptoms.

- Plantar Flexed or Flexible First Metatarsal - This is the most difficult foot type to define. It can act like a Forefoot Varus, Rearfoot Varus, and even a Rigid Valgus in some situations, although not as severely. Everything, including sciatica, is a symptom.
- Equinus - As the center of gravity crosses over the ankle, this foot type is unable to place the foot 10 degrees closer to the shin. A foot that spends too much time in the shock-absorbing phase with little or no conversion to propulsion is one of the symptoms.

Other biomechanical factors include: angles of the shin bones, knee, hip, and leg length differentials, structural, and functional.

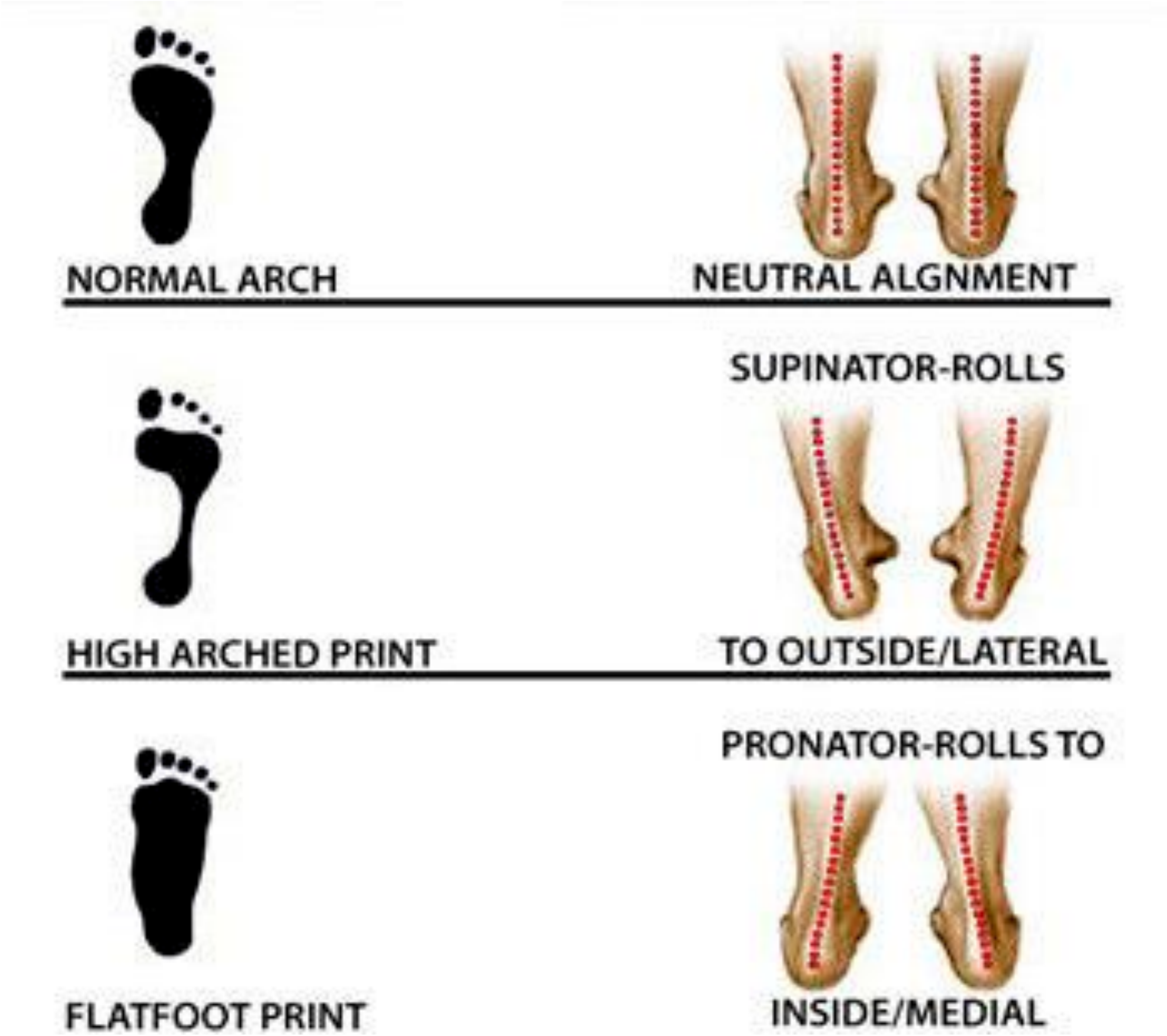


Figure 25: Foot classification (RealyHealth, 2007)

Foot Posture Index (FPI-6)

The foot posture index (FPI-6) is a clinical method of assessing how pronated, neutral, or supinated a foot is [33]. It is a reliable and robust method of assessing static foot structure, and it provides a more valid way to determining static foot structure. The practitioner makes a series of six observations and palpations, and each measure is graded from -2 to 2. A neutral foot has a total score of 0; a pronated foot has a positive score, while a supinated foot has a negative value. The patient is requested to stand steady in a double stance position for scoring. The scoring criteria for FPI are listed in the table below, obtained from Lee et al. There are six criteria for the Foot Posture Index: Talar head position, supra and infra lateral malleolar curvature, calcaneal frontal plane position, prominence in the region of the talonavicular joint, height and congruence of the medial longitudinal arch, abduction/adduction of the forefoot on the rear foot (figure 26).

	-2	-1	0	+1	+2
Talar head palpation	Talar head palpable on lateral side/but not on medial side	Talar head palpable on lateral/slightly palpable on medial side	Talar head equally palpable on lateral and medial side	Talar head slightly palpable on lateral side/palpable on medial side	Talar head not palpable on lateral side/but palpable on medial side
Supra and infra lateral malleoli curvature (viewed from behind)	Curve below the malleolus either straight or convex	Curve below the malleolus concave, but flatter/more than the curve above the malleolus	Both infra and supra malleolar curves roughly equal	Curve below the malleolus more concave than curve above malleolus	Curve below the malleolus markedly more concave than curve above malleolus
Calcaneal frontal plane position (viewed from behind)	More than an estimated 5° inverted (varus)	Between vertical and an estimated 5° inverted (varus)	Vertical	Between vertical and an estimated 5° everted (valgus)	More than an estimated 5° everted (valgus)
Prominence in region of TNJ (viewed at an angle from inside	Area of TNJ markedly concave	Area of TNJ slightly, but definitely concave	Area of TNJ flat	Area of TNJ bulging slightly	Area of TNJ bulging markedly
Congruence of medial longitudinal arch (viewed from inside)	Arch high and acutely angled towards the posterior end of the medial arch	Arch moderately high and slightly acute posteriorly	Arch height normal and concentrically curved	Arch lowered with some flattening in the central position	Arch very low with severe flattening in the central portion - arch making ground contact
Abduction/adduction of forefoot on rearfoot (view from behind)	No lateral toes visible. Medial toes clearly visible	Medial toes clearly more visible than lateral	Medial and lateral toes equally visible	Lateral toes clearly more visible than medial	No medial toes visible. Lateral toes clearly visible.

Figure 26: Foot posture index (Redmond, A. C, 2008)

Medial Longitudinal Arch Angle (MLAA)

The MLAA is a reliable uni-planar measure with a significant level of acceptance within the measure for foot classification and broad foot classification limits [34]. A line should be drawn from the center of the medial malleoli to the navicular tuberosity, and another from the navicular tuberosity to the first metatarsal head (figure 27).

LAA refers to the obtuse angle formed by these two lines. The normal maximum LAA ranges from 131 to 152 degrees. Low-arch refers to a foot with a lower LAA, whereas high-arch refers to a foot with an angle greater than 152°. From the center of the medial malleoli to the head of the first metatarsal, a Feiss line is drawn. In a high-arched foot, the navicular tuberosity should be above the arch, while in a low-arched foot, it is below the line.

Rearfoot Angle (RFA)

A skin marker pen is used to precisely locate and mark four areas. These are: the calcaneus base; the Achilles tendon attachment; the Achilles tendon center at the height of the medial malleoli; and the center of the posterior part of the calf 15 cm above marker three [35], as shown in figure 27.

A goniometer was used to determine the RFA. The RFA is the acute angle formed by the projections of lines one and two. 5° valgus is a pronated foot, 4° valgus to 4° varus is a neutral foot, and 5° varus is a supinated foot, according to RFA.

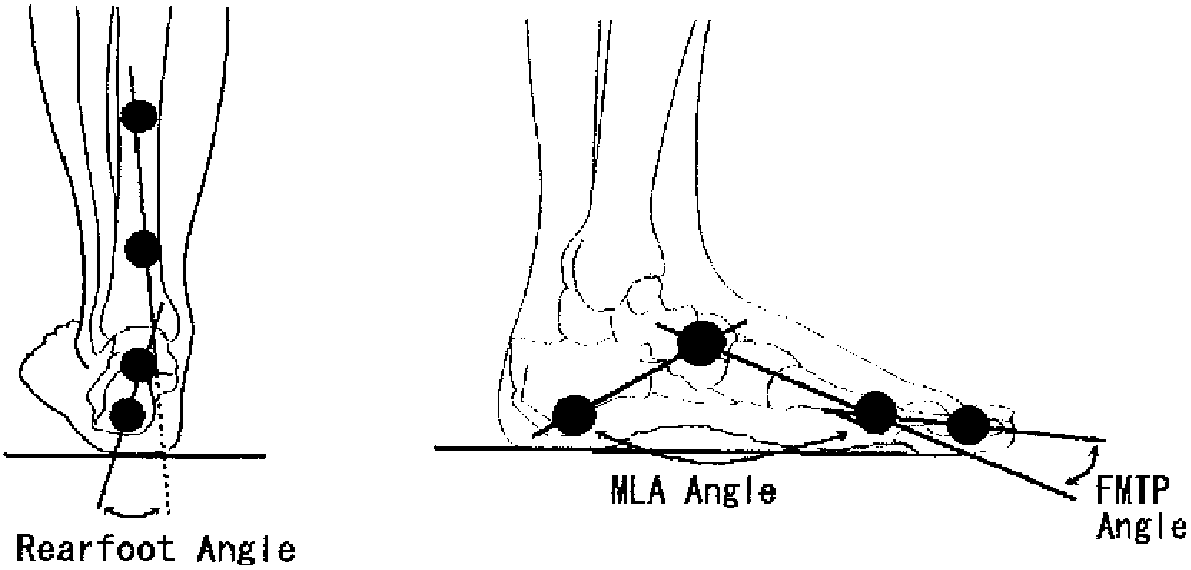


Figure 27: MLAA and RFA (Hiroshige Nakamura, 2003)

Navicular Drop Test (ND)

The navicular drop test [36] is a method for examining the function of the medial longitudinal arch, which is crucial for examining individuals with sports injuries. Differences in navicular drop between healthy and impaired participants have yielded conflicting results. Foot length, age, gender, and Body Mass Index (BMI) may all influence the navicular drop, hence normal values have yet to be defined. It is not an acceptable measure for defining the foot, according to Langley et al.

Subtalar Joint Neutral (STJN)

Dr. Merton L. Root was the first person to conceptualize the root theory, and he did it throughout the latter half of the 1950s and the early 1960s. It is also known as "the foot morphology hypothesis," "the subtalar joint neutral theory," or simply as the "Rootian theory" [13]. Simply put, the root theory is founded on a set of static measures, from which the author derives the belief that these measurements may accurately anticipate kinematic function. When walking, the foot should be in a normal position where the subtalar joint (STJ) is in a neutral position and the midtarsal joint is completely locked. This should take place between the mid-stance and heel-off phases of the gait cycle. It is deemed abnormal for there to be any deviations from the indicated STJ alignment, and as a result, there ought to be mechanical dysfunction present (figure 28). The root hypothesis has been called into question in terms of its reliability, validity, and practical application, despite the fact that it has received clinical acceptance and is utilized in the majority of podiatry and orthopedic literature. The 1950s and 1960s were a time when more sophisticated measuring equipment was not readily available. In addition, the majority of the research that has been done using this method has been carried out using approaches that are distinct from those that Dr. Root has advocated. It is also important to note that Dr. Root has prescribed orthotics in his practice; however, the specific procedures that he used have not been published or documented; as a result, it is impossible to rely on interpretations of such approaches.

Jarvis et al. conducted a study in 2017 [37] to investigate foot kinematics between normal and abnormal feet classified according to Root et al., determine if the degree of structural deformity is associated with the degree of compensations, and measure the position of the subtalar joint during gait in pain-free feet. The study was published in 2017. According to the findings of this study's kinematic analysis, there is no correlation between the abnormalities proposed by Root et al. and the variations in foot kinematics that occur during gait. Since STJ inversion in the neutral calcaneal standing position (NCSP) shows no link to rearfoot kinematics, this indicates that the clinical use of "subtalar joint neutral" might not provide clinicians with information that is accurate [38]. The first MPJ dorsiflexion during gait propulsion was discovered by Jarvis et al. to be significantly less than 65 degrees, which was Root's postulated value. Because it is obvious that not all feet with structural defects should present symptoms, even though their function would be impacted, the classification of the root appears to be erroneous, and it is felt that it is no longer fit for professional practice. The fact that the static assessment is performed in a non-weight bearing position, which may not be similar to the applied external and muscular force when the subject is weight bearing, is a straightforward reason for why it does not necessarily reflect kinematic features.

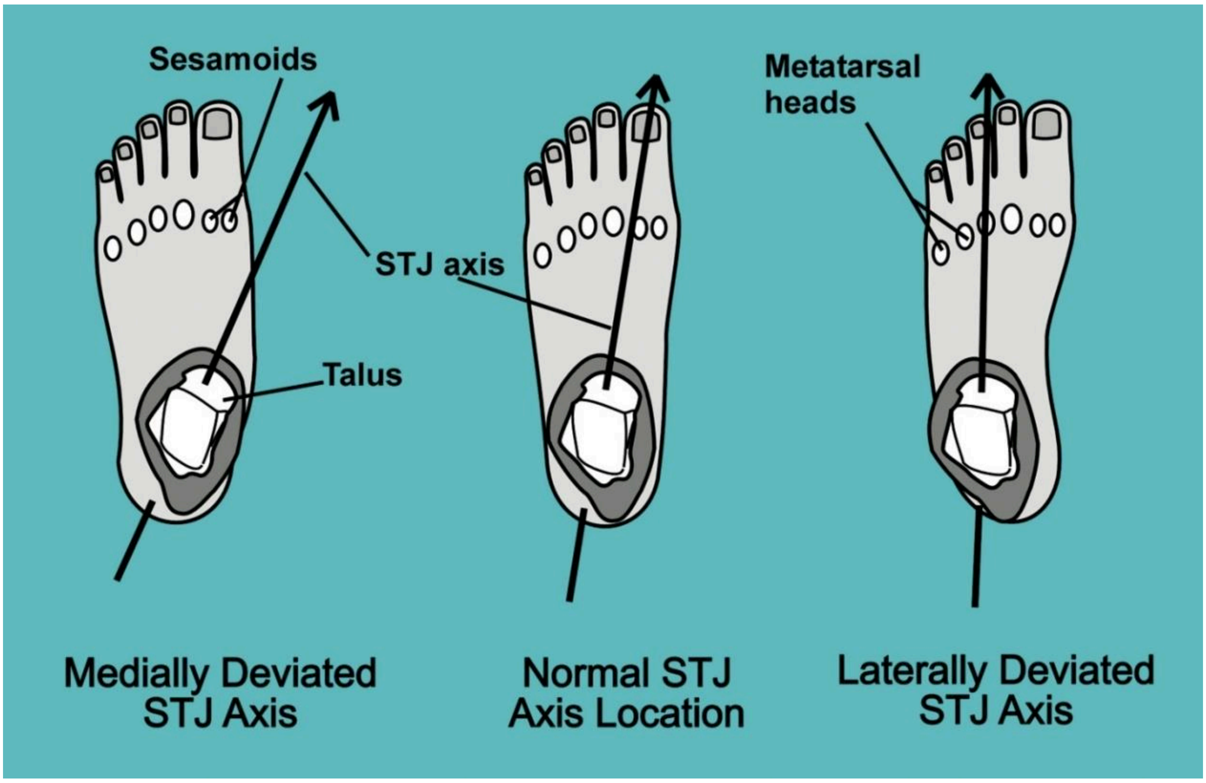


Figure 28: STJ axis (Kirby, 2001)

Forefoot Angle

To quantify forefoot varus or forefoot valgus, the relationship between the forefoot and the rearfoot has to be assessed [39]. The patient is in a prone laying position for the non-examined lower extremity to measure the correlation. The correlation between the forefoot and the rearfoot is noticed once the STJN is attained. The goniometer's stationary arm is parallel to the line bisecting the calcaneus, with the fulcrum on the point bisecting the calcaneus. The goniometer's movable arm is parallel to an imaginary line passing through the metatarsal heads. A forefoot angle of 0° is considered neutral, whereas forefoot varus is positive and forefoot valgus is negative.

Arch Height Index (AHI)

The AHI is a measurement of the medial longitudinal arch, based on which the foot can be classified as high-arched, normal, or low-arched [40]. Williams and McClay investigated several foot measures and ratios and concluded that height of dorsal surface of foot at 50% of foot length divided by truncated foot length was a reliable and valid method for determining AHI [41]. The measurement can be done with a caliper and a graph sheet. As indicated in figure 29, the patient is standing and a caliper is used to assess foot length, the height of the dorsum of the foot at 50% of foot length, and shortened foot length.

AHI = Height of the dorsum of foot at 50% of foot length ÷ Truncated foot length

The foot is deemed high arched if the ratio is 0.356 or above, and low arched if the ratio is less than or equal to 0.275. AHI at ten percent and ninety percent of body weight can be used to determine arch mobility. The Arch Rigidity Index (ARI), which is determined as standing AHI/sitting ARI, has also been suggested as a validated and accurate alternative to the navicular drop test.

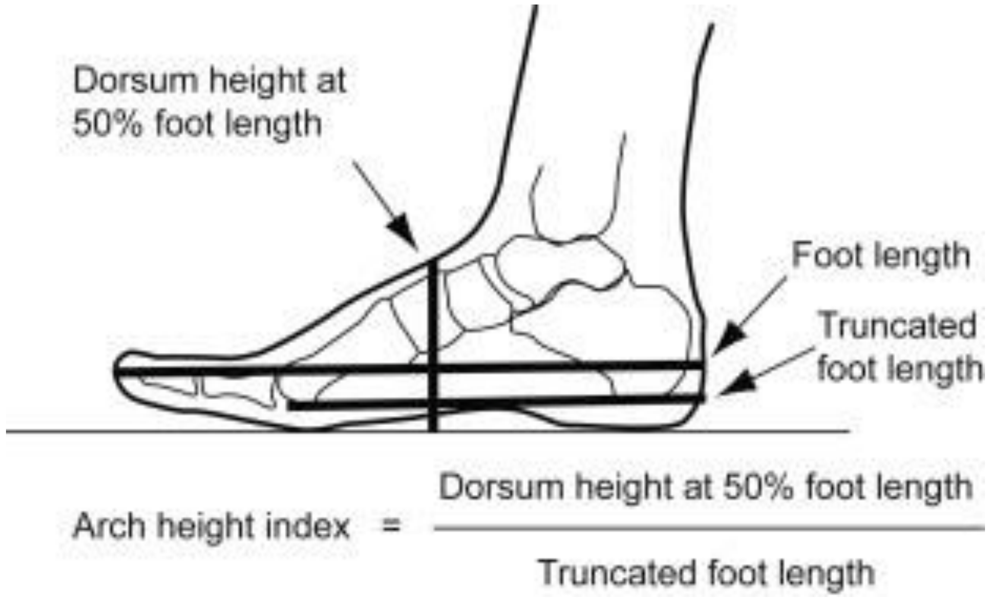


Figure 29: Arch height index (K.K. Whitcome, 2014)

Tibial Torsion Measurement/Thigh-foot angle (TFA)

The patient is placed in prone laying with knees flexed to 90° to determine internal or external tibial torsion. Between the line dividing the posterior thigh and another line dividing the foot, a thigh-foot-ankle (TFA) measurement is taken. TFA greater than 30 degrees indicates significant external tibial torsion, while TFA less than 0 degrees indicates internal tibial torsion. [42].

3. Gait

The medical term for human movement or the manner in which humans walk is gait. Interestingly, each person has a distinct walking pattern. Injuries, diseases, and ill-fitting footwear can have a profound effect on a person's gait. Walking utilizes a series of repetitive limb movements to propel the body forward while maintaining stance stability. As the body advances, one limb serves as a mobile support while the other limb advances to a new support location, and then the functions of the limbs are reversed. Both feet must be in contact with the ground for the transfer of body weight from one limb to another. This sequence of actions is repeated by each limb in a reciprocal fashion till the individual reaches their destination. It is a mechanism that requires the coordinated movement of the subject's bones, muscles, joints, and neurological system. The conditions for gait include the capacity to sustain or take an upright stance, the capacity to retain balance in an upright position under static and dynamic situations, and the capacity to initiate or generate a new forward step [10].

The gait cycle (GC) is used to describe the intricate process of walking or our gait pattern [43]. This cycle covers the motions from the initial contact of the heel with the ground to the second touch of the same heel with the ground. A GC is a single sequence of these functions performed by a single limb. Each GC is broken into two distinct phases: stance and swing, as depicted in Figure 30. Stance is the term used to describe the entire time a foot is planted on the ground.

The stance phase of a nonpathological gait begins with the initial contact (IC) of the heel. Swing refers to the time the foot is in the air during limb advancement; swing begins when the foot leaves the ground during the toe-off phase. The normal distribution of floor contact periods during a GC is around 60% stance and 40% swing [44].

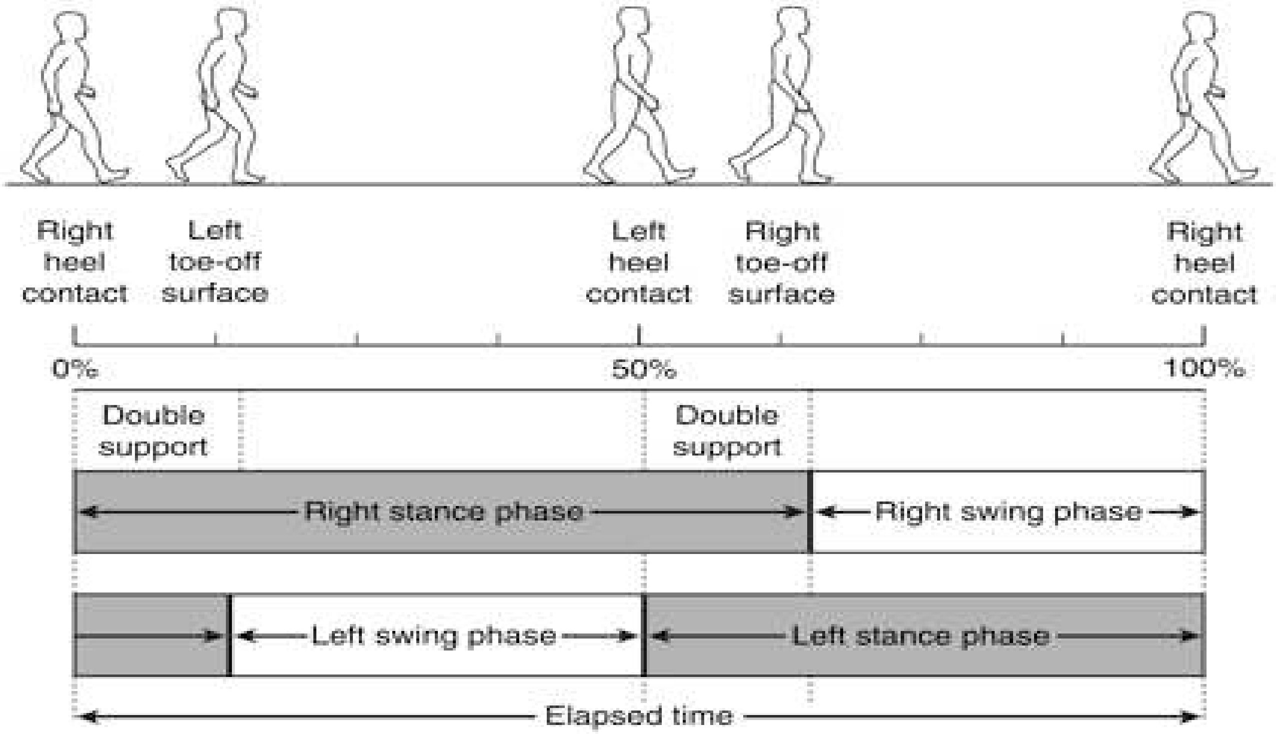


Figure 30: Simplified gait cycle (Perry, J., 1992. Gait analysis: normal and pathological function)

Stance is separated into three intervals based on the order in which both feet make contact with the ground. The IC begins the initial double stance, which is the first time both feet are on the ground throughout the GC; we will refer to this as the double limb stance (DLS). The single limb support (SLS) occurs when the opposite foot is lifted for the swing; during the SLS interval, the full body weight is supported by a single limb.

The terminal DLS is the third division of the stance phase, beginning with the initial contact of the contralateral foot and continuing until the original stance limb is raised for the swing phase. During the stance phase, the floor contact times are as follows: 10% for beginning DLS, 40% for SLS, and 10% as seen in figure 31 for terminal DLS. [10].

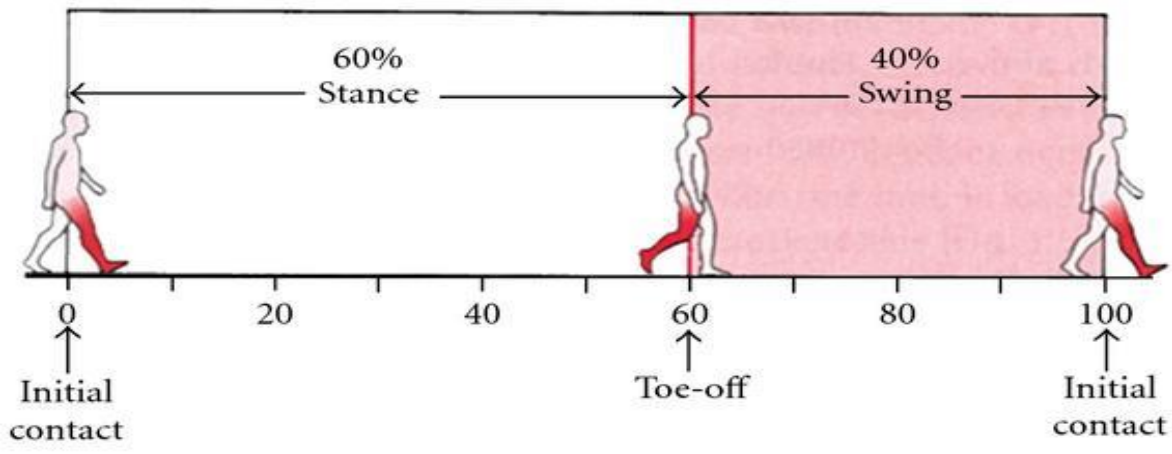


Figure 31: Stance and swing phase (Perry, J., 1992. Gait analysis: normal and pathological function)

It is important to know that walking faster proportionally lengthens SLS and shorten the two DLS. When DLS is omitted, the person has entered the running mode of locomotion. The GC has also been identified as a stride; a stride is the equivalent of a GC. At the midpoint of a stride the opposite foot contacts the ground for its initial DLS, the interval between the IC of each foot is a step. There are 2 steps in each stride (figure 32).

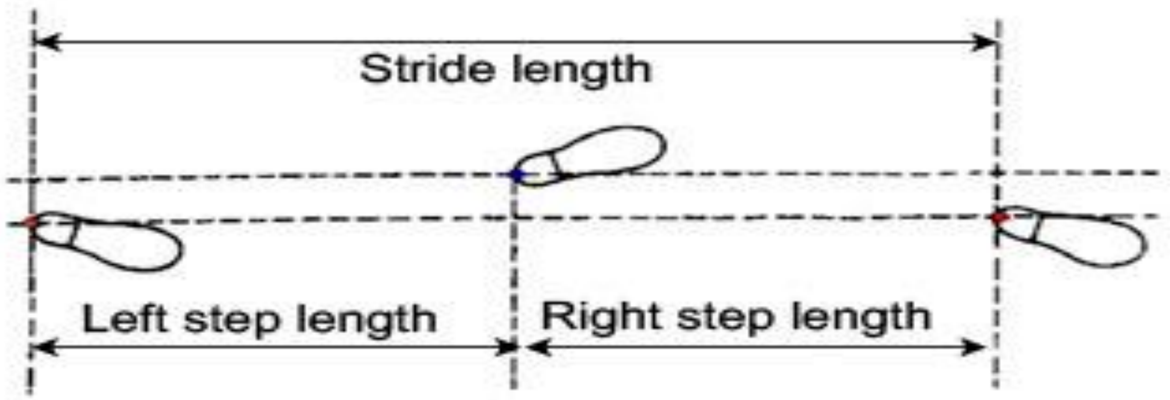


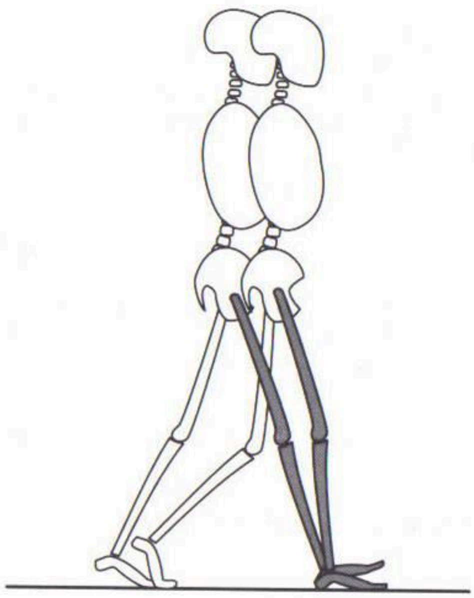
Figure 32: Stride and step (Perry, J., 1992. Gait analysis: normal and pathological function)

The GC is divided in phases of gait and each stride contains 8 functional patterns according to Perry, J. (1992. Gait analysis: normal and pathological function). Each of the 8 gait phases has a functional objective and a critical pattern of selective synergistic motion to accomplish this goal. The limb has 3 basic tasks to accomplish: weight acceptance, limb stability and the preservation of progression [10].

Phase 1 - Initial contact
Interval: 0% to 2% of the GC
Initial contact represents the beginning of the stance phase and of the initial DLS. This phase includes the foot drop and the immediate ground force reaction. This phase is often referred to as heel strike in normal gait.

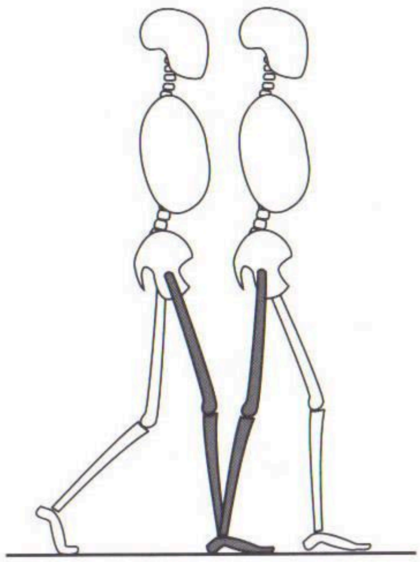
Initial Contact

Phase 2 – Loading response
Interval: 2% to 12% of the GC
This phase follows the IC and continues until the other limb is lifted for swing (end of the initial DLS). During loading response, the foot comes in full contact with the floor and body weight is fully transferred onto the stance limb.



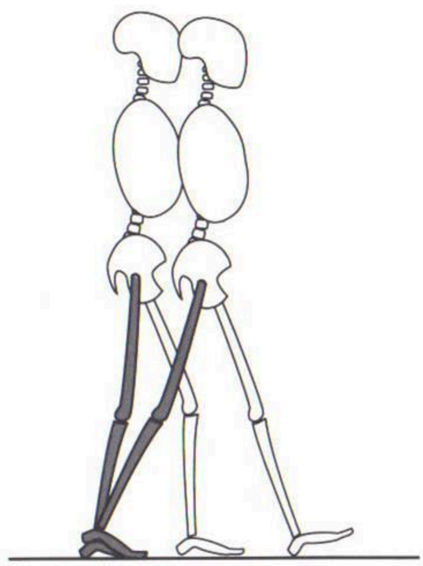
Loading Response

Phase 3 – Mid stance
Interval: 12% to 31% of the GC
Mid-stance represents the first half of single limb support. It begins when the contra-lateral foot leaves the ground and continues as the body weight travels along the length of the foot until it is aligned over the forefoot.



Mid Stance

Phase 4 – Terminal stance
Interval: 31% to 50% of the GC
Terminal stance constitutes the second half and end of single-limb support. It begins with heel rise and ends when the contra-lateral foot contacts the ground. During this phase, body weight moves ahead of the forefoot.

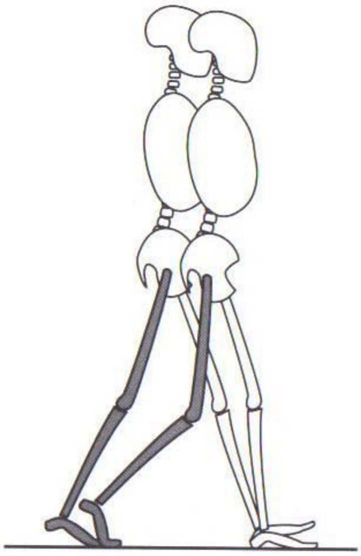


Terminal Stance

Phase 5 – Pre-swing

Interval: 50% to 62% of the GC

Pre-swing is the terminal double-limb support period. It begins when the contra-lateral foot contacts the ground and ends with ipsilateral toe off. During this period, the stance limb is unloaded and body weight is transferred onto the contra-lateral limb. The termination of stance and the onset of swing is defined as the point where all portions of the foot have achieved motion relative to the floor. Likewise, the termination of swing and the onset of stance may be defined as the point when the foot ends motion relative to the floor. Toe off occurs when terminal contact is made with the toe.

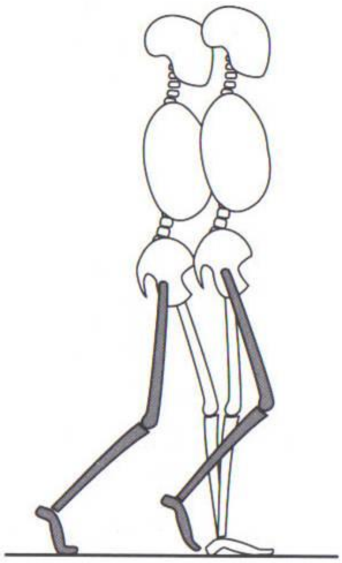


Pre-Swing

Phase 6 – Initial swing

Interval: 62% to 75% of the GC

Initial swing begins the moment the foot leaves the ground and continues until maximum knee flexion occurs, when the swinging extremity is directly under the body and directly opposite the stance limb.

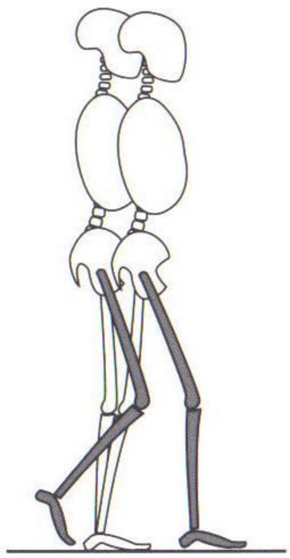


Initial Swing

Phase 7 – Mid-swing

Interval 75% to 87% of the GC

Mid-swing begins as the swinging of the foot is opposite to the stance limb. This phase begins following maximum knee flexion and ends when the swinging limb is forward and the tibia is in a vertical position.

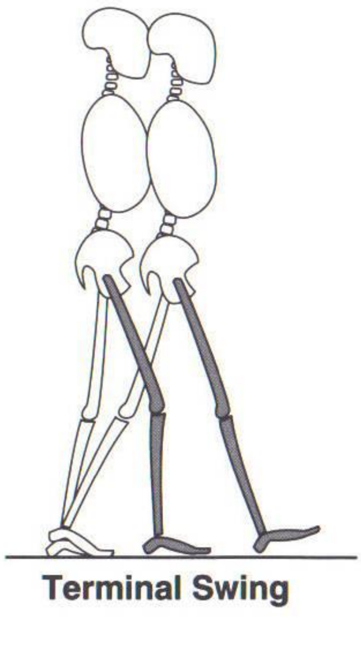


Mid Swing

Phase 8 – Terminal swing

Interval: 87% to 100% of the GC

In the final phase of terminal swing the tibia passes beyond perpendicular and the knee fully extends in preparation for heel contact. Limb advancement is completed as the leg moves ahead of the thigh.



Terminal Swing

In the swing phase we can recognize two extra phases: acceleration and deceleration (figure 33). The acceleration phase goes from toe-off to mid-swing, while deceleration goes from mid-swing to heel strike. In the acceleration phase, the swinging leg makes an accelerated forward movement with the goal of propelling the body's weight forward. The deceleration phase decreases the velocity of this forward body movement in order to absorb the shock.

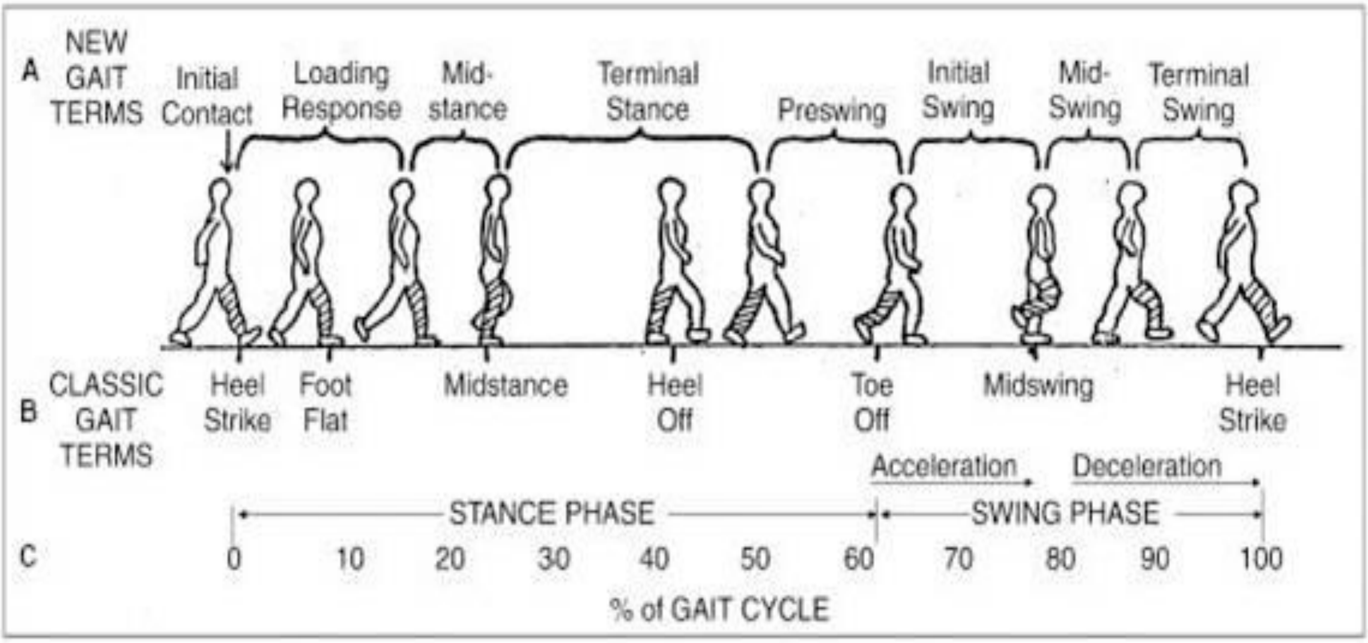


Figure 33: Detailed gait cycle (Perry, J., 1992. Gait analysis: normal and pathological function)

The joints of the foot are controlled by extrinsic and intrinsic muscles of the lower limb. As with all joints, motion occurs by rotation about an axis in a plane of motion [45]. The three planes of motion in the foot are: sagittal plane (Sp), frontal plane (Fp) and transverse plane (Tp) as shown in figure 34. The foot (or any part of the foot) is considered as being:

- adducted when its distal aspect is angulated towards the midline of the body in the Tp,

- abducted when the distal aspect is angulated away from the midline (Figure 34 A),

- plantar flexed when the distal aspect is angulated downwards in the Sp away from the tibia,

- dorsiflexed when the distal aspect is angulated towards the tibia in the Sp (Figure 34 B),

- inverted when it is tilted in the Fp, with its plantar face towards the midline of the body,

- everted when its plantar faces away from the midline of the body (Figure 34 C),

- supinated when it is simultaneously adducted, inverted and plantar flexed,

- pronated when it is abducted, everted and dorsi flexed (Figure 34 D).

The talocalcaneal joint and the talocalcaneonavicular joint are included in the subtalar joint. Its axis of motion, known as Henke's axis, travels obliquely through the subtalar joint. Supination and pronation are the natural motions at this joint. The midtarsal joint is formed by the talonavicular and calcaneocuboid joints working together. This joint has two axes of motion, an oblique and a longitudinal axis, which results in forefoot supination/pronation movements.

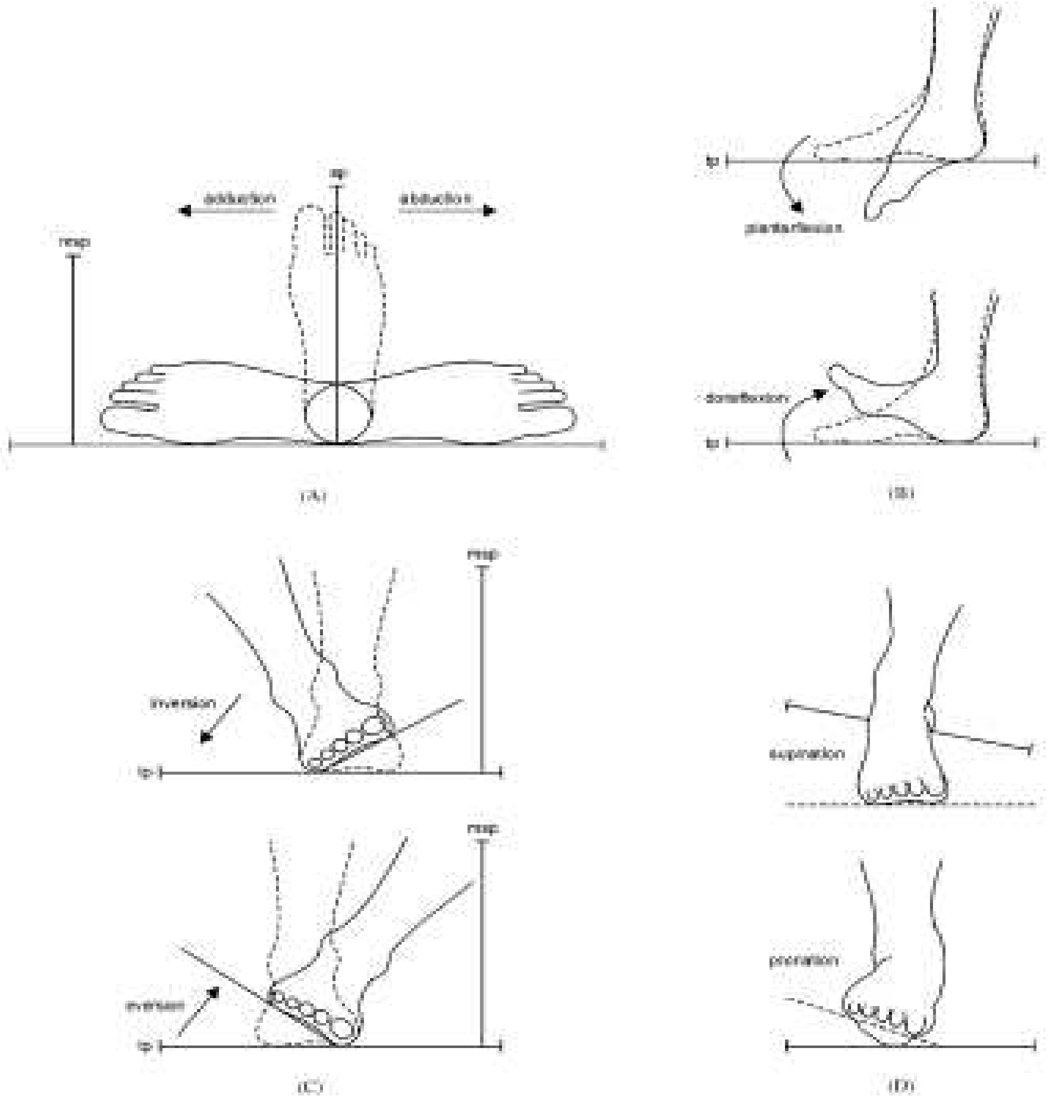


Figure 34: The movements of the foot-ankle region. (A) adduction-abduction, (B) plantarflexion-dorsi flexion, (C) inversion-eversion, (D) supination-pronation (Abboud, 2002)

Only the joint between the first metatarsal bone and the medial cuneiform exhibits significant movement at the interfaces between the metatarsal bones and the lesser tarsus. Up to 90 degrees of extension is achievable at the MTP joints, but only a few degrees of flexion. Each interphalangeal joint is extensible and flexile.

The ankle joint is the articulation between the distal portion of the tibia and the body of the talus, allowing the foot to dorsiflex and plantarflex. During plantar flexion, the ankle joint is also slightly mobile at the Tp, resulting in joint instability in this position.

All of the aforementioned definitions are the most prevalent. In the section under "Other aspects affecting foot-ankle mechanics," some extra information regarding nuances when describing foot-ankle movements, positions, or kinds will be offered.

Foot and ankle motion is the most essential technique for easing this pathway. Due to the lever arm of the heel, the ankle is lifted at initial contact but descends as the foot becomes plantar grade. At heel rise, the ankle is again lifted, a position that maintains throughout terminal stance and pre-swing. During stance phase, these ankle motions, which are coordinated with the knee and controlled by the pretibial and triceps surae muscles, smooth the route of the center of mass.

The controlled lever arm of the forefoot during pre-swing is especially advantageous because it softens the sudden downward reversal of the center of mass. Throughout stance phase, the foot and ankle motion support the course of the CG, maintaining it approximately horizontal.

Normal gait is characterized by Jacqueline Perry [10] in terms of the function of the heel, ankle, and forefoot rocker mechanisms (figure 35). The initial rocker is known as the heel rocker. This heel rocker preserves the momentum generated by the body's weight transfer to the stance leg.

The normal initial contact is formed by the calcaneal tuberosity, which serves as the pivot around which the foot and tibia travel. As the body's weight is transferred to the stance foot, the section of bone between the fulcrum and the middle of the ankle rolls toward the ground to preserve forward motion. Ankle rocker is the second rocker.

The pivotal arc of the ankle rocker propels the tibia over the stationary foot. Once the forefoot contacts the ground, the ankle becomes the fulcrum. Soleus eccentric contraction controls the advancement of the tibia. The third rocker is the rocker for the forefoot. The metatarsal heads become the pivot for tibial progress when the heel lifts. As body weight falls below the area of foot support, the rate of progression increases.

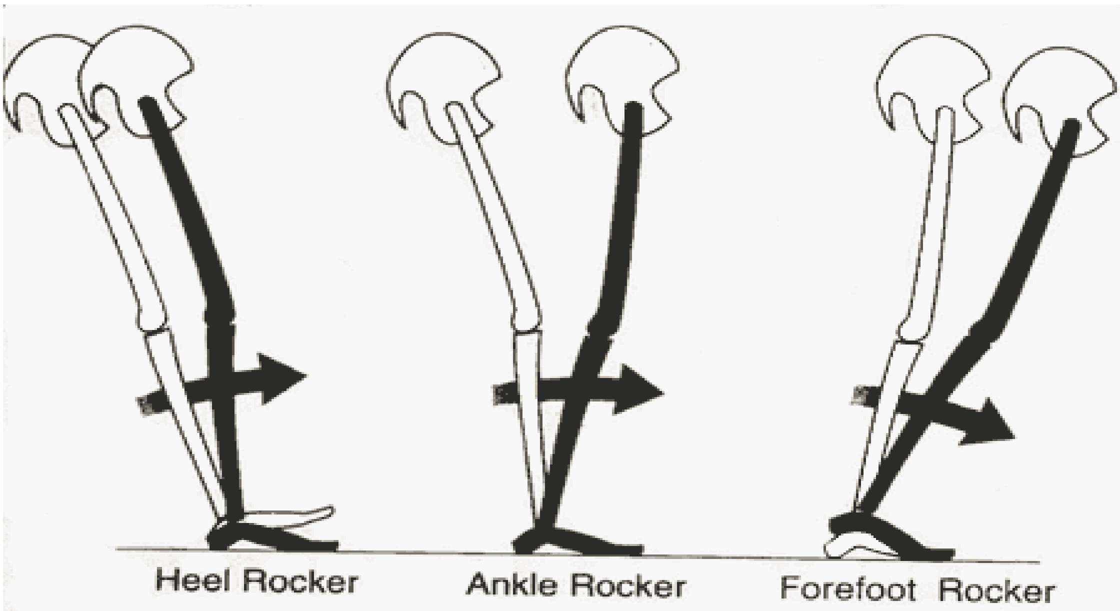


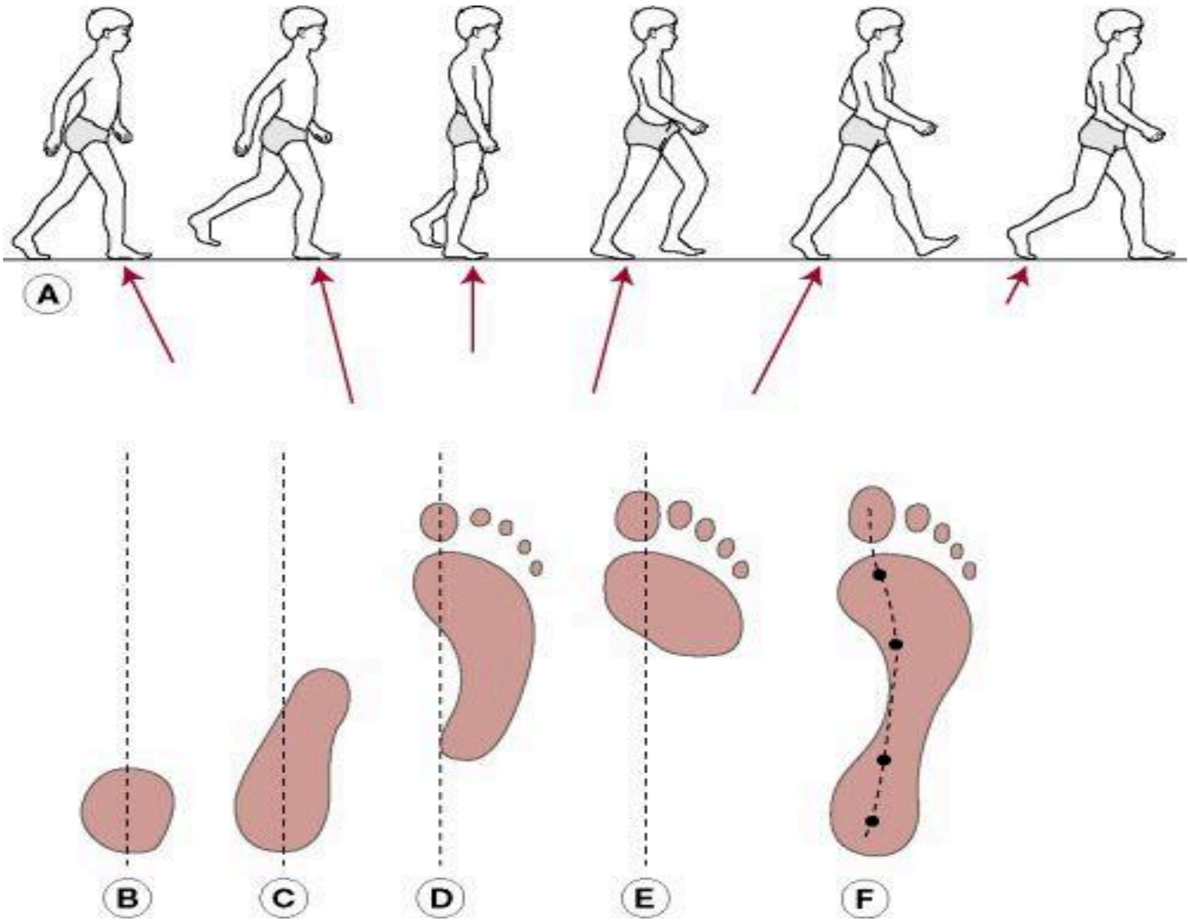
Figure 35: Ankle/foot rockers (Perry, J., 1992. Gait analysis: normal and pathological function)

For all of its activities, the foot requires enough mobility and stability. Mobility is required to absorb the body's ground response force. During first heel contact, subtalar pronation acts as a shock absorber. Pronation is also required to allow for leg rotation and to absorb the impact of such rotation. Through eccentric regulation of the supinators, subtalar pronation aids in shock absorption. Chopart's joint, on the other hand, becomes freed, allowing the forefoot to remain loose and flexible. The foot requires mobility in midstance to respond to surface variations.

To create a sturdy platform for the body, foot stability is required. The foot must be able to support the body's weight while also acting as a stable lever to move the body forward. This function necessitates subtalar joint pronation control.

Normal foot function [45] allows the foot to transition from a movable adaptor to a rigid lever at the appropriate time. The foot must have enough mobility to go into all of the gait cycle's postures while retaining mobility and stability. Physiological mobility is necessary because the foot would not be able to maintain stability if mobility was too great. The joint can support standing in the stable maximally close packed position when this requirement is met. Many overload injuries can occur when the usual transition between the two tasks is disrupted, such as in the foot, underleg, upper leg, and lower back. As a result, the three stages of ground contact must occur within the typical time interval; otherwise, compensatory mechanisms will be activated, potentially leading to overuse disorders.

The systems frequently fail during the transition from midstance to propulsion. The tibialis posterior muscle aids in the shift from eversion to inversion. The muscle is stretched out like a spring, and potential energy is stored as a result. The muscle switches from eccentric to concentric activity toward the conclusion of the midstance, and the energy is released. The tibialis posterior muscle then induces caput tali abduction and dorsiflexion, causing the hindquarter to evert. At the same time, the peroneus longus muscle will pull the forefoot forward with a plantar flexion of the first toe at the conclusion of the midstance. [18]



A: Beginning of the GC; B: Calcaneograde; C and D Plantigrade; E: Digigrade; F: Trajectory of the center of pressure during stance (Neale's Disorders of the Foot, Lorimer, 2006)

Figure 36: Foot support phases

The forefoot becomes stable in this manner. The metatarsophalangeal joints begin to dorsiflex, the plantar fascia is stressed, and the calcaneum becomes vertical and tips over in inversion when the forefoot moves in the propulsion phase. Differential timing of heel and forefoot contact with the floor results in three periods of foot support in the following order: heel, flat foot, and forefoot. The first metatarsal and the great toe are the last to make touch with the earth.

- Heel support or calcaneograde: this phase initiates stance (IC and loading response), the heel continues as the sole source of support until approximately 12% of the GC.
- Foot flat support or plantigrade: the forefoot contact through the 5th metatarsal head terminates the heel only support period and introduces a foot flat posture that goes from 12% to approximately 31% of the GC (mid-stance).
- Forefoot support or digigrade: this phase begins when the heel leaves the floor; the body weight is divided over the metatarsal heads. All the metatarsal heads are involved until the final support on the great toe. This phase goes from 31% to 60% of the GC (terminal stance and pre-swing).

As can be seen in figure 36, the center of pressure shifts forward along the length of the foot during stance in normal gait, moving from the heel to the forefoot before coming to rest on the first metatarsal and the great toe. This trajectory suggests that the ankle will bend plantarward, and that the foot will pronate, which will allow the center of pressure to shift forward and inward.

The windlass mechanism in the foot:

This mechanism in the foot was first described by John Hicks (1954) [12]. The foot and its ligaments were first described by Hicks as an arch-like triangular structure. The truss' arch is made up of the calcaneus, midtarsal joints, and metatarsals, as well as the medial longitudinal arch. As depicted in figure 37, the plantar fascia serves as a tie-rod that connects the calcaneus to the phalanges. The vertical pressures of the body weight move downward along the tibia, flattening the medial longitudinal arch. Ground-reaction forces push up on the calcaneus and metatarsal heads at the same moment, further flattening the tibia because these forces fall both anterior and posterior to it.

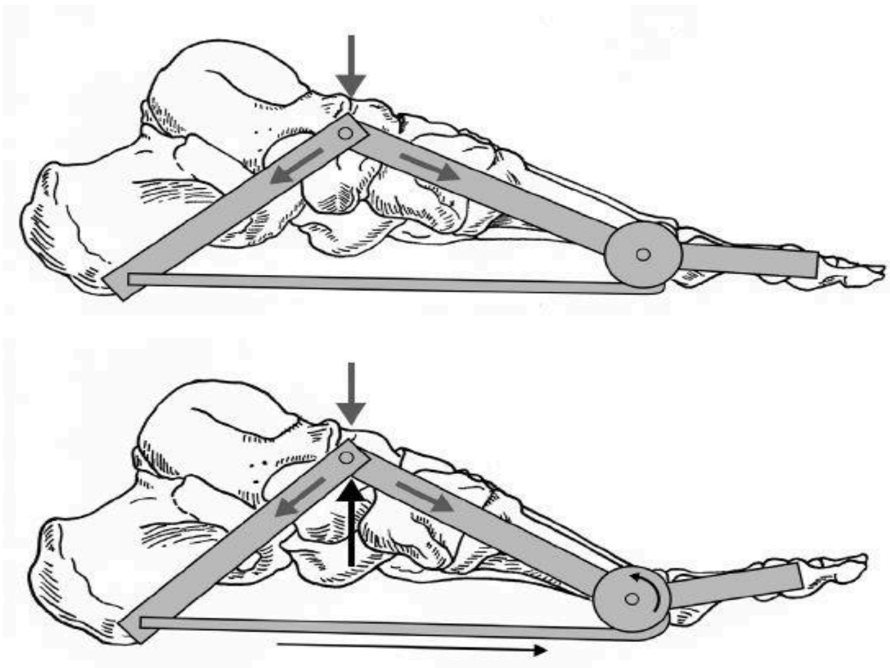


Figure 37: Windlass mechanism (Hicks JH. The mechanics of the foot, II: the plantar aponeurosis and the arch. J Anat. 1954)

The plantar fascia, which extends from the base of the calcaneus to the phalanges, inhibits foot collapse due to its anatomic placement and tensile strength. The plantar fascia's stretch tension keeps the calcaneus and metatarsals from expanding and maintaining the medial longitudinal arch. At heel strike, the GC begins with the foot in a supinated position. When transitioning from heel strike to weight acceptance, the subtalar joint quickly pronates. Increased foot movement is required to absorb ground reaction forces and adjust to uneven terrain during this stage of pronation.

Following the weight-acceptance phase, the foot reaches maximum pronation, and the subtalar joint supinates the foot from midstance to toe-off. Supination converts the foot into a stiff lever that is required for propulsion. The plantar fascia's direction aids in maintaining the arch throughout stride and contributes considerably to the proper timing of pronation and supination during the gait cycle.

Pronation increases the space between the calcaneus and the metatarsals from heel strike to weight acceptance, putting tension on the plantar fascia. Supination occurs from midstance to the propulsive phase, allowing the foot to become a rigid lever that can move the body forward utilizing the windlass mechanism. Supination forces exert tension to the plantar fascia in the same way that pronation forces do. Plantar fascia tension is increased by the forces exerted during pronation and supination. Increased tissue stress can be caused by inefficient foot function [38].

Recent research has shown new paradigms when it comes to the role of the plantar fascia and the windlass mechanism. It is commonly said that the rigidity of the human foot is a trait of our evolution for upright walking. This rigidity is supported by a thick plantar aponeurosis that connects the heel to the toes. In previous accounts of the function of the human foot have considered stretching the plantar aponeurosis via toe extension (also known as the windlass mechanism) in order to stiffen the foot when it is levered against the ground in preparation for push-off during walking. Recent work from Kelly and Farris [46] indicated that the tensioning and stiffening of the foot that is seen with higher foot-ground contact pressures and push-off effort could not be explained by plantar aponeurosis strain via the "windlass mechanism." Instead, electromyographic recordings revealed that the source of tension in the plantar aponeurosis comes from active ankle plantar flexor contractions. Additionally, the produced tension along the plantar portion of the foot was being influenced by the plantar intrinsic foot muscles. Their conclusion was that the foot's major source of rigidity for push-off against the ground during bipedal walking is active muscular contraction, not the passive windlass mechanism.

They also discovered that as the arch-spring collects and releases elastic energy during locomotion, the plantar fascia also extends and then shortens [47]. They found that toe plantarflexion delays the stretching of the plantar fascia at foot striking, which possibly changes how the stress is distributed across other arch tissues. A quasi-isometric plantar fascia's shortening is postponed to later in stance by a pure windlass effect in propulsion. After that, the windlass mechanism shortens the plantar fascia, which probably increases arch recoil during push-off.

4. Running biomechanics

When periods of double support during the stance phase of the gait cycle [48] (both feet are simultaneously in contact with the ground) give way to two periods of double float (Fig. 38) at the beginning and end of the swing phase of gait, the transition from walking to running occurs (neither foot is touching the ground). Despite the fact that modern "grounded" running suggests that certain types of runners may have a brief double support phase. The majority of grounded running gaits occur at speeds (2 to 3 m s-1) where neither running nor walking is "pleasant" or energetically efficient. As speed rises, the initial contact typically shifts from the hindfoot to the forefoot.

As mentioned previously, the gait cycle is the time between the initial contact of one foot and the initial touch of the same foot. It is comprised of a stance phase, during which the foot is in contact with the ground, and a swing phase, during which the foot is not in contact with the ground.

Other useful characteristics of gait include stride length, step length, and cadence. The distance between the initial touch of one foot and the initial contact of the opposite foot is the stride length.

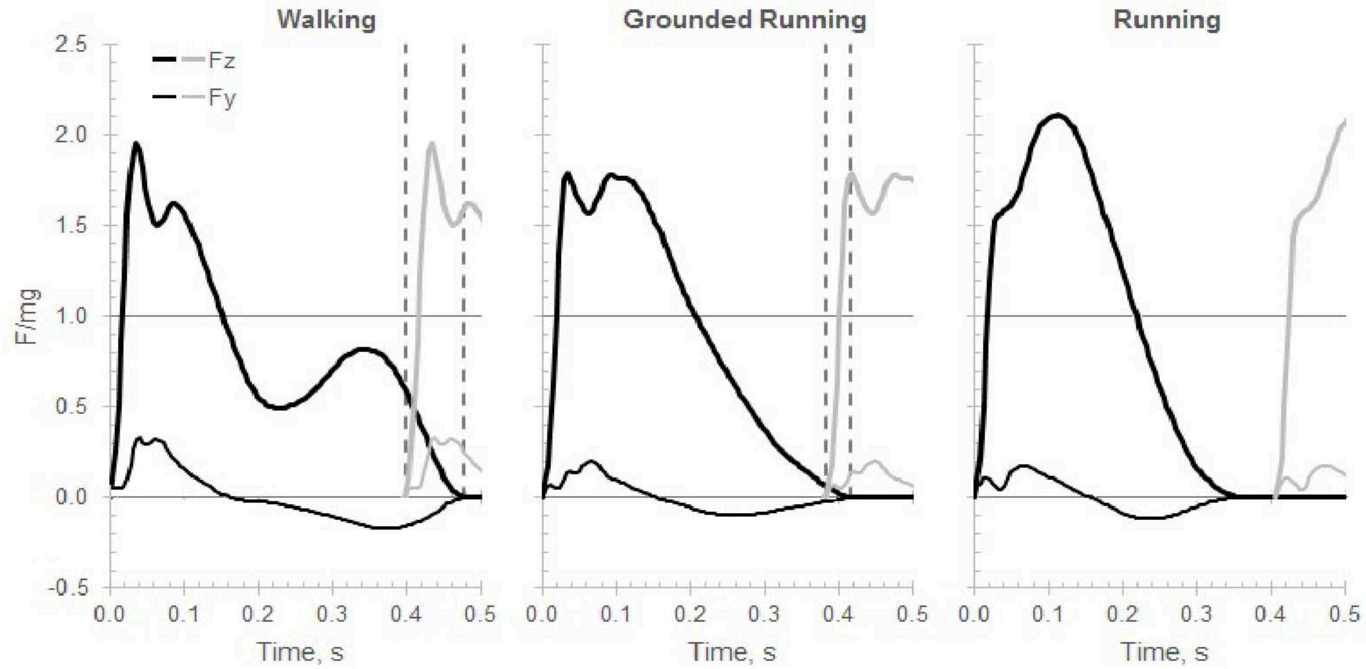


Figure 38: Vertical and anterior-posterior forces during walking, grounded running and running. Dashed lines indicate periods of double support. (Bas Kluitenberg, 2012)

Step length is the distance between the initial contact of one foot and the initial contact of the same foot, or the distance covered in a single gait cycle. Cadence is the number of steps performed in a particular amount of time. Normal walking can be defined as double support, when both feet are in touch with the ground, and single support, when only one foot is on the ground. In a walking gait cycle, the stance phase typically accounts for 60 percent and the swing phase for 40 percent.

During the early and last 10 percent of the stance phase, double support occurs. Single support corresponds to the swing time period of the contralatera extremity by definition. The stance phase is further broken into four stages, beginning with initial limb contact. The initial phase is the loading reaction, followed by the midstance, terminal posture, and preswing phases. The swing phase begins at toe-off and consists of initial swing, midswing, and terminal swing.

As this is the performance benchmark for an athlete, running is typically defined by its velocity, which makes empirical sense. Running is distinguished from walking by its float phase, during which the runner gains aerial momentum.

As seen in figure 40, the running gait cycle is also separated into a stance and swing phase. The stance phase is broken into the absorption and propulsion phases, while the swing phase is subdivided into the starting and terminal swing phases. At the beginning and end of each swing phase, neither limb makes contact with the ground during a double-float interval. In order to accommodate this airborne interval, the stance phase must account for less than fifty percent of the gait cycle. Initially, increased velocity is accomplished by raising the step length, and then by increasing the cadence. With an increase in velocity, the stance phase is shortened and the swing time is lengthened, resulting in a longer double-float phase.

As velocity increases from jogging to sprinting, step length and cadence also increase. In addition, the sprinter initially strikes the ground with his toes as opposed to his mid- or hindfoot; Figure 39 depicts the regular running cycle.

Approximately 40% of the running gait cycle is spent in the stance phase, while 60% of the running gait cycle is spent in the swing phase. Due to the increased range in velocity, these values are more variable in running.

The average step length increased by 20%, although the gait cycle time dropped by nearly one-third (0.9 to 0.6 seconds). Approximately a three-quarters increase in cadence and a near doubling in velocity.

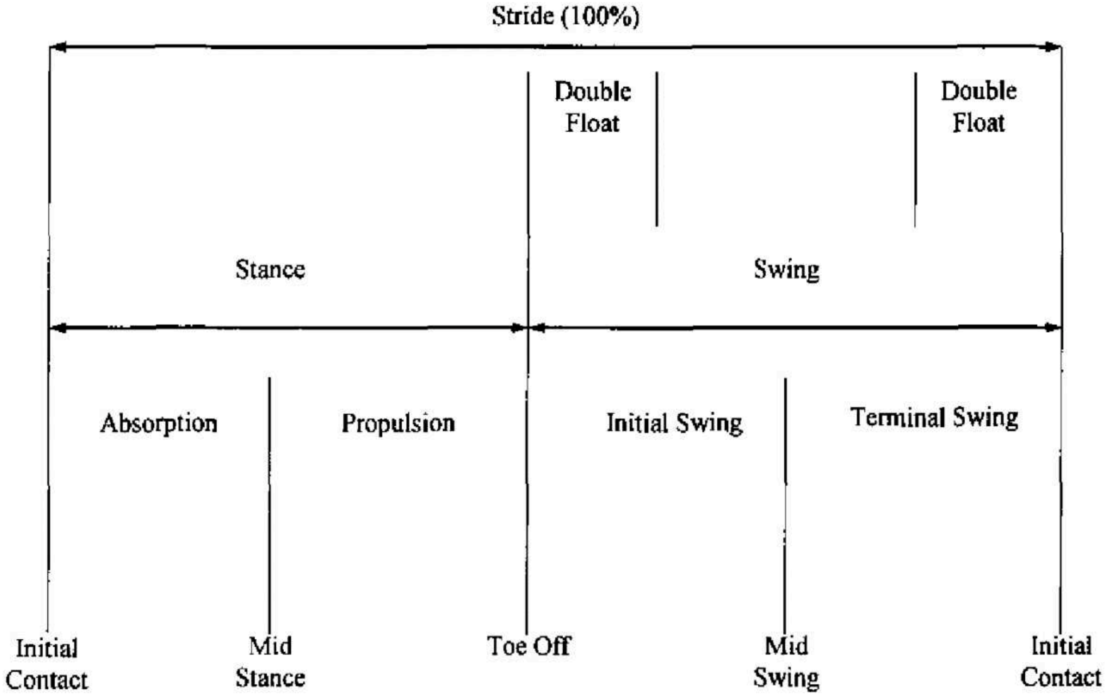


Figure 39 : The events of the running gait cycle (Sheila A Dugan, 2005)

Kinematic is defined as the description of motion of joints or body segments that occurs independent of forces that cause the motion to occur [49]. The motions of all of the joints in the lower extremity are interrelated and occur in a similar fashion with each gait cycle. For instance, the pelvis, femur and tibia initially internally rotate during stance phase, which corresponds with eversion and unlocking of the subtalar joint during walking. This is followed by external rotation of the pelvis, femur, and tibia, which is linked to inversion of the subtalar joint.

The inversion of the subtalar joint locks the hindfoot in a rigid position, thus stabilizing the foot for toe-off. The pelvis, femur, and tibia again internally rotate during the swing phase of walking, which facilitates eversion of the subtalar joint, placing the foot in a supple position to act as a shock absorber at heel strike, which starts the next stance phase. The concept of a mitered hinge to describe the linkage of internal and external rotation of the tibia with eversion and inversion of the hindfoot, respectively, was popularized by Mann, eversion of the subtalar joint, which results in pronation of the forefoot, the foot becomes supple because the axes of the calcanocuboid and talonavicular joints become parallel, thus allowing increased motion at the transverse tarsal joint. Other factors stabilizing the foot during the heel rise and toe-off portions of gait are the plantar fascia and the metatarsal break.

The plantar fascia augments the arch and inverts the heel as the toes are dorsiflexed during the latter portion of the stance phase. The oblique orientation of the metatarsophalangeal joints is referred to as the metatarsal break. This orientation promotes hindfoot inversion during toe-off, thus, facilitating external rotation of the leg due to the mitered hinge linkage of the tibia to the hindfoot. This increase in inversion increases the rigidity of the foot to facilitate toe-off. The kinematics of running differ substantially from those of walking. In general, joint range of motion increases significantly as velocity increases. The body lowers its center of gravity with increased speed by increasing flexion of the hips and knees and by increasing dorsiflexion at the ankle joint. Both the knee and ankle flex during the absorption phase of stance during running. At the time of heel strike, rapid dorsiflexion also occurs at the ankle, along with the hip and knee flexion, which helps absorb the force of impact.

This is in contrast to walking, where plantarflexion occurs at the ankle initially at heel strike. At the midstance of running, knee and ankle motion reverse and both begin extending, which begins the propulsion phase. Although the hip also demonstrates an increased range in motion with increased velocity, the increase of motion is in flexion and the degree of extension actually decreases slightly. At toe-off, the body is preparing to go into the double float phase; therefore, the hip, knee, and ankle are all at maximal extension. Another major difference between walking, running, and sprinting is the manner in which the foot contacts the ground. In walking, the heel contacts the ground first, followed by a foot-flat stance phase. During running, the majority (approximately 80%) of runners (rear foot or heel strikers) make initial ground contact along the posterolateral border of the foot. The other runners initially contact the ground with the midlateral border of the foot (midfoot strikers). Sprinters initially contact the ground with their toes. Upon initial contact the hindfoot is in slight inversion, but rapidly migrates into eversion to act as a shock absorber.

This eversion, or pronation, of the foot occurs in 30 milliseconds, which is five times faster than during walking. Runners with rigid cavus feet generally absorb forces more poorly than runners with normal feet because they are unable to pronate their feet to function as shock absorbers. Interestingly, running barefoot normally causes increased pronation, which may be secondary to the need for the lower extremity to absorb force that would otherwise be dissipated by the shoe. Any alteration in the normal kinematics of the lower extremity can lead to decreased force dissipation and increased stresses, thus increasing the risk of injuries, such as stress fractures.

Kinetics is defined as the study of the internal (muscle forces) and exterior (earth reactive forces) factors that generate movement [49]. Electromyographic activity quantitatively measures muscle activity. Compared to walking, running results in an increase in the electromyographic activity of all examined muscles. In addition, muscles are often active for a bigger part of a running gait cycle. During walking, electromyography reveals that the quadriceps become active in the final 10% of the swing phase and remain active in the first 15% of the stance phase. During running, the quadriceps become active during the final 20% of the swing phase and stay active during the first 50% of the stance phase. Throughout sprinting, the quadriceps are engaged during the first 80% of stance and the last 50% to 60% of swing phase. After toe-off, sprinting action continues during the balance of the swing phase and the first 80 percent of the stance phase.

In general, the knee joint acts during walking and running by increasing knee flexion during impact absorption and then knee extension during propulsion. An assessment of the lower leg muscles reveals those muscles prior to the ankle (anterior compartment muscles) act as one unit, but muscles posterior to the ankle (posterior compartment muscles) function as a separate unit. 2 The posterior calf muscles are active throughout approximately the middle 50 percent of walking's stance phase. During running, the posterior calf muscles increase their activity so that they are active during 80% of the stance phase and 25% of the swing phase. In addition, when running while shortening to plantarflex the foot, it is estimated that the posterior compartment muscles exert a tension equivalent to 250% of body weight. Similar to the posterior compartment muscles, the anterior compartment muscles become engaged shortly before to toe-off and stay active during the swing phase and the initial 10% of stance phase. During the stance phase of running, the ankle actively dorsiflexes after heel strike.

In contrast to the eccentric contraction of the anterior compartment muscles following heel strike in walking, which decelerates the foot when the initial plantar flexion occurs in the stance phase, the anterior compartment muscles contract concentrically during running to stabilise the ankle and assist in dorsiflexion. Thus, one can observe that when velocity increases, the front and posterior compartment muscles' phasic activity reverses as their activity changes. At contact, therefore, both the anterior and posterior leg compartment muscles activate, resulting in a stable foot. Ground response forces are the forces exerted on the foot by the ground when the foot makes contact with the ground. There are three components to the force: forward-backward, medial-lateral, and vertical. The vertical ground response force is the largest of these. During the loading and push-off phases of gait, the vertical component often exceeds 1.3 to 1.5 times body weight. The magnitude of the vertical ground response force can approach three to four times human weight during jogging. The fore-aft ground response force corresponds to deceleration during the initial 50% of stance followed by acceleration during the propulsive period of the last 50% of stance. The amplitude of medial-lateral forces is low during both running and walking in normal persons.

In addition to being roughly double that of walking, the vertical ground reaction force felt when running occurs at least twice as quickly. Therefore, the forces exerted on the supporting tissues and skeleton would indicate a fourfold increase in strain, which explains why running is significantly more likely to cause symptoms due to accumulated microtrauma than walking. Runners with certain foot shapes are more susceptible to injury. Studies indicate that feet with a high arch (cavus) concentrate pressure beneath the heel and forefoot. There is no midfoot loading, despite the increased loading circumstances associated with greater running speed. Normal, flexible feet demonstrated pressure dispersion, including in the midfoot region.

These results demonstrate why runners with cavus feet are more susceptible to injury. Others have established through pedobarographic research that patients with a long second metatarsal (Morton's foot) have significantly greater second metatarsal head pressures. It was hypothesised that these individuals were more susceptible to second metatarsal pressure issues, such as metatarsalgia or stress fracture. Various authors have identified differences between runners who are rearfoot strikers (contacting the ground with the posterolateral border of the foot) and midfoot strikers (contacting the ground with the medial border of the foot) (contacting with the midlateral border of the foot). It has been observed that forefoot strikers have a shorter average foot contact time than rear-foot strikers. Midfoot strikers do not exhibit the same dramatic initial peak in the vertical ground response force as rear-foot batters. Cavanagh and LaFortune assessed the centroid of pressure, which is the instantaneous average pressure point in runners.

In rear-foot strikers, the centroid migrated from the back lateral border of the foot to the midline in the centre of the forefoot in a gradual manner, with the centroid localised over the forefoot for about two-thirds of the stance phase. In contrast, midfoot strikers running at the same speed made initial contact with their midfoot, followed by a quick shift to their forefoot. Although these changes were observed between midfoot and heel strikers, the authors did not examine their clinical importance or level of function difference. It has been shown that footwear reduces peak vertical ground reaction forces and modifies the path of the centre of pressure. During the initial 25 milliseconds after foot contact when running, a rapid loading rate has been observed. This has been suggested as a factor in joint deterioration. Therefore, appropriate footwear can reduce the forces to which the foot is subjected.

In running, the hip and knee joints consistently reverse power generation and absorption [7]. During the absorption phase of stance, the beginning swing phase, and the final swing phase, the ipsilateral knee absorbs energy while the contralateral hip generates energy. During the initial swing phase, for instance, the hip flexes and generates force by concentric contraction of the rectus femoris, whereas the knee flexes and absorbs force through eccentric contraction of the distal rectus femoris. During the final portion of the swing, the hamstring muscles fulfil a similar job by contracting concentrically at the hip and eccentrically at the knee. Consequently, during running, the double-joint muscles (rectus femoris and hamstrings), which both bridge two joints, contract concentrically at one end and eccentrically at the other end, allowing for an effective transfer of energy across joints. This phenomenon is not observed while walking.

During walking, the body's centre of gravity is highest when walking slowly and lowest when walking quickly [7]. Thus, potential energy is highest during the single support phase (i.e., when the centre of gravity is highest), while kinetic energy is lowest, and potential energy is lowest during the double support phase (i.e., when the centre of gravity is lowest), when kinetic energy is highest. As a result, walking has a relatively low energy cost. Running involves a distinct form of energy conservation.

During the double-float phase of running, the centre of gravity is at its highest point (i.e., highest potential energy) when the body is going the quickest (i.e., highest kinetic energy). In contrast, the centre of gravity is lowest during the absorption phase of gait (lowest potential energy), when the body is moving the slowest and has the lowest kinetic energy; however, Cavanagh and LaFortune' estimate that 50% of the total work performed in running is due to the storage of energy as elastic strain. This is due to the fact that power production via concentric muscular contraction is always preceded by power absorption (i.e., eccentric contraction of muscles). The role of the double-joint muscles (rectus femoris and hamstrings) as an essential energy-saving component has been explored earlier.

The footstrike pattern is defined as a biomechanical analysis of the way the foot touches the ground. A common interpretation of this definition uses the distinction of three footstrike patterns (Fig. 40): forefoot strike (FFS), midfoot strike (MFS), rearfoot strike (RFS) [7].

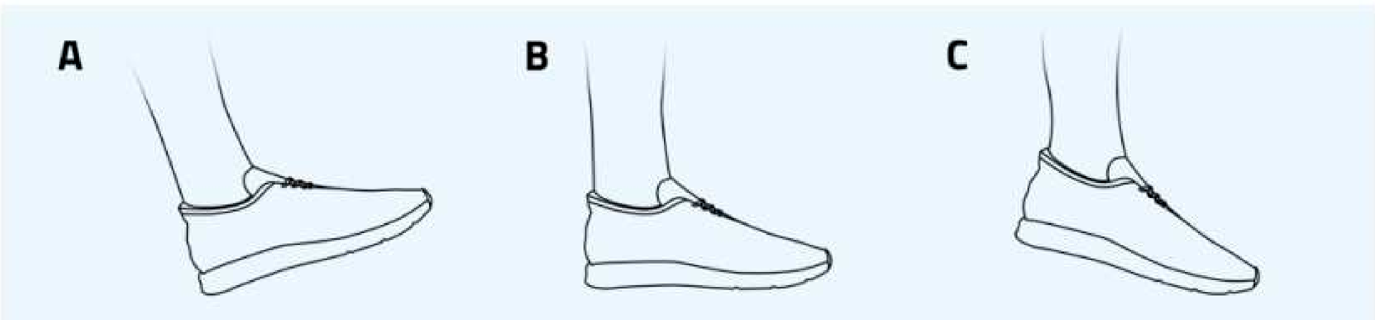


Figure 40 : A: RFS, B: MFS and C: RFS (T Hoenig, 2020)

This popular classification has undergone several changes. Some authors, for instance, distinguish between an FFS in which the toe strikes the ground before the heel and an FFS (so-called toe strike) in which the heel never reaches the ground. The footstrike angle (FSA) is a measurement of the initial contact angle between the foot and the ground. Instead of identifying nominal variables (FFS, MFS, RFS), the exact footstrike angle is measured (continuous variable). Two markers can be placed on the shoe or foot of the athlete: (A) the calcaneus and (B) the head of the fifth metatarsal. During stance and initial ground contact when running, the angle between the vector AB and the ground is calculated and measured. In these instances, the (sole) angle between the ground and the running shoe's sole at initial ground contact can be utilised to calculate FSA (Fig. 41).

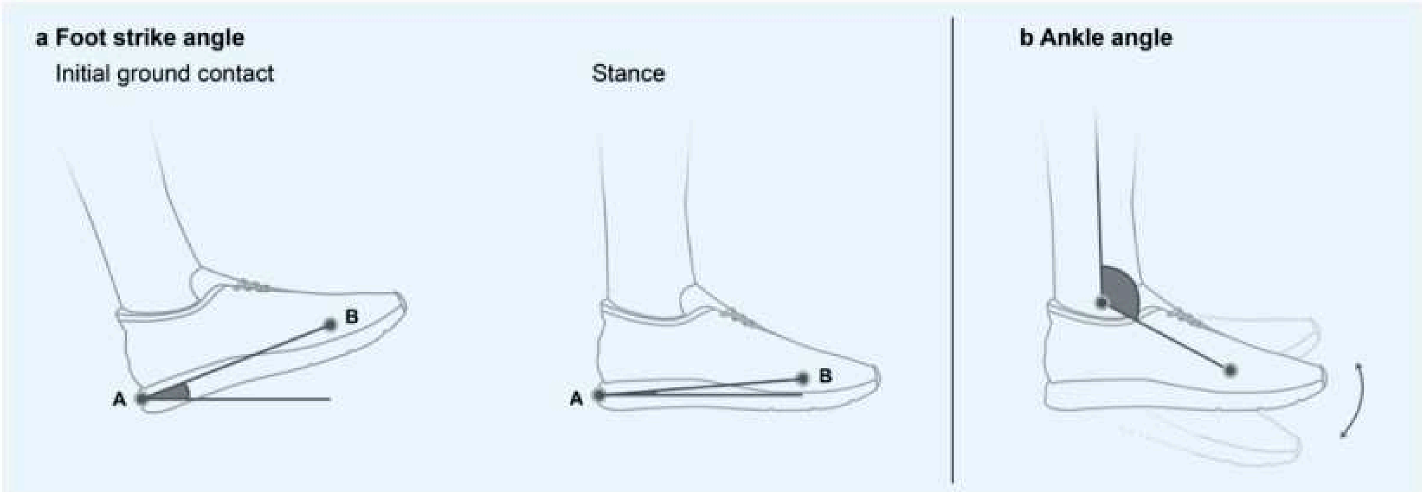


Figure 41 : Foot Strike Angle (T Hoenig, 2020)

To classify FSA under the traditional footstrike definitions (FFS, MFS, RFS), Altman & Davis used the following conversion:

- FFS: FSA< -1.6°
- MFS: -1.6° <FSA< 8.0°
- RFS: FSA> 8.0°

As stated previously, the FSA is determined using a foot segment and the earth horizontal. In contrast, the ankle angle is determined using a foot vector relative to a shin vector. As with the FSA, measurements of the ankle angle typically need at least two-dimensional sagittal plane kinematics. FFS and MFS are distinguished by a more plantarflexed foot position (e.g. ankle angle below 0° relative to a neutral standing posture) at touchdown, whereas RFS is distinguished by a more dorsiflexed foot position (e.g. ankle angle above 0°). It can be assumed that the determination of the ankle angle in runners is accurate. To the best of our knowledge, this method has not yet been validated for classifying footstrike patterns. Marker placement is prone to inaccuracy, thus the examiner should be knowledgeable. Garofolini et al. presented a novel classification that incorporates the vertical ground reaction force (vGRF) measurement and joint kinetics.

The ankle moment was determined during an initial contact period characterised by an increase in vGRF. In RFS, the rearfoot makes contact with the ground first. Following first contact, the forefoot transitions to a flat foot position (plantarflexion movement).

The instant of external plantarflexion is countered by an early moment of internal dorsiflexion. Thus, athletes with predominantly internal dorsiflexion moments were categorised as RFS. In FFS, the forefoot makes contact with the ground first.

The moment of external dorsiflexion is matched by the moment of internal plantarflexion. Thus, athletes with predominantly internal plantarflexion moments were classified as FFS. This classification may be advantageous because it more accurately reflects force vectors during footstrike and the unique anatomy of each foot.

The cyclogram of the stride describes the relative movement of the distal ends of the upper and lower limb segments relative to the shoulder and the hip. Indeed, if we observe the trajectory of the foot or the ankle, we note that they describe a repetitive movement of a reel with each stride such. This "circular" movement allows the translation of the body as a whole.

The cyclogram represents the result of the mobilization of three joints: ankle, knee and hip, corresponding to the shape of the stride. It allows in particular to characterize the anteroposterior distribution of the stride, that is to say to differentiate anterior cycle and posterior cycle as shown in figure 42.

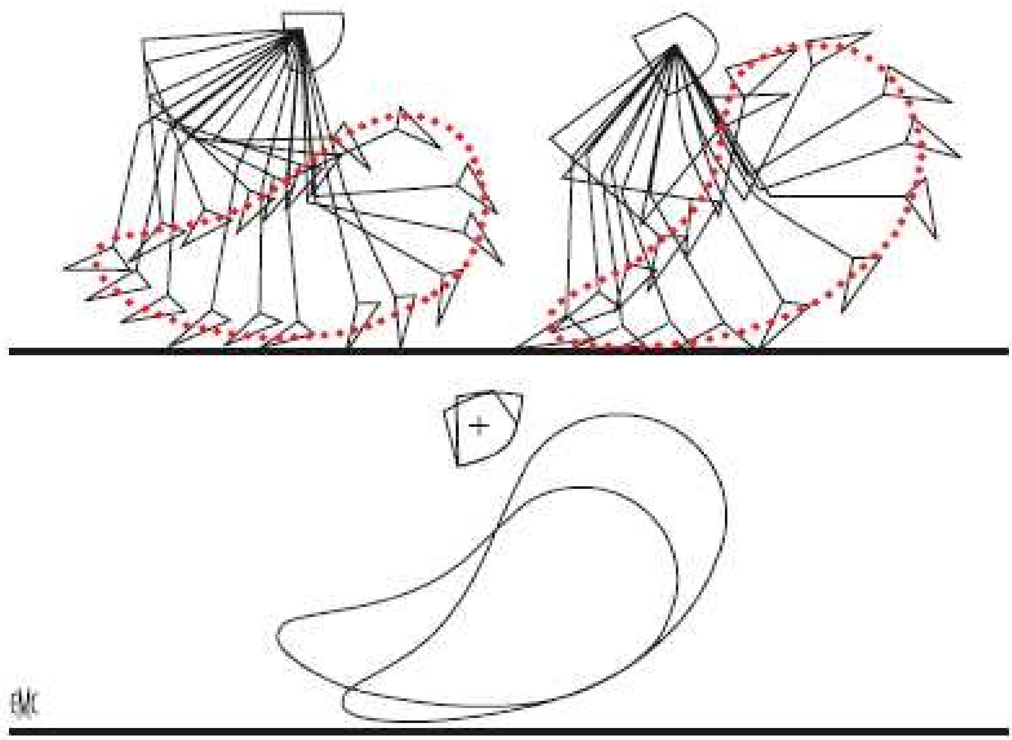


Figure 42: Representation of the poulaine-shaped cyclogram for anterior running cycle (left) and posterior running cycle (right). (Leboeuf et al., 2006)

As the FSI relies on force/pressure data, an instrumented treadmill or sensor insole are typically required for its application. Nonetheless, indirect measurements with inertial measurement units (IMUs) have become a promising technique in recent years. A RFS can be derived based on the FSI when the COP is positioned between 0 and 33 percent of the total foot length (Fig. 43). A COP between 34% and 67% denotes an MFS, but a COP between 68% and 100% denotes an FF. The FSI is frequently utilised to evaluate shod and barefoot runners.

The significance of the FSI is that it can reveal how the ground reaction force vector affects the foot. Repetitive forces that are applied to the human body are regarded as a primary cause of overuse injuries associated with running. The FSI may be called the "gold standard" in the analysis of footstrike patterns not just for this reason. In contrast to other categories, the FSI does not adequately address the musculoskeletal system of the foot during foot strike.

Consequently, few academics recommended using values other than the COP. In addition to the physical state of the ankle, they advised taking the vGRF into account. This concept assumes that RFS runners often experience a higher impact peak than MFS/FFS runners. The assessment of footstrike patterns based on impact peaks cannot, however, be considered a conventional process.

Currently, instrumented treadmills are expensive and restricted to biomechanical laboratories, making them impractical in a real-world situation. This may change as wearable technology (e.g., shoe insoles with the ability to record plantar pressure) becomes more popular.

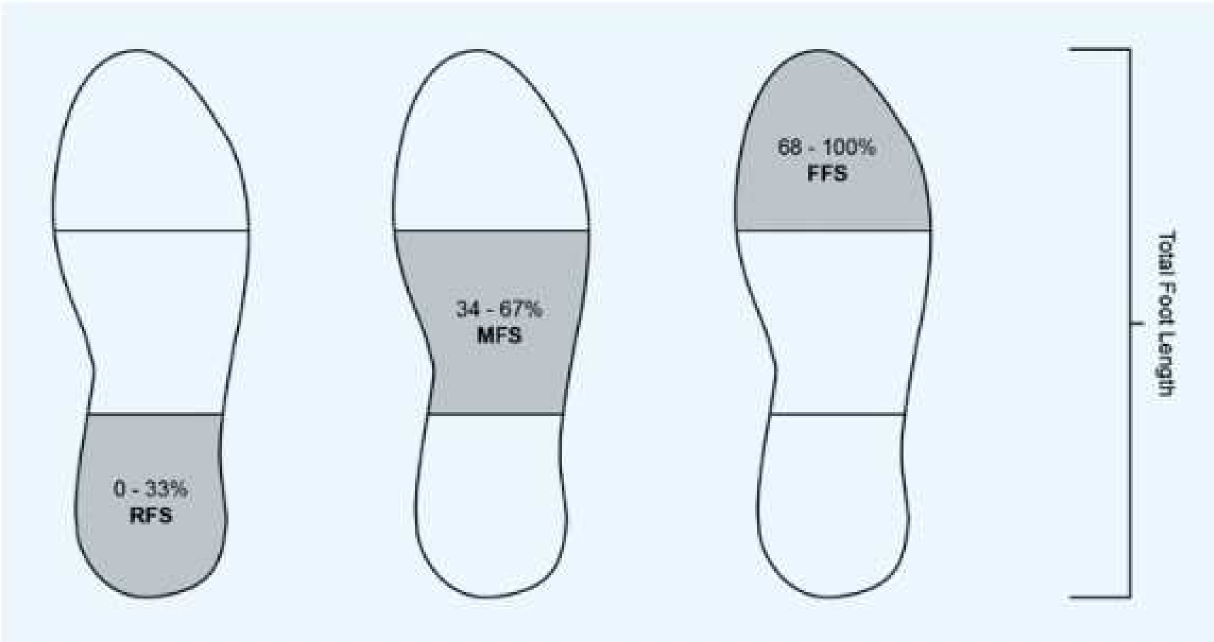


Figure 43 : Foot Strike Index (T Hoenig, 2020)

Nearly all ailments of the lower limbs have been attributed in sports medical literature to excessive pronation (and the spine, for that matter). It is believed that improper movement of this joint over thousands of repetitive cycles results in overuse symptoms due to increased internal rotation of the tibia as a result of the mitered hinge effect. This theory is supported by a substantial amount of clinical evidence, as shoes or orthotics designed to reduce hyperpronation typically eliminate painful symptoms.

Several review papers on foot and ankle biomechanics investigate the normal and pathological mechanics of hindfoot and midfoot motion. Roger Mann has been a pioneer in educating biomechanists and medical professionals in this field. Combining two distinct areas of research, Czerniecki's review links foot and ankle biomechanics with Perry's three stance phase ankle rockers. Due to the oblique orientation of the ankle joint, sagittal plane ankle motion is accompanied by rotation in the transverse plane and rotation of the foot about its long axis (figure 44).

Ankle dorsiflexion promotes internal rotation of the tibia and pronation of the foot during walking, while the foot is fixed to the ground. In addition to having an oblique axis of rotation, the subtalar joint is also responsible for the complex pronation/supination of the foot in relation to the tibia. Again, rotational torques around the longitudinal axis of the foot are communicated to the tibia via the mitered hinge effect, resulting in rotational torques about the tibia's longitudinal axis.

The hindfoot is often inverted upon initial contact. As the limb is thus loaded during the absorption period, pronation ensues. Pronation "unlocks" the transverse tarsal joint, enhancing the foot's flexibility and allowing it to absorb trauma more effectively. Typically, peak pronation occurs at 40 percent of stance phase.

At 70 percent of stance, the foot begins to supinate and reaches a neutral position. Then, the transverse tarsal joint is "locked." The generating phase has been achieved, and the foot is now stiffer, enabling it to act as a lever for push-off more effectively. The 'hyperpronator' may not begin to supinate or attain a neutral position until much later, well after power generation was scheduled to commence. In this situation, the foot would be ineffective as a lever [17].

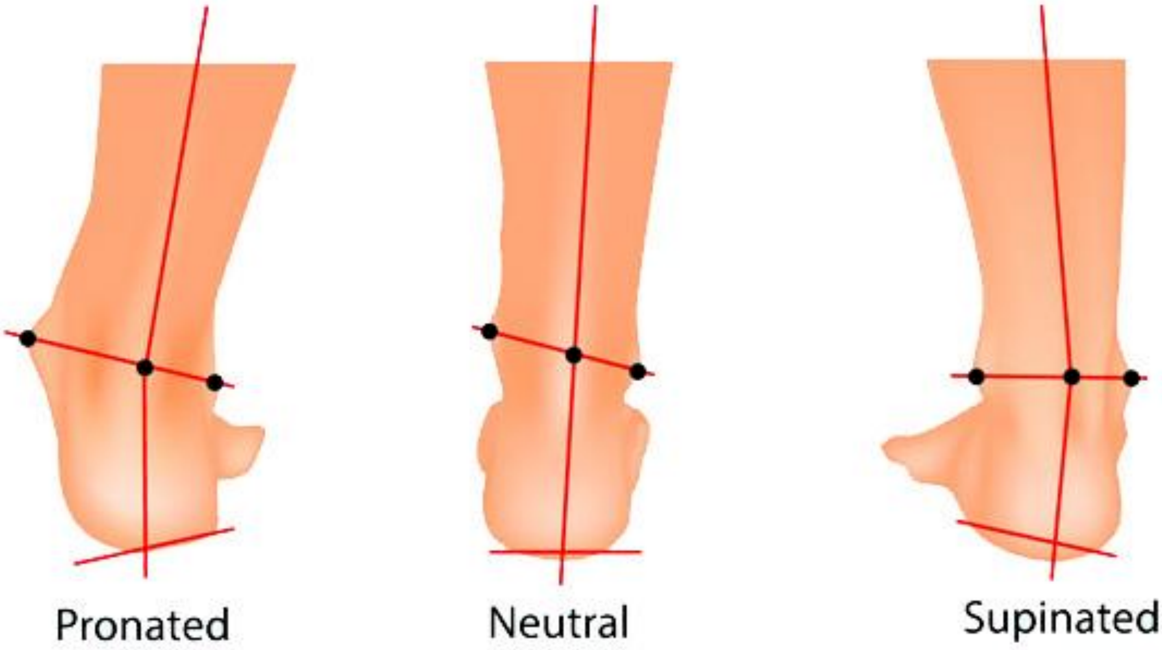


Figure 44 : Foot biomechanics (https://www.healthline.com)

To conclude this chapter, we have covered the functional anatomy of the HLS, all the basic aspects of gait (walking and running) and the concepts of the foot function which is considered as podiatric biomechanics. This leads on to the next chapter where we will get a deeper understanding of running gait analysis, fatigue, impacts and loading rate and how to gauge them.

CHAPTER 3

Material, methods and results

In this chapter we will first explore the current modern methods used to collect data on running motion. We will also describe how fatigue, impacts and loading rate may hamper a runner's performance and health. We will then go in depth in the protocols we implemented to curate date on athletes. Finally, we will explain the different steps we used to get to the calculation of the Shock Response Spectrum (SRS) of our runners. This last step is a novel approach to running gait analysis and a potential first step towards a predictive method to avoid overuse injuries.

1. Running gait analysis

The systematic study of human locomotion is known as gait analysis. This study makes use of the eye and the brain of observers, in addition to instrumentation for monitoring body movements, body mechanics, and the activity of the muscles [50]. Individuals who suffer from conditions that impair their walking abilities are evaluated and treated with the use of gait analysis. It is also extensively used in the field of sports biomechanics to assist athletes in running more effectively and to discover posture-related or movement-related difficulties in those who have sustained injuries. Quantification, which involves the introduction and analysis of measurable parameters of gaits, and interpretation, which refers to drawing various conclusions about the animal based on its gait pattern (such as its health, age, size, weight, and speed, among other things), are both included in the study.

Research and analysis of running gaits are becoming increasingly popular as the number of people participating in running events increases. This capability is currently being made more generally available to a wider variety of experts as a result of recent advancements and proliferations in the technology utilised in the study of running gait. Analysis of a runner's gait is no longer restricted to clinical gait analysis and research laboratories that have adequate funding. Attending a major running competition, such as the Boston Marathon, is all that is required to observe firsthand the application of cutting-edge technology that is able to deliver real-time feedback on the motion of the runner's rear foot while running and perform quantitative analysis of the arch structure in order to recommend the most suitable running shoe for a specific person. It's possible that the utility of these technologies is even more significant than the fact that they are readily available. The data collection technology that is utilised for the study of running gait is continuing to become more portable, inexpensive, and compact. Additionally, the gathering of this kind of data is no longer limited to research or clinical settings.

The dramatic rise in the number of people running for recreation and competition over the past few years has had obvious repercussions for professionals such as clinicians, physical therapists, and coaches who provide services aimed at the evaluation and rehabilitation of running-related injuries and performance enhancement strategies. These professionals have seen an increase in business as a direct result of this rise. Recent advances in technology have made it possible for a wider variety of people to currently perform these services. It is possible that clinicians will become increasingly involved in the prevention and rehabilitation of running-related injuries as new information regarding the processes of damage becomes available. Because of the rise in demand for these services, it is becoming increasingly important to have a solid understanding of the many forms of technology that are now on the market. As a result, the purpose of this review is to provide a concise summary of the technologies that are now utilised in the analysis of a runner's gait, with an emphasis on the most recent innovations and pieces of equipment.

The analysis of a runner's gait can be made easier with the use of a number of different technologies. These include the more traditional motion capture systems that are used to describe motion of the body, force plates that quantify the forces acting on the body, and electromyography (EMG), which is used to estimate the level of muscle activity during motion and is shown in figure 45. All of these are examples of motion analysis technologies.

In more recent times, smaller and more portable sensors have been developed and effectively employed to assess the parameters of running gait. Accelerometers, electro goniometers, gyroscopes, and in-sole pressure sensors are some examples of these types of devices. MEMS stands for micro-electromechanical systems. These tools have been put to good use in the investigation of running shoe and orthotic performance, factors that increase the likelihood of injury, running performance, effects of fatigue, and gait adjustments to different running styles.

Sensor/System	Measured Parameters
Motion analysis systems	Segment position and orientation, linear and angular velocity, and acceleration
Force platforms	Ground reaction force, loading rates, center of pressure, joint moment, and power (when used with segment position and orientation data)
Pressure sensors	Pressure distribution, vertical force, center of pressure, spatiotemporal gait parameters
Electromyography	Muscle activation and timing patterns, muscle fatigue
Accelerometers	Segment acceleration and orientation, spatiotemporal gait parameters
Electrogoniometers	Relative joint angles
Gyroscopes	Segment orientation, angular velocity, and acceleration

Figure 45 : Summary of running gait parameters measured by each system (Higginson, B. K. 2009).

MOTION ANALYSIS

Frequently, gait analysis demands the quantification of the two- or three-dimensional movements of distinct body segments. Motion capture technology [51], in which markers are placed to the person and monitored during the motion of interest, is the most prevalent approach for capturing this information (figure 46). These systems commonly employ passive markers that reflect ambient or infrared light, active markers that generate light (light-emitting diodes), or electromagnetic systems that can detect the position and orientation of a receiver placed on a body segment relative to an externally fixed transmitter.

Through manual or automatic digitization procedures, the two- or three-dimensional coordinate location of the markers can be identified. The velocity and acceleration can then be derived from this location data by taking the time derivative of the velocity and position, respectively Nevertheless, each system has its own limitations. Optical systems are susceptible to marker occlusion if there are a large number of markers or an inadequate number of cameras This technology can have significant limitations, including a requirement for expert operators, prohibitive expense, a relatively modest capture volume, and the need for a regulated operating environment.

Although treadmills are frequently employed in the research of walking and running gait to overcome challenges with tiny capture volumes, it is hypothesised that their use induces gait adaptations, such as an increase in stance phase time, that are not generally found in overground running. These modifications appear to be reliant on speed, with walking pace evoking little or no change in gait patterns, whilst changes in running gait appear to be determined more by the running style, running speed, and shoe/treadmill interaction of the particular subject. Protocols used for the examination of running gait often include a treadmill familiarization period to reduce any variations in gait between running modes; this may be especially important when evaluating treadmill-naive athletes.

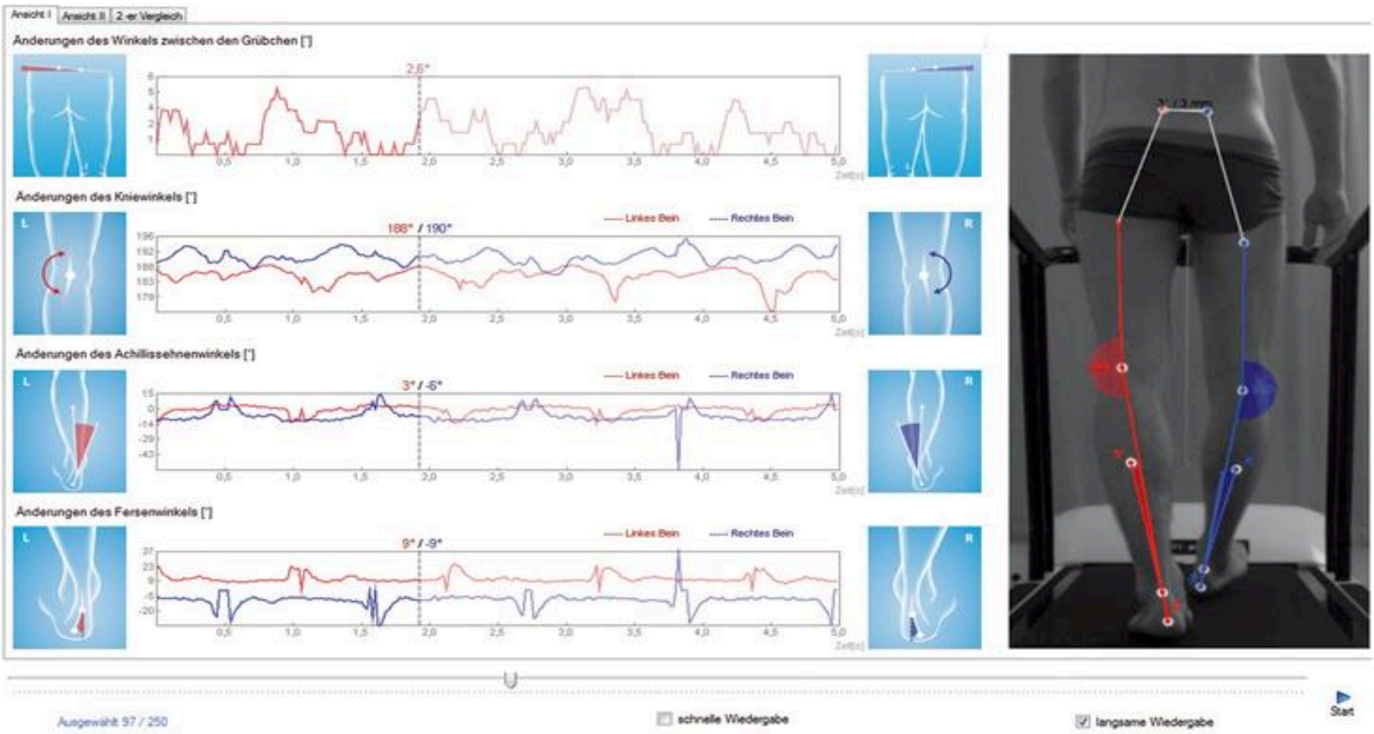


Figure 46 : Optical motion analysis system (https://diers.eu)

FORCE ANALYSIS

As depicted in figure 47, force plates are widely employed to monitor contact forces between the foot and ground during the stance phase of gait [51]. This data can be utilised to calculate impact forces, loading rates, propulsive and braking forces, and to monitor changes in the centre of pressure (CoP) over time. Due to their relatively tiny size, however, they put limits on foot placement, which may force participants to adopt a "targeting" strategy while running, so modifying their natural gait mechanics. Recent advancements in instrumented treadmills have enabled the rapid gathering of ground response forces across multiple gait cycles, allowing for highly controlled gait speed and reducing the potential inaccuracy provided by targeting tactics.

During the double support phase of walking, instrumented treadmills with multiple force plates have mostly been employed to solve difficulties surrounding the summation of forces from both feet. Due to the fact that running is defined by its single support and flight phases, all measured forces are typical of the loads under a single foot, allowing analysis to be conducted using the single force plate design. Although instrumented treadmills have been used to precisely record ground reaction forces during running, treadmills with force plates beneath the belt can be sensitive to noise due to belt friction. This noise, however, is generally confined to the direction of belt passage and has a negligible effect on vertical or medial/lateral force measurements.

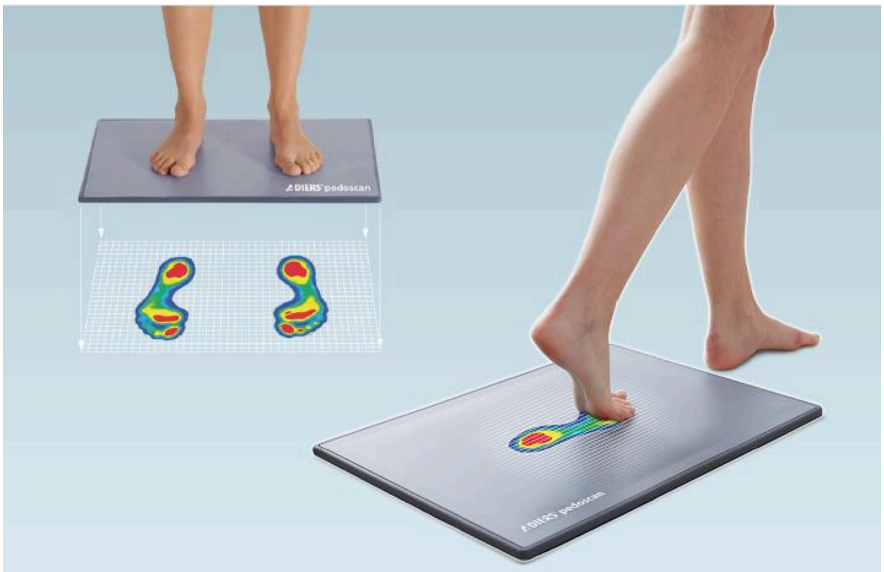


Figure 47 : Force platform (https://diers.eu)

PRESSURE ANALYSIS

Using in-shoe pressure sensors (Figure 48) is a lightweight, portable, and user-friendly alternative for analysing running gait [51]. In contrast to force platforms, they can quantify the distribution of force along the plantar surface of the foot, hence providing more information on the loading of the foot during locomotion than force measures alone. Because this device is positioned in the shoe, loads acting on the foot surface can be measured directly, whereas a normal force platform measures the force acting on the bottom of the shoe. Because of this, in-shoe pressure sensors are commonly used to quantify the effect of shoe design on foot loading and have been used to compare foot loading between similar shoe types and between shoes with different midsole designs, as well as changes in a shoe's impact-absorbing capabilities over repeated impact cycles. In-shoe pressure sensors also allow for the measurement of vertical forces encountered by the foot during prolonged running and the detection of common gait parameters required for gait analysis, such as heel strike and toe-off, which define the stance phase of gait. Although force platforms are regarded as the gold standard method for collecting these measurements, as stated above, they are limited in the number of steps that may be sampled and are normally relegated to laboratory use. In-shoe pressure sensors allow a researcher or clinician to collect data from repetitive foot strikes in an environment conducive to a natural running stride.

Figure 48 : In-shoe pressure insole (https://www.runvi.io)

ELECTROMYOGRAPHY

Electromyography (EMG) is a standard method for measuring muscle activity levels during walking or running gait [51]. Typically, the major parameters of interest are the timing of muscle activation and relative intensity, which can be acquired using surface or indwelling (fine-wire) electrodes (figure 49). This method can be used to detect aberrant gait and evaluate a runner's neuromuscular control. Normal muscle activation during the stride and stance phases of running and sprinting, as well as the influence of walking and running speeds on muscle activation and timing patterns, have been observed elsewhere. In addition to providing information regarding muscle activation levels and timing, the frequency content of the EMG signal can be evaluated to estimate relative muscle fatigue, which can be utilised to detect potential running-related injuries early on.

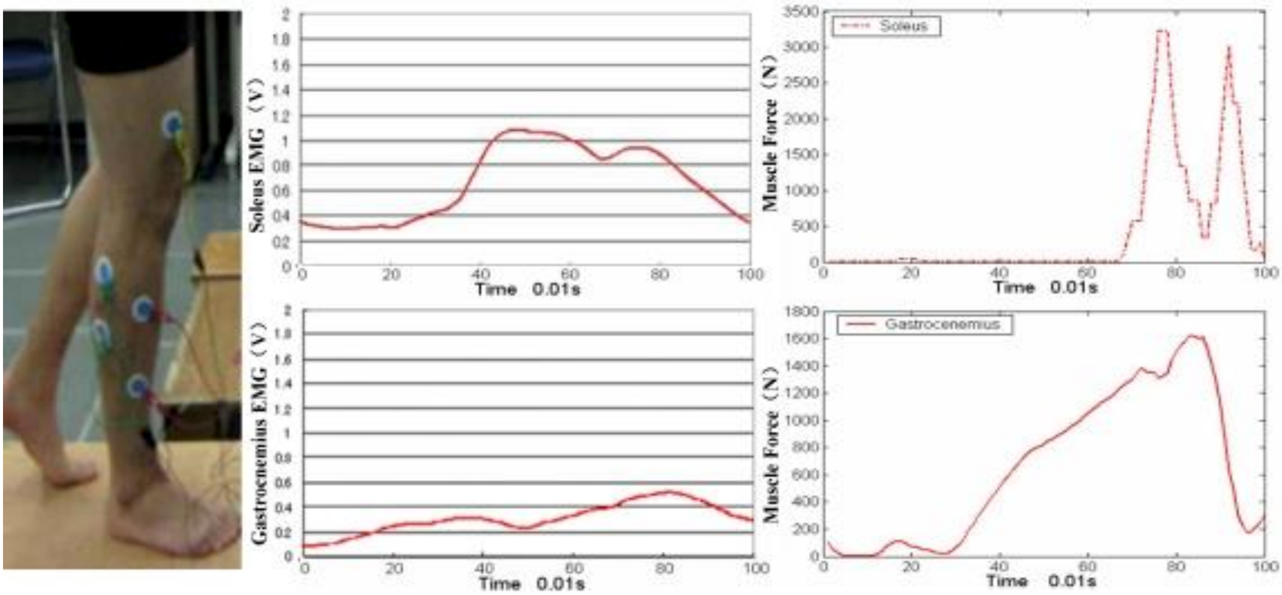


Figure 49 : Electromyogram gait analysis (Weijun Tao, 2012)

MICRO-ELECTRO-MECHANICAL SYSTEMS (MEMS)

For the measurement of human mobility, the use of body-fixed sensors such as accelerometers is increasingly becoming a feasible alternative to more conventional gait analysis methodologies. Accelerometers are inertial sensors that offer a direct measurement of acceleration along a single or many axes (figure 50), eliminating the error associated with the distinction of displacement and velocity data acquired from sources like motion capture systems [52].

Accelerometer-based systems have been successfully used to quantify the shock experienced by the lower extremity during walking and running, to evaluate the effect of footwear and insoles on tibial shock during running, to evaluate shock attenuation between body segments during running, and to investigate the effects of fatigue on running gait patterns [6]. Some accelerometers' capacity to respond to both gravitational acceleration and movement-induced acceleration enables them to be utilised for measuring segment orientation under static settings. Accelerometers, when combined with rate gyroscopes, have been discovered to produce similar joint angle, angular velocity, and angular acceleration derived from motion capture systems under dynamic conditions, and have been utilised effectively to estimate walking speed and surface inclination angles [53]. The capacity of accelerometers to be employed in the estimation of spatiotemporal gait parameters, which until recently required the use of a force plate, motion analysis systems, or footswitches, is perhaps the most enticing advantage.

As discussed in previous sections, motion capture devices and force plates are limited in their ability to measure successive strides while analysing running gait. Due to their light weight and portability, accelerometers are able to record data that can be recorded continuously throughout numerous stride cycles over an extended time period.

This method has been successfully utilised to identify modifications in running patterns following the start of exhaustion in middle-distance runners running on a track, without modifying the runner's running patterns [54]. Although mechanical testing has confirmed the validity and reliability of accelerometers for measuring accelerations within the frequency and amplitude range of human body motion, evidence suggests that they are sensitive to the attachment site and method, with skin-mounted accelerometers resulting in significantly greater peak accelerations than bone-mounted accelerometers.

Gyroscopes are small angular rate sensors that can be fitted to individual body segments to directly monitor the angular velocity of the segment [56]. Integrating the angular velocity data enables the calculation of the angular orientation. This sensing technology has been discovered to be a low-cost alternative to motion analysis systems, and algorithms have recently been developed to determine spatiotemporal gait parameters using the angular velocity measurements provided by these sensors.

Figure 50 : MEMS placement (https://runscribe.com)

2. Fatigue

Previously, fatigue was described as an exercise-induced decrease in muscle's ability to create force or power, regardless of whether the task can be sustained [57]. Enoka and Duchateau [58] observed more subsequently that a crucial aspect of this definition was the separation between muscle exhaustion and the ability to continue the work. Consequently, muscle fatigue is not the point at which a task fails or when muscles become exhausted. Rather, muscular fatigue is characterised by a steady decline in the maximal force or power that the affected muscles are able to produce shortly after the commencement of persistent physical exercise.

A standard strategy for quantifying the onset of muscular fatigue is to interrupt the fatigue-inducing exercise with brief maximal contractions (voluntary or electrically triggered) to measure the loss in maximal force capacity. The drop in maximal force or power recorded immediately after a fatiguing contraction can be used to quantify the degree of muscular fatigue produced by an intervention.

The location of weariness, which is separated into peripheral and central levels (figure 51), determines its effect on motor tasks. Peripheral fatigue was defined as the drop in force generated by a decrease in the contractility of muscle fibres, which was mostly driven by metabolic events within the muscle. By weakening the system at and distal to the neuromuscular junction, it limits the muscles' ability to produce torque. Specific processes of peripheral fatigue include a decrease in the propagation of action potentials over the neuromuscular junction and along the muscle fibres, as well as alterations in the excitation–contraction coupling mechanisms inside the muscle fibres. Central fatigue was described as an exercise-induced drop in muscular force resulting from a decrease in recruitment. This occurs proximal to the neuromuscular junction and gradually affects the CNS's ability to stimulate a muscle to its full potential.

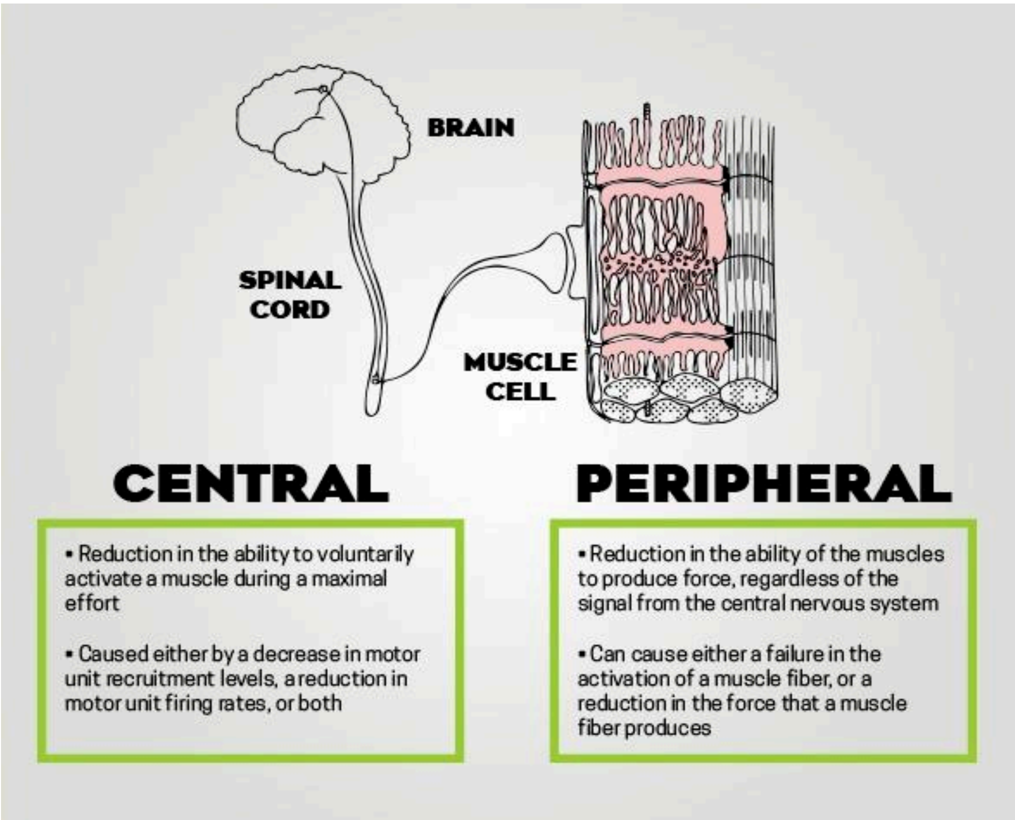


Figure 51 : Central and peripheral fatigue (S.Boyas, 2011)

This disability could be the result of physiologic or cognitive causes. Identifying central and peripheral fatigue is complicated by the involvement of both supraspinal (i.e., changes in neurotransmitter concentrations and flux) and spinal (i.e., inhibition of motoneuron excitability) mechanisms, as well as the positive and negative influences of afferent sensory feedbacks on spinal fatigue.

It has been postulated that fatigue alters biomechanical and neuromuscular function, including reaction time, movement coordination and motor control precision, muscle force production capability, and running performance. It was predicted that modifications caused by localised muscle exhaustion could result in aberrant loading and, subsequently, altered stress distribution on musculoskeletal systems. Additionally, localised muscular exhaustion may have detrimental impacts on performance.

It has been suggested that running-related fatigue modifies biomechanical and neuromuscular function, including reaction time, movement coordination and motor control precision, muscle force generation capacity, and running performance [59]. Several mechanical variables contain multiple neuromuscular and mechanical processes concurrently describing the running system: stride length and frequency, contact and flight duration, vertical peak force, centre of mass displacement, and stiffness. Examining these characteristics is of utmost importance when understanding how fatigue may alter the mechanics of running across long distances.

Friesenbichler et al. [60] explored whether fatigue influenced the damping and frequency of these soft-tissue vibrations (i.e. time into an exhaustive exercise). Fatigue may diminish this vibration damping mechanism of muscle tuning, according to their findings. These results verified the hypotheses of previous researchers: Verbitski et al. [61] observed a striking correlation between weariness and an increase in heel strike-induced shock waves. Nyland et al. [60] claimed that one of the effects of running while exhausted was a reduction in the muscles' capacity for stabilisation. Instead of the muscles, inert internal tissues such as ligaments, cartilage, and bones must absorb the loading phases while running. Mizrahi et al. discovered, while evaluating the fatigue-induced changes sustained throughout the stance phase by participants jogging 30 minutes at their anaerobic limits, that an imbalance between the muscles leads to a loss in the muscles' protective properties. They determined that fatigue in long-distance running at speeds beyond the anaerobic threshold is caused by a steady increase in the impact loading on the shin and an imbalance in the muscle contractions of the shin muscles. The combination of these two situations may impair the tibia's load-bearing equilibrium since the bone is subjected to greater bending forces.

The study of fatigue during activities such as running is complex due to the lack of knowledge regarding the precise nature of fatigue and the muscles that experience fatigue. During exhausted running, it is possible that important muscles that are active during running perform at various capacities relative to one another. Several physiological and neurological factors are associated with the inability of a muscle to generate the anticipated force production.

In addition, fatigue has a psychological component, and during a dynamic exercise, such as a graded exercise test, in which an individual stops exercising voluntarily, it is unknown if the system fails to generate muscle force at any one muscle or group of muscles, or if the individual stops exercising for a reason other than fatigue. Although challenging, the study of weariness is essential because it is a common occurrence among runners and may be linked to running injuries. For instance, Nordin and Frankel [63] provided a model indicating that overuse injuries are associated with muscle performance due to a loss of shock-absorbing capacity or a change in movement pattern. We hypothesise that shock attenuation is sensitive to fatigue, with less shock attenuated during fatigued running compared to running without fatigue.

In running, elastic mechanics are mostly utilised in two ways. Initially, elastic systems limit the work required of the muscles, so conserving energy [64]. A runner ascends and descends while running, earning and losing gravitational potential energy. Additionally, the body's centre of mass accelerates and decelerates, causing the athlete to acquire and lose external kinetic energy. Cavagna et al. [65] shown over fifty years ago that not all losses in (potential and kinetic) energy were caused by the transformation of mechanical energy into heat. A portion of the energy was initially stored as elastic strain energy and subsequently recovered during elastic recoil.

The second function of elastic structures is as suspension springs. Some lower limb structures (mostly the foot) have elastic qualities that reduce the impact force, similar to how car springs protect the driver from experiencing a strong shock while driving over a bump. During the landing portion of a stride, this second mechanism prevents rapid deceleration of the foot and potential anatomical damage. It consists of the fatty pad on the heel and flexible soles of shoes. This section does not elaborate on the second class of elastic structures, as its concentration is on energy storage/recoil processes (Figure 52).

In running, muscle and tendon are typically regarded as strain energy reserves. Alexander and Bennet-Clark [66] noted that a muscle and its tendon transmit identical forces, but that the tendon is more likely to experience bigger elastic strains. In the case of small muscle fibres and long tendon (e.g. triceps surae), the majority of strain energy is stored in the tendon.

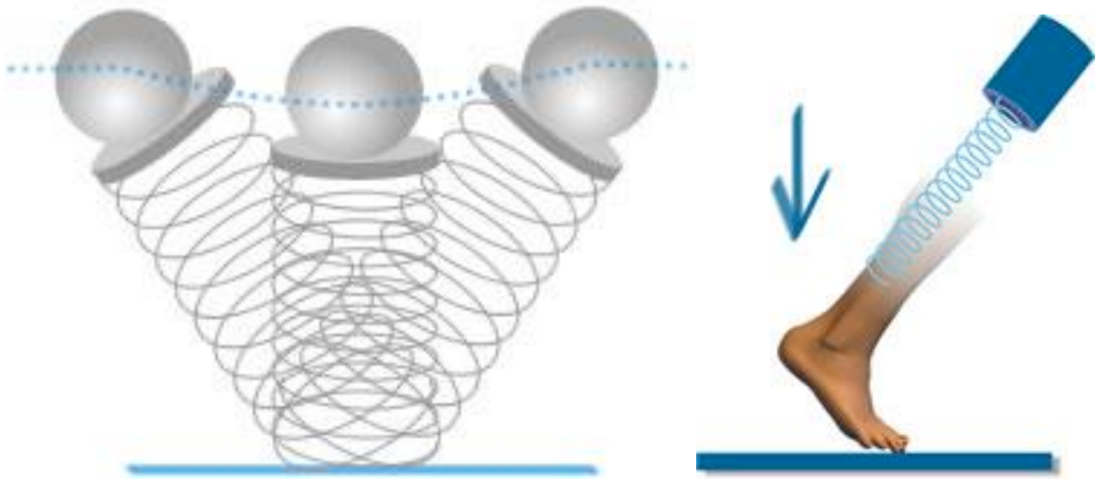


Figure 52 : Energy store/recoil running mechanism (https://volodalen.com)

At three lower limb levels, from proximal to distal, three main storing/reusing "springs" energy have been documented in the literature. The elastic characteristics of the quadriceps muscle and its tendons are represented by the most proximal of these three springs. The second spring symbolises the triceps surae and its proximal and distal (i.e. Achilles) tendons' elastic characteristics.

The structures of the medial arch of the foot are represented by the most distal spring (ligaments, plantar fascia, intrinsic muscles). According to Alexander and Novacheck [46] in a literature analysis on the biomechanics of running, the total energy turnover in each stance phase of a 70 kg man running at 4.5m.s-1 is 100 J. It is believed that 35 J of strain energy is stored in the heel cord and 17 J in the arch of the foot. The remainder of the energy must be stored in the quadriceps and its tendons. These and Farley et al observations' supported the idea that the body's system of muscle, tendon (and ligament) springs operates like a single linear spring ("leg spring").

During the stretch-shortening cycle, running is a mechanically normal human movement in which the musculotendinous tissues of the lower limb alternately store and replenish elastic energy. This takes place as the lower limb is moving forward. As a consequence of this, the lower limbs have the potential to be conceptualised as springs that are loaded by the mass and inertia of the body. This approach makes use of the "spring-mass model" (SMM) [67] to characterise the behaviour (stiffness regulation) of the musculoskeletal system of the lower limb when it is subjected to bouncing and running gaits respectively.

The leg spring stiffness is defined in this model as the ratio of the maximal force to the maximum leg compression at the middle of the stance phase. This term "leg stiffness" refers to the average overall stiffness of the integrated musculoskeletal system during the ground-contact phase. The leg spring stiffness represents the average overall stiffness of the system. In addition to this, the vertical stiffness is used to depict the vertical motion of the centre of mass (COM) while the object is in contact with another. It is measured as the ratio of the greatest force to the greatest vertical downward displacement of the COM when it is in its lowest position (i.e. at the middle of the stance phase). During the first fifty percent of the ground-contact phase, the leg spring is compressed, and then it lengthens during the second fifty percent (Figure 53).

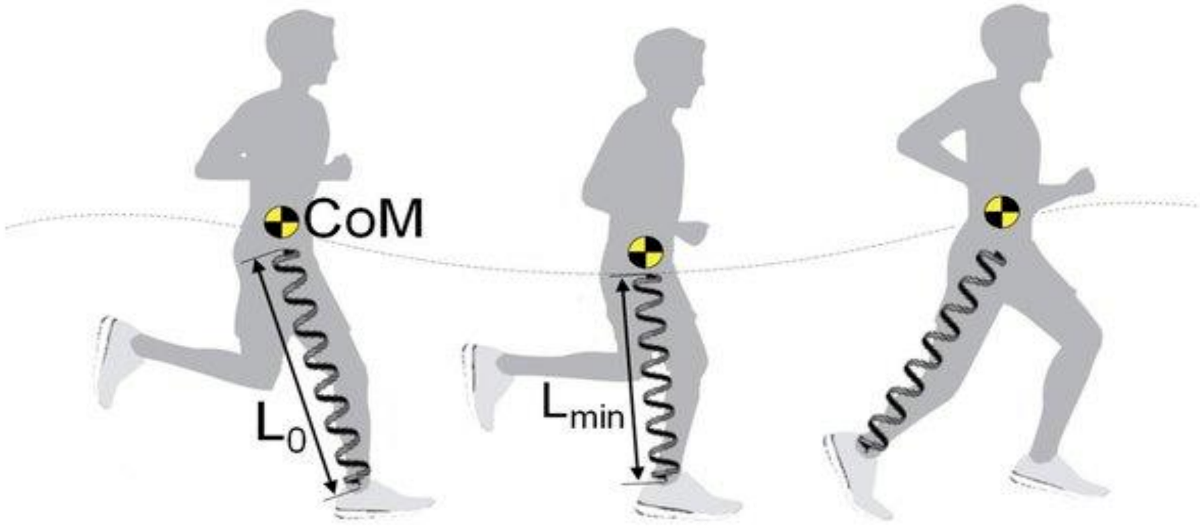


Figure 53 : Spring mass model (Juha-Pekka Kulmala, 2018)

Recently, the SMM has been utilised in a number of studies in order to explain the behaviour (stiffness regulation) of the lower limb musculoskeletal system during fatiguing runs. These studies have been conducted by a number of different researchers. Several other researchers contributed to the completion of these studies. Some of these research looked at the SMM parameters over extremely long distances and found that fatigue led to increased leg and possibly vertical stiffness as well as step frequency. Other writers, on the other hand, analysed sprint running repeats and reported, on the other hand, a constant peak vertical force and constant or decreased stride frequency and vertical stiffness in tired state. These findings contradict the findings of the previous authors. The results of experiments conducted on SMM characteristics during periods of intermediate-distance running have showed inconsistent results, which can be thought of as falling somewhere in the midst of these two extremes.

On an athletic track, Slawinski et al. [68] found no changes in the key SMM parameters despite the subject's increased level of fatigue. However, their measurements were collected at moderate speeds before and after a maximal running test that consisted of 2000 metres and lasted somewhere in the neighbourhood of seven minutes. However, due to the fact that the running phase of this experiment was self-paced, the participants had the flexibility to change both their speed and their running pattern (especially near the end of the race), which may have had an effect on the outcomes of the experiment.

Alterations in SMM have also been looked into by other authors during hard runs performed at a constant speed. Two studies discovered that runners who were exhausted while using a treadmill had reduced stride frequency and vertical stiffness, although this was only the case when they ran at a pace that was relatively slow (i.e., less than 80 percent of the velocity associated with the maximal oxygen consumption). To the best of our knowledge, Rabita et al. [69] is the only group to have investigated the SMM parameter changes at severe constant velocity (i.e. 95 percent of the velocity associated with the maximal oxygen uptake).

Their findings, which they obtained on a track, not only demonstrated a greater stride frequency, consistent vertical stiffness, and decreased leg stiffness in a fatigued state, but they also corroborated the decrease in peak vertical force that Slawinski et al. described as occurring with fatigue. Furthermore, their findings demonstrated that this decrease in peak vertical force occurs with fatigue. These more recent findings, which seem to be exclusive to high-intensity running speeds, appear to be at odds with those discovered in prior study. It is important to note that there has been no research conducted on the impact of fatigue on the SMM traits of adolescent runners; this is something that should be taken into consideration. Ratel et al. [70] conducted the only study that reported the influence that fatiguing intermittent jogging had on the stride characteristics of both adults and children. They observed a smaller decline in step frequency with rising levels of fatigue in younger subjects, but the underlying mechanisms remained a mystery to them.

3. Impact running and loading rate

On average, each foot makes contact with the ground 80-100 times each minute while running. This corresponds to a stride rate/cadence of 160 to 200 steps per minute, however cadence varies from person to person and to a certain extent with running pace. Every time you land, your foot applies a particular amount of force to the ground, which is countered by the ground applying an equal and opposite amount of force to your foot [71].

This equal and opposite force is referred to as the GRF or ground reaction force. The ground response force has a number of components, which are commonly classified as anterior-posterior (in the direction of motion), horizontal (side-to-side), and vertical (straight up and down). The vertical GRF has the largest magnitude and will be the emphasis of this section. Figure 54 illustrates the vertical ground reaction force curve for a heel-to-toe runner (modeled after Cavanagh and LaFortune, 1980) [72].

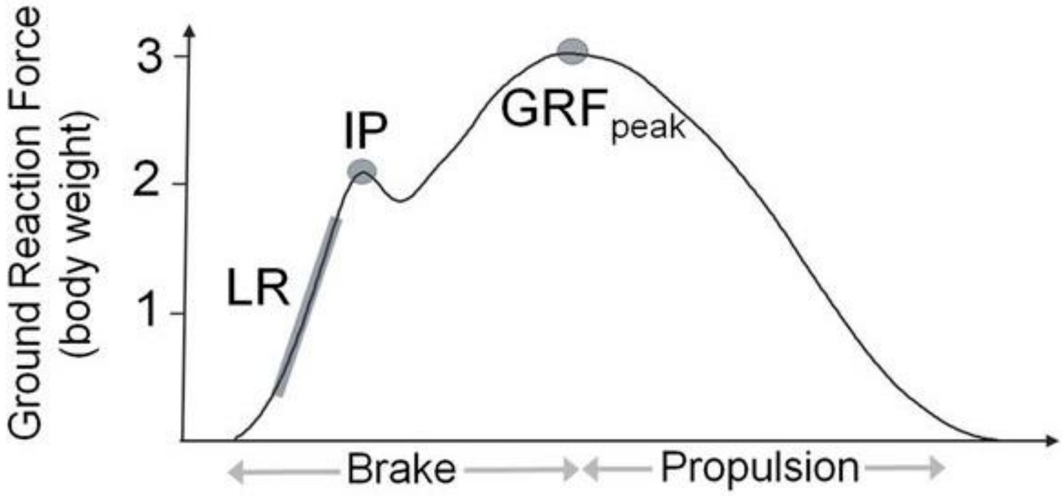


Figure 54 : Loading Rate (LR), Impact Peak and Ground Force Reaction (GRF) for a heel to toe running pattern (Cavanagh and LaFortune, 1980)

The vertical ground reaction force as a function of body weight is plotted on the vertical (Y) axis. Consider this a proxy for how much force your foot applies to the ground when you run. The horizontal (X) axis displays time in milliseconds - the amount of time each foot spends in touch with the ground varies across people and running speeds, but 300 milliseconds (about 1/3 of a second) is a reasonable value for a normal runner. The graph's curve depicts how vertical GRF changes from the point where the foot first makes contact with the ground (time 0) to the point when the foot leaves the ground on toe-off (about 300 ms).

There are two separate force maxima in this graph. To begin, the impact peak is the first force applied to the ground by the foot and lower leg upon heel contact (this graph is for a heel-toe runner). Because full body weight is not being delivered at this point, the impact peak is mostly determined by the weight of the foot and lower leg striking the ground. The active peak indicates the force applied by the foot and body weight during around mid-stance, and it is larger than the impact peak — this is the typical pattern. Vertical loading rate in the graph above is essentially the slope of the line from initial contact to impact peak (in practice, it is usually measured in the region from 20-80 percent between these points). The loading rate simply indicates how rapidly the impact force is applied — a steeper slope indicates a faster collision. A smoother slope indicates that force applied during impact is spread out across a longer time period. A good analogy would be pounding the wall with your bare fist — your fist comes to a sudden halt and the force is applied rapidly.

Forefoot strikers face a unique circumstance in which the impact peak is completely minimised by eliminating the heel strike (Figure 55). The reason for this is that, instead of absorbing impact through a collision between the heel and the ground, the arch of the foot and the Achilles tendon/calf muscle complex absorb it. Because the lower leg is significantly more supple in a forefoot strike, the collision is undetectable. A cushioned heel is unneeded if you run with a forefoot strike, as most barefoot runners do. In a roundabout approach, the cushioned heel is essentially a remedy to a problem caused by cushioned heel shoes.

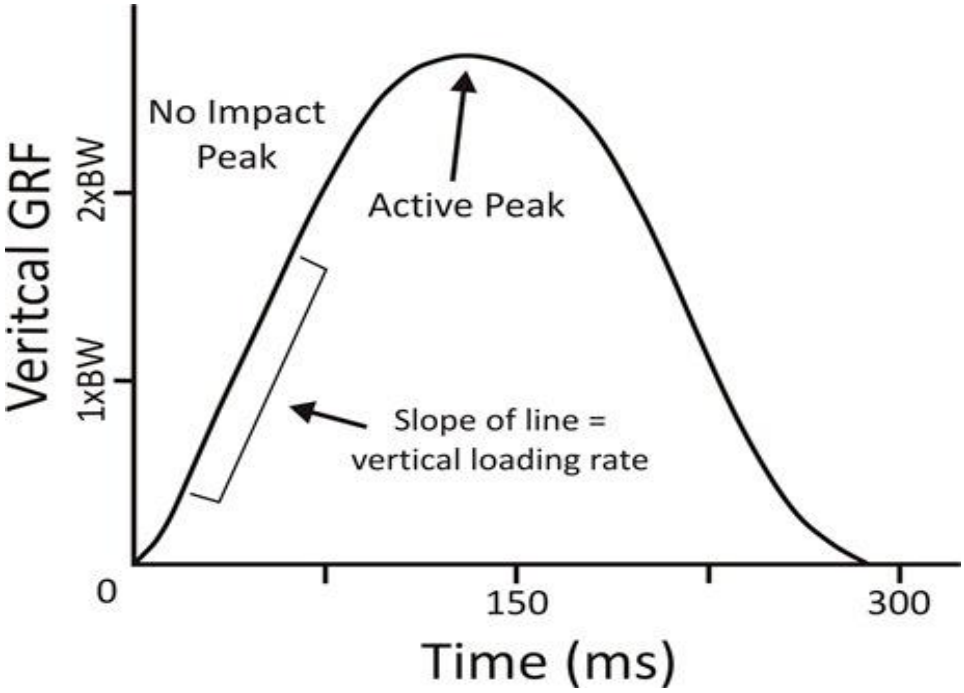


Figure 55 : Loading Rate (LR), Impact Peak and Ground Force Reaction (GRF) for a forefoot strike running pattern (Juha-Pekka Kulmala, 2018)

It is essential to note that running in shoes with a cushioned heel reduces loading rate. In Lieberman's study [4], he demonstrated that the loading rate of shod heel-to-toe runners and barefoot forefoot strikers was identical. Even if the cushioned heel reduces loading rate, it allows/causes you to run differently biomechanically than if you were barefoot (longer stride, different joint angles, pronounced heel strike, etc.). Consequently, forces are applied in a manner that the body may not have evolved to handle as effectively (e.g., joint torques are greater in shoes – Kerrigan et al. [73], 2009; a shorter, quicker stride lowers joint impacts – Heiderscheit et al. [74], 2011). Regarding the application of this understanding to injury risk, we are attempting to comprehend the repercussions of this. Higher vertical impact loading rates, but not impact peaks, have been associated with injuries such as stress fractures in the lower extremities (see this review paper by Zadpoor and Nikooyan, 2011 [75]).

Nigg [69], 1997 reported the results of a prospective study from a master's thesis indicating that impact force and loading rate are not connected with harm, and that greater loading rate was actually associated with less injuries. However, this investigation only examined short-term injuries and did not categorise them by type. In contrast to Nigg's findings, Davis et al. [76] disclose in a conference abstract the results of a prospective study indicating that vertical impact peak and vertical loading rate are associated with an increased risk of running injury.

Vertical impact variables, such as the magnitude and rate of the vertical impact peak and impact shock, have been at the focus of the running injury discussion [77] for decades. Recent research has shown that the forefoot (FF) and midfoot (MF) running footfall patterns are related with reduced injury rates than rearfoot (RF) running [78]. The lack or diminution of the vertical ground reaction force (GRF) impact peak during FF and MF running has been proposed as an explanation for these results. However, impact variables, such as vertical GRF features and impact shock, have been linked to injury in some but not all investigations. One study, for instance, discovered a reduced relative injury frequency among persons deemed to have high vertical impact force magnitudes or loading rates compared to those deemed to have low vertical impact force magnitudes or loading rates. Other vertical GRF characteristics, such as the amplitude of the active peak, may potentially be associated with the development of running injuries; however, this element has been completely overlooked in the running injury debate.

Each foot strike's impact with the ground generates a shock wave that travels throughout the body. Shock attenuation is the process of absorbing impact energy and so decreasing the impact energy between the foot and head. Understanding the elements that influence the attenuation of the impact during running is essential [59,61], as two to three times the body weight act through the foot with each foot strike and around 5000 footstrikes occur during a typical 30-minute run. The energy of the shock wave is absorbed by components including running shoes, surfaces, muscle, bone, and other structural tissues [79]. Joint motions such as ankle, knee, and hip flexion also contribute to absorb shock wave energy, underscoring the crucial role of muscles in this regard. Since shock attenuation is achieved by muscle contraction and the energy-absorbing capabilities of anatomical structures such as bone and the calcaneal fat pad, certain anatomical structures may be subjected to greater stress during impact if muscles are fatigued [80], where muscle fatigue is defined as the inability to maintain the required or expected force or power output. Nordin and Frankel [63] theorised that bone overuse injuries are associated with exhausted muscle, either as a result of a decrease in the shock-absorbing capacity of muscles or as a result of a change in movement pattern to compensate for the change in muscle ability.

Regardless of footfall rhythm, one thing remains unmistakable: running injuries are the result of complicated interactions between many elements. Further investigation of impact-related variables may reveal that the joints or tissues prone to injury may vary between footfall patterns [81]. The main source of the impact shock that is transferred through the leg and the rest of the body during running is the events surrounding the foot-ground collision. This impact shock has a close relationship with vertical GRF features and running kinematics. Everything that affects segment velocity in the instant before initial contact, such as running speed, stride frequency, and joint orientation, will determine the change in momentum of the foot and leg at initial contact, and consequently the magnitude and rate of the vertical impact peak and impact shock. The frequency composition of the impact shock is dependent upon the size and time of the vertical GRF. Given the changes in vertical GRF characteristics and kinematics between footfall patterns, the frequency content of the impact shock resulting from each footfall pattern may vary.

Because the ability of different tissues and processes to transmit and attenuate the impact shock may be frequency dependent [82], the frequency content of impact parameters may be a substantial contribution to running-related injuries. The frequency content and signal strength of the impact shock and tibial acceleration during stance are principally determined by the acceleration of the leg segments and the centre of mass of the entire body (COM). In particular, the tibial acceleration profile in RF running contains a lower frequency range (4e8 Hz) representing voluntary lower extremity motion and the vertical acceleration of the centre of mass (COM) during the stance phase and a higher frequency range (10e20 Hz) representing the rapid deceleration of the foot and leg at initial ground contact. These lower and higher frequency ranges are also typical of the vertical GRF's active peak and impact peak, respectively. In the time domain, the presence of a strong impact peak in RF running but a larger active peak magnitude in FF running suggests that the signal power contained in these lower and higher frequency ranges may differ between footfall patterns and also influence the attenuation of these frequencies.

To prevent the disruption of the vestibular and visual systems due to excessive head acceleration, the impact shock must be reduced. Attenuation is primarily caused by the absorption of energy by active muscles, changes in the joint geometry, and deformation of passive tissues. Combining active and passive mechanisms, the body adapts to larger impact magnitudes by increasing attenuation [83]. Depending on the frequency content of the impact shock, several shock attenuation techniques may be utilised. Passive mechanisms, such as the deformation of the heel fat pad, the running shoe, ligaments, bone, muscle oscillation, and articular cartilage, attenuate the higher frequency waveforms produced by first ground contact. Muscle preactivation will change in order to increase the damping of impact shock frequencies above 40 Hz. Due to muscular latency periods, muscle contractions specifically reacting to the impact stimulus and other attenuation mechanisms may only be effective at attenuating frequencies below 10 Hz. Active shock attenuation strategies consist of eccentric muscle contractions, enhanced muscle activation, segment geometry modifications, and joint stiffness adjustments. However, the body's ability to attenuate low-frequency components may be diminished. Capacity and degree of attenuation are determined by the frequency content of the impact shock and the attenuation mechanisms available. A decreased capacity for attenuation by some tissues or mechanisms may lead to a greater reliance on other tissues or mechanisms, which may result in a tissue becoming overloaded.

To our knowledge, differences in impact characteristics between RF and FF running have only been studied in the temporal domain. However, it may be necessary to investigate impact parameters in the frequency domain, since differences in the frequency content of the impact shock may alter the reliance on certain shock attenuation mechanisms in RF versus FF running, as well as the degree of attenuation. Recent research [80] indicated that RF running produced a higher percent difference in peak acceleration between the head and tibia signals in the time domain than FF running. This investigation of shock attenuation between footfall patterns utilising a transfer function in the time domain to calculate shock attenuation was a great starting step. Due to the fact that frequency content determines shock transmissibility, it is possible that critical information regarding the attenuation of specific frequency components and the methods utilised for attenuation is lost when a time domain analysis is performed.

Differences in kinematics and vertical GRF features between footfall patterns in the time domain suggest that the impact shock may contain frequency domain characteristics that are determined by these kinematic and kinetic events. In particular, the presence of the vertical GRF impact peak in RF running and a bigger vertical GRF active peak in FF running may result in changes in the signal power of the higher and lower frequency ranges of the impact shock and the degree of shock attenuation. Traditionally, acceleration-time signals from skin-mounted accelerometers have been used to estimate the amplitude of impact shock waves (figure 56). While the major component of these peaks is the impact shock wave, they also include acceleration components owing to muscular action and noise due to resonance in the flexible accelerometer connection to the body. Consequently, time-domain studies of the impact shock wave are imprecise. Although spectrum or frequency-domain analysis of the shock wave allows for more in-depth investigation and direct determination of shock transmissibility in the human body, the spectral features of the impact shock wave transmitted during sports activities remain mostly unknown [84].

As stated previously, correctly categorizing the runner's footstrike pattern is necessary for analysing shocks, impacts, and loading rate. The objective of our first experiment (appendix A1) was to examine the foot striking pattern of treadmill running utilising a force measuring platform (included directly in the instrumented treadmill) and gyroscopes for any similarities/differences in gait analysis and leg joint kinematics. We expected that despite use a variety of methods to examine running patterns, we would arrive at the same conclusion.

FIRST EXPERIMENTATION

For this first experimentation we wanted to find a methodology enabling us to discriminate RFS, MFS and FFS using MEMs; This research was carried out with the participation of twelve healthy men runners who had no history of musculoskeletal injuries. The runners had a mean age of 30.3 years (with a standard deviation of 4.9 years), a height of 178.3 cm (with a standard deviation of 5.7 cm), and a body mass of 77.7 kg (with a standard deviation of 8.5 kg). The participants were active runners who ran for fun on a regular basis. They trained a minimum of twice per week, ran a minimum of 20 kilometres per week, and had prior experience running on treadmills. The participants exhibited a diverse collection of foot strike characteristics. Based on a moderate effect size and a power measure of 80 percent, an a priori power analysis was carried out using the Hopkins technique. The results revealed that 12 participants were sufficient for the design. Twelve male runners run with MEMs attached to their tibias on the instrumented treadmill. During a 5-minute adaptation phase, participants ran at the determined speed of 3.30 metres per second. The treadmill was then stopped for 30 seconds, and individuals disembarked before remounting for data collection. When participants indicated they were ready to begin, the treadmill was restarted and the belt speed was gradually increased until it reached 2.2 m/s (BF 8 and C8) for a first recording of 30 seconds, and then again until it reached 3.3 m/s (BF 11 and C11) for a second recording of 30 seconds. Each participant completed a shod trial (C) and a barefoot trial (D) (BF). During both trials, an accelerometer attached to the right distal anteromedial tibia takes data on the three-axis acceleration measure. For signal comparison we also attached a MEMs on 5th lumbar prominence.

Figure 56: Experimental shock and impact MEMs RunScribe measurement

We employed a commercial Force Distribution Measurement Treadmill (FDM-T, Zebris®, Germany) that had a suitable conveyor belt that measured 1500 millimetres in length and 500 millimetres in width. The driving mechanism has a range of 0 to 22 kilometres per hour (with minimum increments of 0.1 kilometres per hour). The sensor unit has more than 5,000 capacitive pressure/force sensors, which allows it to measure a treading area that is 150 cm by 50 cm. The movement of the treadmill can be accounted for using a particular design for an ergometer treadmill. This makes it possible to conduct an analysis of perfectly stable gait and roll off patterns. There was just one inertial sensor utilised (a Hikob Fox®), and it had a sample rate that varied from 0 Hz to 1.3 kHz. The sensor had a tri-axial accelerometer. According to the findings of previous studies done in the lab by Thomas Provot and Xavier Chiementin [85], the distal anteromedial section of the tibia is the optimal location for implant implantation because it mitigates the effect of angular acceleration and rotational movement. Using an adhesive strapping band, the MEMS gyroscope was adhered to the surface of the skin (figure 57).

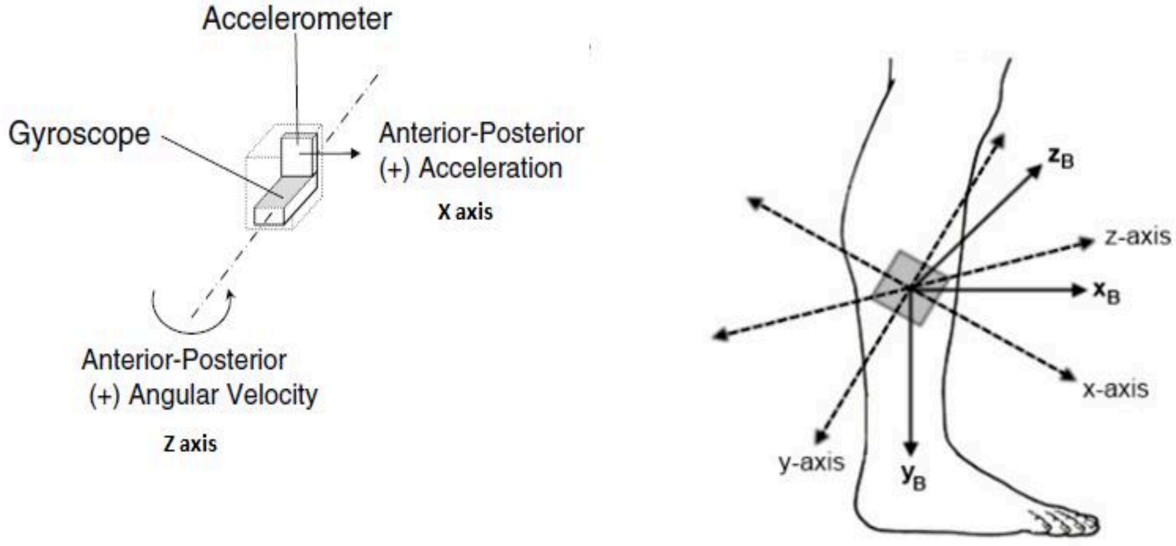


Figure 57: MEMS placement

In order to conduct this analysis of the quantitative and qualitative data side-by-side, we utilised a model based on the Support Vector Machine and applied the kernel approach. This class of algorithms is utilised most frequently for pattern analysis. As a result, we are able to investigate and evaluate many general sorts of relations that exist within the foot striking patterns. For the purpose of this study, we classified the foot strike as follows: rearfoot strike (RFS), midfoot strike (MFS), and forefoot strike (FS) (FFS).

According to Tong et al. (1999), [86] gyroscopes have the potential to be utilised in the process of calculating the inclination of a segment as well as the relative angle of a joint. A single gyroscope mounted on the shank segment can provide information on the range of segment tilt, the number of steps, the cadence, as well as an estimate of stride length and walking speed. It has not yet been applied to determine how people strike the ground with their feet. Gait analysis makes extensive use of joint angles, which can be obtained through the integration of angular acceleration or angular velocity, respectively.

As shown in figure 58, we employed a Support Vector Machine model that made use of the Kernel approach [87] for this comparative evaluation of qualitative and quantitative data. This class of algorithms is utilised most frequently for pattern analysis. As a result, we are able to investigate and evaluate many general sorts of relations that exist within the foot striking patterns. The classification of each foot strike that we employed was as follows: rearfoot strike (RFS), midfoot strike (MFS), and forefoot strike (FS) (FFS). The goal of this study was to discover whether or not the level of agreement between a single inertial sensor and an established method to quantify running gait, more especially foot strike pattern, increases with increasing velocity and whether or not the subject is shod or barefoot. Strong agreement between the two approaches is demonstrated by the fact that 80.6% of the data were in agreement, there was very little bias, and the correlations ranged from extremely high to practically perfect. The possibility exists that the 19.4 percent margin of error is connected to the positioning of the MEMS gyroscope and the method by which it was attached.

A single gyroscope mounted on the shank segment can provide information on the range of segment tilt, the number of steps, the cadence, as well as an estimate of stride length and walking speed. It has not yet been applied to determine how people strike the ground with their feet. Gait analysis makes extensive use of joint angles, which can be obtained through the integration of angular acceleration or angular velocity, respectively. Offsets and drifts of any kind have the potential to corrupt the data that is received during integration. Gyroscopes have the potential to be utilised in the process of determining the inclination of a segment as well as the relative joint angle.

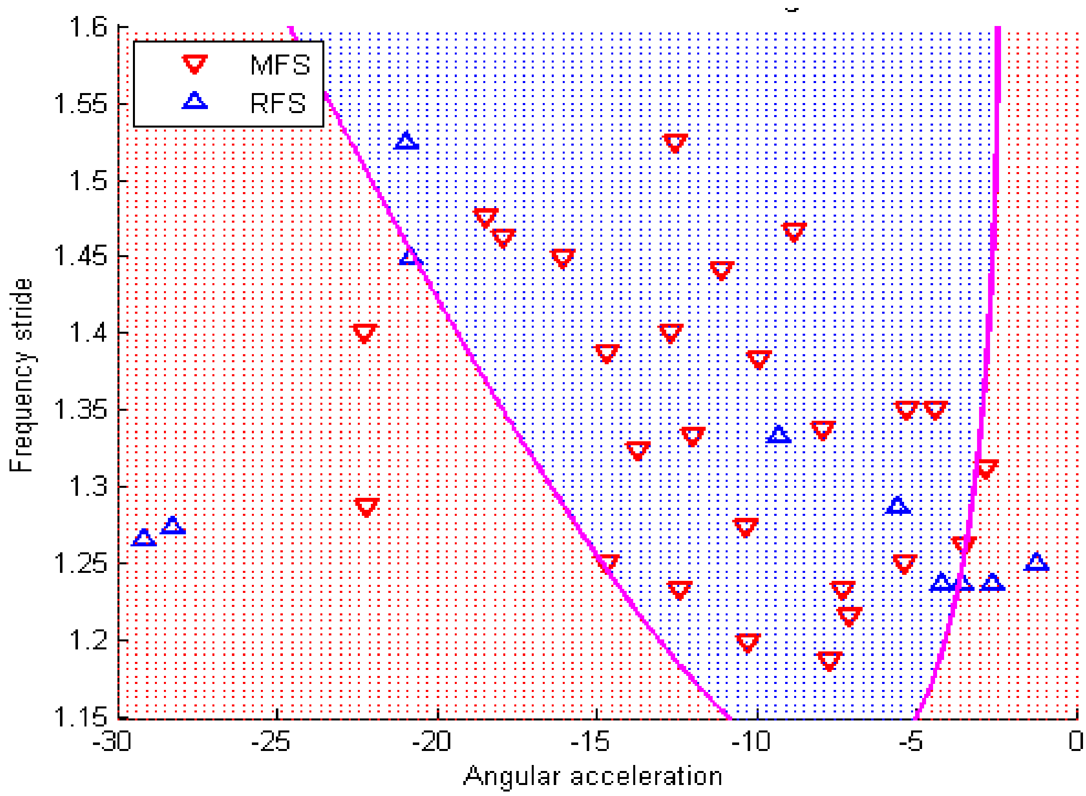


Figure 58: Classification using Kernel method and results correlation

Using gyroscopes, it is possible to discriminate different activities and in addition provide angular information. A system using a gyroscope on the shank provides rich information for gait analysis. A high pass filter to correct any drift and offset, inclination derived from the gyroscope signal is used to calculate the segment inclination range, cadence, number of steps taken and foot strike patterns. The pattern on the shank showed two minima, one occurs when foot flat and the other occurs when toe off. There were also two peaks in this pattern. The large one occurred during mid-swing and the small one occurred at heel off. This pattern provides information that can be used to identify different gait events and may be useful for developing control systems.

		Results	
		RFS	MFS
Réel	RFS	6	4
	MFS	3	23

$$Overall\ Accuracy\ de\ \frac{6+23}{36} = 80.6\%$$

The results of this first experimentation were not very conclusive probably due to the positioning of the IMU, in the spirit of collaboration and continuity we followed the protocol previously used by the lab. It would have probably been much more conclusive to position the MEMs on the foot and simply use the foot strike angle to properly determine the foot strike pattern.

4. Harmonic decomposition

Although the results of our first study were not very conclusive, we had collected a lot of good quality data on our twelve runners especially linear acceleration data thanks to the accelerometer. This is when I was introduced to Pr Boussad ABBES and Mr Serge ODOF who are specialists in mechanics, material resistance and shocks/vibrations. They had previously worked on on the Effect of Repetitive Shock [88] and Damage Estimation Method Based on Power Spectrum Densities Using Spectral Moments [89]. With their collaboration and thanks to their knowledge we decided to study the power spectrum density of the acceleration signal from our runners. This is the first step to enable to then study the effects of repetitive shocks through the shock response spectrum methodology and should be considered as an intermediary protocol. Running is a predominantly periodic endeavour [90]. Consequently, it seems natural to use harmonic components as gait characteristics. We utilise the Fourier decomposition [91] of the time series of the gait data characteristics to extract the fundamental and higher order harmonics [92]. The magnitude measured at the fundamental frequency is a measure of the total change undergone by the associated feature, whereas the relative phase between different time series indicates the temporal delay between the various features. The nonsinusoidal but nonetheless periodic trajectory of a feature is characterised by the higher harmonics assessed relative to the fundamental harmonic. A whole running cycle, or stride, consists of two steps: the left and the right.

Therefore, the fundamental period of any data series consists of either the left step or the right step of a running cycle. On the other hand, tiny discrepancies in the lengths of the two legs and their weight-bearing capacities cause the majority of humans to have a somewhat asymmetrical gait between the left and right steps. To assess if the asymmetric gait is discernible from the data series of the gait acceleration features, we extracted the fundamental period from the Fourier components of the time series. The majority of humans complete a full stride in little less than one second. To separate the harmonic decomposition feature from the average appearance feature, the DC component of the Fourier transform is set to 0 and the mean of all components is removed. In reality, the zeroth harmonic components represent the mean components of the gait's average appearance attributes. The harmonic analysis of acceleration series is performed only to shoe and leg features.

Some of the power spectra appear to contain dominating peaks, but others do not. As depicted in Figure 59, the majority of spectra have a fundamental frequency, and a few even exhibit a sizeable magnitude in the second harmonic, but only a few exhibits a discernible third harmonic. Furthermore, there are numerous reasons why we can only expect to retrieve the first and second harmonics. First, the reduced amplitude of the higher harmonics makes them more vulnerable to noise. Second, because our subjects do not have perfectly periodic running strides, the localization of the fundamental frequency contains mistakes that are magnified at the higher harmonics, hence increasing the noise in the magnitude and phase estimations at the higher harmonics.

We began comparing signals between the shank and the L5 MEMs, time signals are very different. We measure the norm of the acceleration a(t). The overall intensity of the L5 IMU below the sensor on the right tibia. The curves are “smoother” and the impacts are obviously twice as numerous on L5 as on the right tibia. The frequency analysis (PSD power spectral density measurement) also shows very different measurements in quality. The fundamental frequencies are different, the right tibia gives a fundamental frequency of 1.53 Hz and for the measurement in L5 we are on a fundamental of 3.06 Hz (twice that of the right tibia). This reflects the fact that with a positioning in L5 bilateral information is recorded, whereas with the sensor on the right tibia, there is a strong isolation of information.

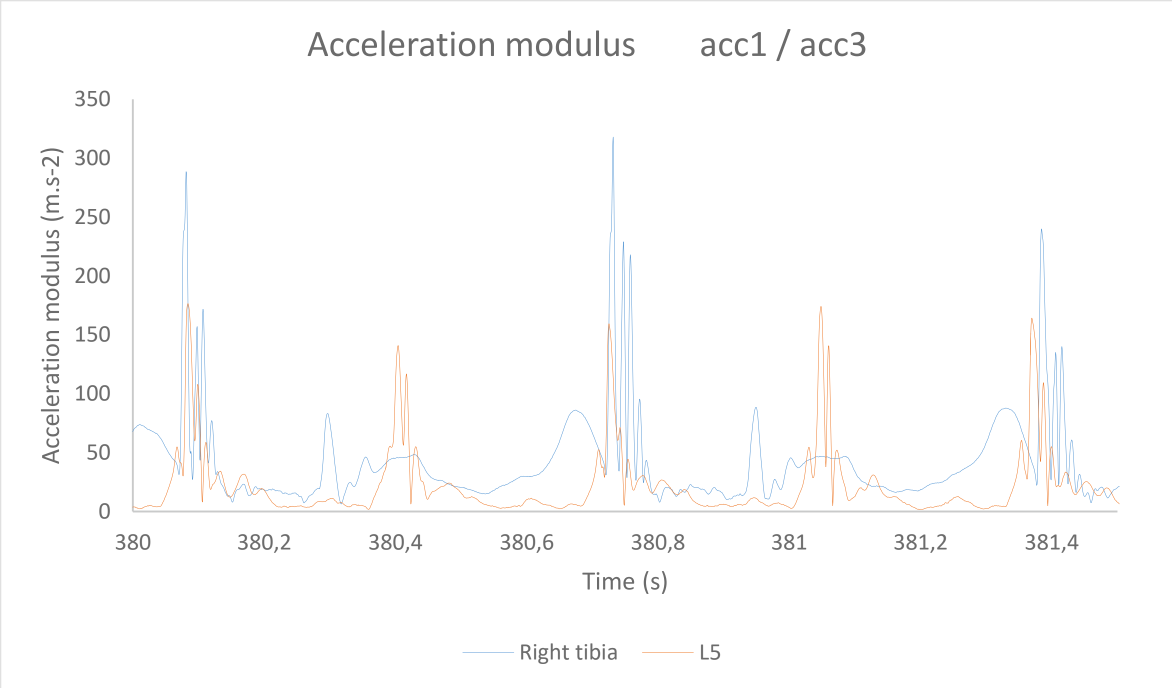


Figure 59: Experimental acceleration modulus over time comparison between tibia and L5

In order to compare the content of the signals in frequency quality and to overcome this fundamental frequency difference, each PSD is normalized on the fundamental frequency, to compare the harmonic content. We see on the graph the disappearance of the second harmonic for L5, the signal is smoothed, we lose the details. Putting the sensor in L5 provides information on both shins at the same time but does not allow fine measurement at high frequencies. The muscles and the skeleton filter the signal. For these reasons we did not use the data collected on L5 because to study the shock response spectrum we need data from shocks and impacts not filtered by the natural damping of the musculoskeletal system.

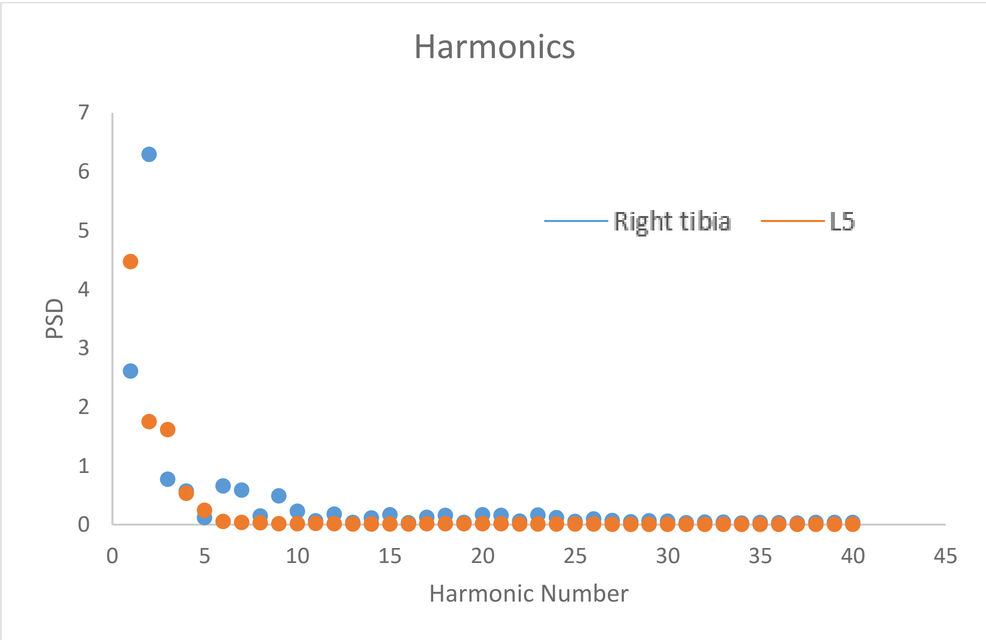


Figure 60: Experimental harmonics comparison between tibia and L5

The fundamental harmonic

Our gait fundamental spectral decomposition feature vector for a sequence is:

t = (Ω1, |Xi(Ω1)|, phase(Xi(Ω1)

Ω1 is the fundamental running frequency of a given sequence – corresponds to a single step–and i indicates the type of feature from the four ellipse descriptions of acceleration series. Intuitively, the magnitude feature components measure the amount of change in each of the acceleration series due to motion of the running body, and the phase components measure the time delay between the different acceleration series. The time series of region features is additionally noisy due to noise in the acceleration series and the fact that subjects do not have precisely periodic running strides. Consequently, the power spectra of a number of acceleration characteristics lack a strong peak indicating the fundamental operating frequency. Some have a component at a very low frequency whose magnitude is far greater than that of the actual fundamental frequency. Therefore, we employ a normalised averaging of acceleration series power spectra resulting in a considerably more dominating peak frequency, 1, which is also constant across all signals.

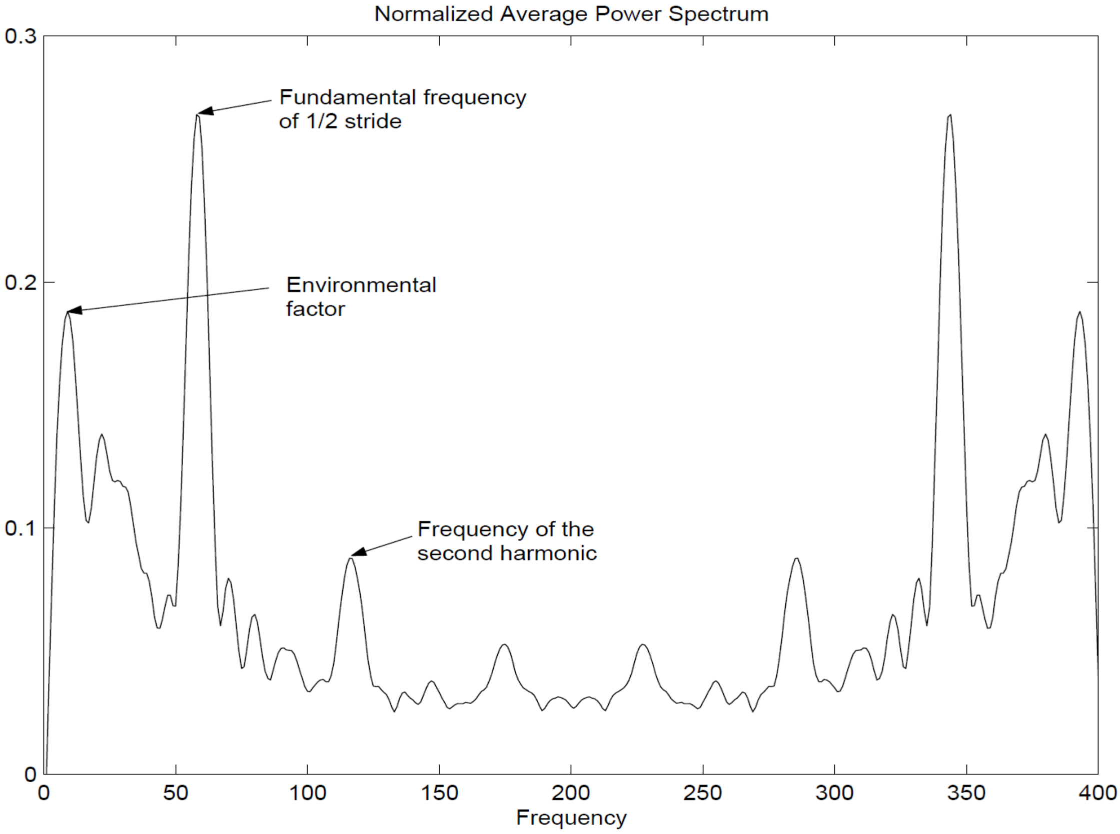


Figure 61 : Theoretical normalized average power spectrum with a global peak at the fundamental running frequency (Lily Lee, 2003)

Due to the fact that the fundamental harmonic properties consist of three different types, the distance between two gait sequences is not a straightforward Euclidean distance. These three types of fundamental harmonic properties are the fundamental period, amplitude, and phase of the fundamental frequency. Given that both the fundamental time and the magnitude are contained inside Euclidean space, it is possible to make use of Euclidean distances. In order to conduct this investigation into the similarities and differences between qualitative and quantitative data, we began by doing a frequency analysis, which allowed us to identify the fundamental frequency.

After that, we proceeded to complete the Grms computation. Grms is the common unit of measurement that is applied when attempting to specify and compare the amount of energy present in repetitive shock vibration systems. Vibration systems that produce repetitive shock (RS) produce a continually varying form of pseudorandom broad-spectrum vibration. The value of this signal's root mean square (rms) can be determined by first quadrupling the magnitude of the signal at each point, locating the average (mean) value of the squared magnitude, and then taking the square root of the average value. This calculation can be done for any number of points. The metric of grms is determined by the final number (figure 60).

	BF 8	BF 11	C8	C11
Fundamental	1.45	1.53	1.47	1.54
Grms	7.10	11.26	11.31	13.06

Figure 62: Fundamental and Grms calculation

The Grms signal is described as a time domain measurement, it is typically thought of as a frequency domain measurement taken from the Power Spectrum curve. When Grms is calculated using Power Spectrum information it is often thought of as the area under the curve of the Power Spectrum display. More accurately, it is the square root of the integral of the Power Spectrum.

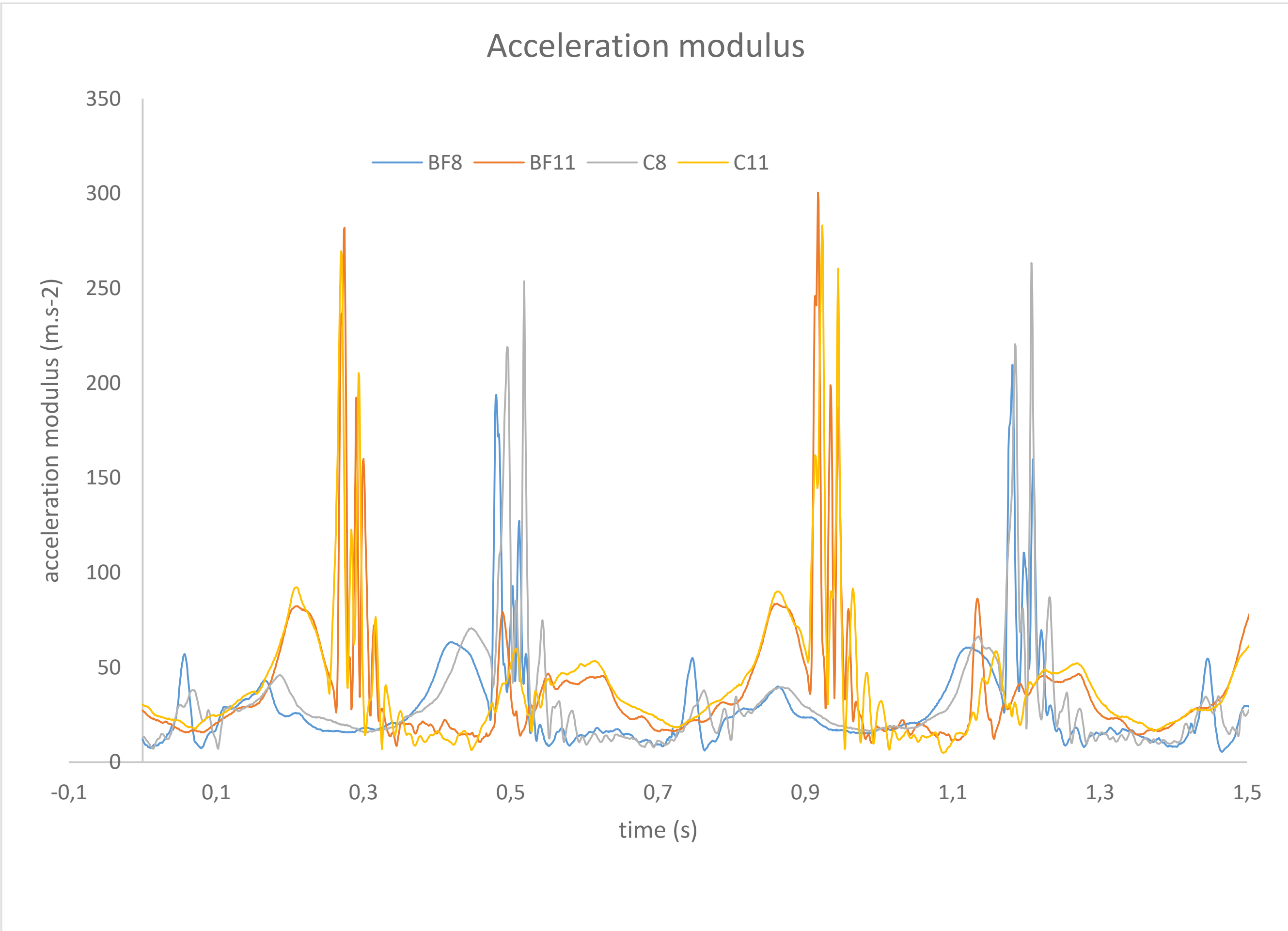


Figure 63: Experimental acceleration modulus over time for each run

The signals are different, even if the shape is globally common. We note the presence of short and intense shocks during strides. The frequency analysis will allow an easier comparison.

$$\text{PSD: } PSD(v) = \frac{1}{T}\left|\int_{-\infty}^{+\infty} a(t)\, e^{-2i\pi vt} dt\right|^2$$

n frequency in Hz and T duration of the signal

In order to compare the content of the signals in frequency quality and to overcome the fundamental differences in frequency, we normalized each PSD on the fundamental frequency to enable us to compare the harmonic's content.

The second harmonic

While the basic harmonic components capture the majority of the acceleration series' information, higher harmonics are required to catch these changes [93]. The amplitude of the fundamental frequency, the magnitude of the second harmonic, and the phase of the second harmonic with respect to the fundamental frequency intuitively offer a translation-independent description of a signal including only the first and second harmonics. Because the sampling rate and the amount of noise in the acceleration series render higher harmonic components unstable, we do not investigate beyond the second harmonic.

A visual examination of clinical gait analysis data reveals that the majority of gait parameter time series are not pure sinusoids. However, it is less obvious if the higher harmonics, specifically the second harmonic, can be easily recovered from acceleration series features that were originally generated from acceleration data. Based on the fundamental frequency computed using the algorithm given in the previous section, the second harmonic is assumed to be at double the frequency of the fundamental frequency, i.e., Ω2 = 2Ω1, even though the local peak may not actually be at that frequency. In most cases, the local peak in the range of the second harmonic occurs in the range (+1,−1) relative to our assumed second harmonic frequency. The magnitude is easily computed, while the relative phase of the second harmonic is measured relative to the phase of the fundamental harmonic as follows:

ϕ2 = phase(Xi(Ω2)) − 2 × phase(Xi(Ω1)

The distance between two sequences is computed in the same way as in the case of the fundamental harmonic, except without the fundamental period component.

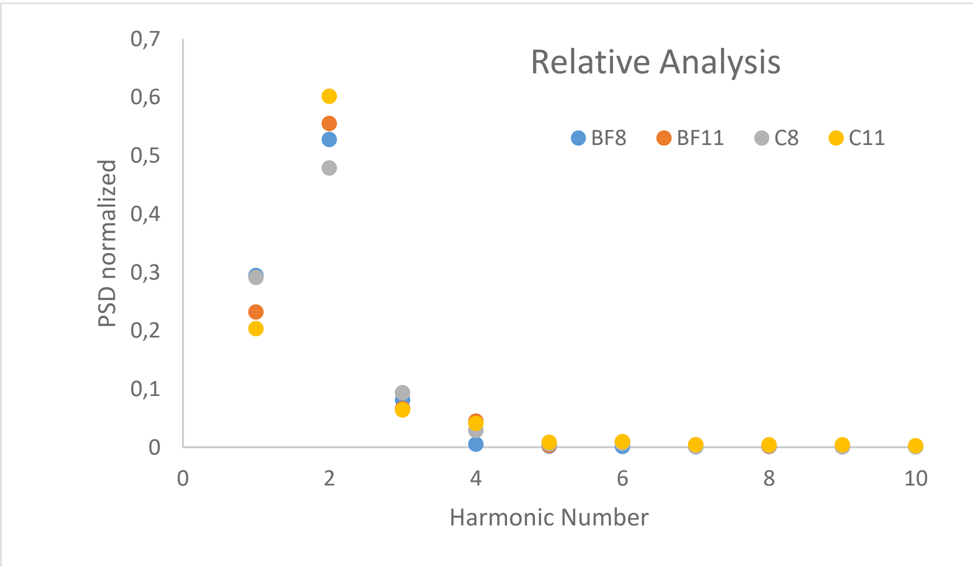


Figure 64: Harmonic analysis

The area of the peaks is a function of the harmonic number that is depicted in this graph. Either the region is relative analysed to the entire area or it is not normalised to the complete area (absolute analysis). The results of the shod running gait analysis (C8 and C11) and the results of the barefoot running gait analysis are noticeably distinct from one another (B8 and B11). The spectral examination of the second harmonic reveals a distinct rise in the frequency. Additionally, the energy at 3.3 metres per second is far more critical. The goal of this study was to determine whether or not the agreement between a single inertial sensor and an established method to measure running gait, more specifically foot strike pattern, varies with increasing velocity and whether or not the runner is wearing shoes or not, as shown in figure 61.

Because we believed that we could readily discern them in their harmonic content, using the difference in impact shock wave between barefoot and shod running was a straightforward choice for us to make. Additionally, utilising speeds that differ from one another would enable us to see a distinct variation in the power spectrum.

5. Shock Response Spectrum

A graphical representation of a shock, or any other brief acceleration input, in terms of how a Single Degree Of Freedom (SDOF) system (like a mass on a spring) would behave to that input is referred to as a Shock Response Spectrum (SRS) [94]. The natural frequency of a hypothetical SDOF is plotted along the horizontal axis, and the peak acceleration that this SDOF would experience as a direct result of the shock input is plotted along the vertical axis of the graph (figure 62).

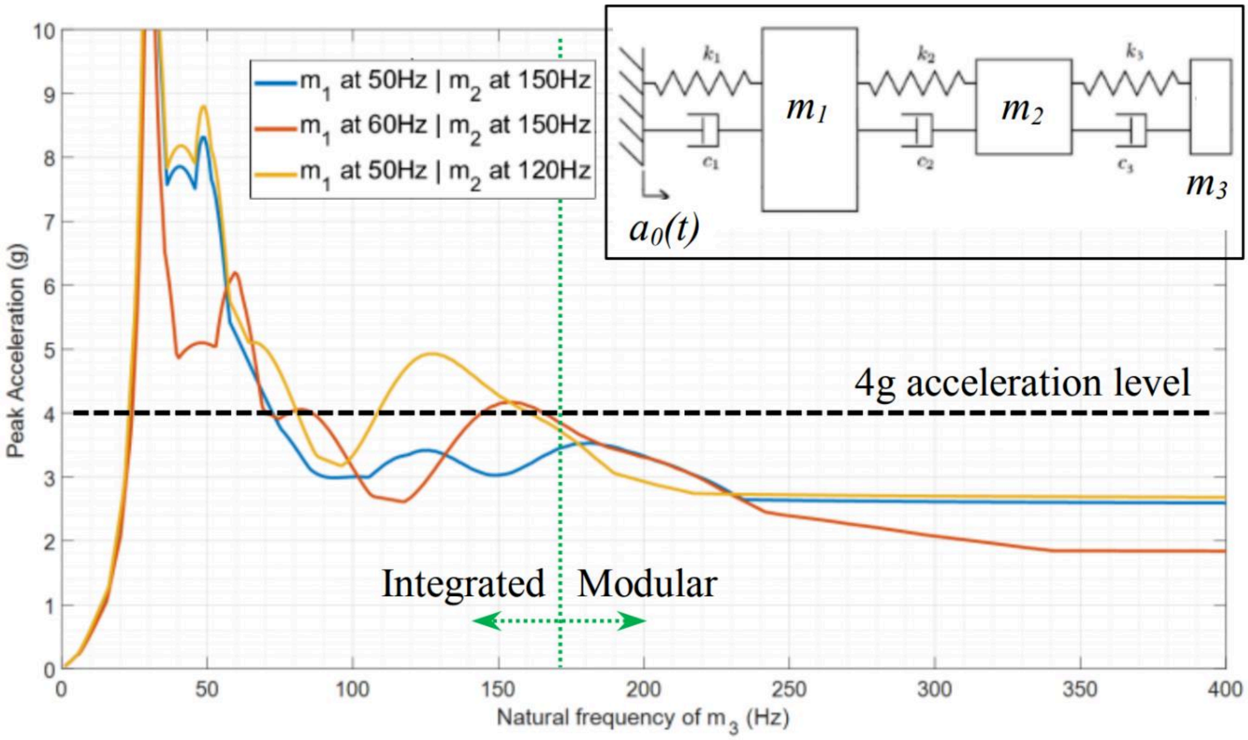


Figure 65 : Theoretical SRS pattern (Styan Zwerus, 2021)

Throughout our research, we propose a new methodology for the analysis of shock events occurring during the proposed experimental procedure. Our approach is based on the Shock Response Spectrum (SRS), which is a frequency-based function that is used to indicate the magnitude of vibration due to a shock or a transient event. The main aim is to analyze the ability of the human musculoskeletal system to attenuate the mechanical stresses resulting from the fatigue effect by Shock Responses Spectrum (SRS) of the foot strike–generated shock waves during running.

Most of previous studies focused on shocks/impacts, ground force reaction or spectral or vertical impact load rate. Using SRS as a measurement in running gait analysis has never been studied as off today. This innovative approach could pave the way to a whole new way of assessing a runner's gait pattern using smart connected shoes.

The Shock Reaction Spectrum (SRS) describes the maximal dynamic response of a specific component of a system as a result of a given loading condition. Traditionally, the system has been modelled as an array of damped linear spring masses, with the element under consideration having a single degree of freedom (SDOF). This element's SRS is then computed and analysed.

This is exemplified in Fig. 62, which shows the response at the component level (m3) in response to a dynamic loading condition applied to the system's base, a0(t), for a given upstream system configuration (m1 and m2). It is critical to understand that the SRS curve is not a continuous function, but rather a series of connected points. The whole dynamic response is computed computationally for each natural (eigen)frequency of m3, and each corresponding peak value is depicted individually in the curve. Peak responses are indicated in acceleration (number of g-forces) in this case; however, peak displacement values may also be supplied.

To compute the SRS of Fig. 62, the natural frequency of the system and subsystem levels, including damping ratios, must be determined in advance. Three design scenarios for m1 and m2 are given, with a natural frequency altered by 20% in each case. The graph shows how slight changes to m1 and m2 can have a significant impact on the response of m3 across the whole spectrum. Furthermore, the final propagation level at higher frequencies varies depending on the scenario. Traditionally, one SRS (i.e. one line of Fig. 62) would be stated as a design requirement for a downstream component, leaving the design team with a disconnected system and thus little design freedom. In this regard, Fig. 1 already provides a more systematic approach by demonstrating the system-level impact and forecasting component sensitivity to upstream design decisions.

The SRS of Fig. 62 is especially relevant during component level design in forecasting peak responses that (electronic) components at this position will be exposed to. For example, when assuming a limiting 4g acceleration at the component level, the designer can determine possible ranges for the natural frequency (i.e. areas where the SRS curve is below the dashed, black line) and adjust the mechanical design accordingly, for example, by considering rigid vs. (semi)suspended interfaces between subsystems. When selecting (i.e. designing for) a natural frequency of m3 above 170 Hz, the upstream design becomes unimportant, according to Fig. 1. As seen in the graph, the design of m3 may be regarded modular under these boundary constraints.

In a more sophisticated instance, where, for example, excessive rigidity must be avoided or is simply impossible, less freedom is accessible, and the design team must preserve a system's perspective (i.e. follow an integrated design approach). This later example demonstrates how (small) design changes upstream can have a significant impact at the component level. However, the impact of these adjustments is not clear, even for skilled designers, and relies on numerical analysis. Furthermore, this technique is uncommon in current design processes and is not discussed in literature.

Shock response spectra are used in various industries [95] where dynamic loading conditions are crucial. Sheet metal blanking, high speed machining, bolted and bearing joints are examples of manufacturing processes. An SRS is used in spindle rotor design to measure radial displacement and vibrations. Similarly, an SRS is used to analyse spacecraft and launch vehicle components that are subjected to dynamic loading conditions from a number of sources. An SRS is used to model vibration-induced failures in order to better understand how automotive electronics behave to actual driving conditions. An SRS analyses the ability of the human musculoskeletal system to attenuate mechanical stressors caused by the fatigue effect in biomechanics.

<u>Shock Response Spectrum calculation in running biomechanics</u>

Running Running is the preferred form of exercise for millions of individuals across the globe and of all ages. Its simplicity is a major factor in its widespread acceptance. However, running increases the risk of musculoskeletal injuries, and it is necessary to understand the cause of these injuries in order to prevent them effectively [100]. One of the essential roles of the human musculoskeletal system is to attenuate and disperse shock waves created by foot contact with the ground [96].

These shock waves are generated by the vast majority of motions, including walking and running. Walking and running are distinguished when periods of double support during the stance phase of the gait cycle (both feet are simultaneously in contact with the ground) give way to two periods of double float at the beginning and end of the swing phase of gait (neither foot is touching the ground) [48]. As speed rises, the initial point of contact typically shifts from the hindfoot to the forefoot.

Running requires frequent collisions between the foot and the ground with a single leg. Such impacts are distinguished by a transient peak in the ground reaction force (impact force), fast deceleration of the lower extremity (impact shock), and the propagation of an acceleration and deceleration wave (impact shock wave) through the body [80].

Several body structures and systems, such as bone, synovial fluids, cartilage, soft tissues, joint kinematics, and muscular activity, must attenuate the impact shock wave experienced by the body during landings. Soft tissues and bone provide passive attenuation of shock. Active shock attenuation is accomplished via eccentric muscular activity [82].

This active mechanism is believed to be far more important than the passive mechanism in shock attenuation. Since muscles are believed to play a significant role in energy and shock absorption during landing, it has been suggested that diminished muscular activity due to exhaustion reduces the body's shock absorption capacity and thus increases the risk of injury [97]. Fatigue has been characterised as any decline in the overall neuromuscular system's capacity to generate force, regardless of the force necessary in a given situation [61].

Repetitive impact loads have been associated with degenerative joint disorders and athletic overuse injuries, such as stress fractures, shin splints, osteoarthritis, and lower back pain. Although the particular mechanisms of impact-related damage are relatively understood and contentious, there is substantial evidence relating impact, fatigue, and injuries [7-59-71-98].

In these experiments, we suggest a new methodology for the study of shock events that occur during the experimental process that has been proposed (appendix A3 and A4). Our method is predicated on the Shock Response Spectrum (appendix A5) (SRS) [99], which is a frequency-based function that is utilised to represent the magnitude of vibration that is caused as a result of a shock or a transient event [99, 100]. The primary objective is to investigate how well the human musculoskeletal system is able to dampen the mechanical stresses brought on by the fatigue effect, as measured by the Shock Responses Spectrum (SRS) of the shock waves created by the foot strike when the subject is running. The majority of earlier investigations concentrated on shocks or impacts, ground force reaction, spectral or vertical impact load rate, or any combination of the three. To this day [104], there has never been any research done on the gait analysis of running that uses SRS as a measurement. This novel methodology might pave the way for an entirely new method of analysing a runner's gait pattern that makes use of smart shoes that are connected to the internet.

These investigations were conducted with the goals of determining the influence that fatigue has on the attenuation of impact shock waves and analysing how human biomechanics relate to the shock attenuation that occurs during running. It was expected that exhaustion would produce a decrease in the shock attenuation capacity of the musculoskeletal system, which would consequently involve a potentially increased risk of overuse injury.

SECOND EXPERIMENTATION

In our second experimentation fatigue was induced to validate our hypothesis that exhaustion would produce a decrease in the shock attenuation capacity of the musculoskeletal system. This study included five high-level CrossFit athletes who volunteered to take part in it. There were four men and one woman among them, and all of them ran at least three times per week. None of them had any musculoskeletal injuries. The athletes had a mean age of 26.4 years (with a standard deviation of 3.9 years), a height of 182.3 cm (with a standard deviation of 5.7 cm), and a body mass of 81.7 kg (with a standard deviation of 8.5 kg). Two times each week, the participants ran distances ranging from 10 to 15 kilometres and participated in interval training consisting of sprints of varying lengths and intensities. During this particular experiment, a total of two Micromachined microelectromechanical systems (MEMS) accelerometers (RunScribe®) were utilised. On top of the foot, in the shoelaces, were mounted the two RunScribe pods (figure 63).

Figure 66: MEMS accelerometers placement.

The RunScribe pods include 9-Axis Motion Tracking, which integrates a 3-axis gyroscope, 3-axis accelerometer, and 3-axis compass into a single device, in addition to an integrated Digital Motion Processor. This allows the pods to track motion in nine different axes simultaneously. This gives us the ability to assess other things at a sample rate of 500 Hz, including efficiency (measured as stride rate, contact time, and flight ratio), motion (measured as feettrike type, pronation, and pronation velocity), shock (measured as impact Gs and braking Gs), symmetry, and power. After a brief period of warming up, the participants were given the task of running 800 metres at their absolute fastest possible pace. They immediately hopped on an Assault AirBike (Rogue, Columbus, OH, USA) (figure 64) right after the first run, and for the next two minutes they were challenged to perform at their absolute peak effort (the power had to stay above 400 Watts for the 2 min). After getting off the Assault AirBike, participants were given the instruction to complete a second 800 metre run at their highest possible intensity level. The identical routine was then performed three times, beginning with a two-minute warm-up on the Assault AirBike, followed by a run at the highest possible intensity. The three 800 metre run intervals were the only ones that were being recorded by the RunScribe pods, as they were turned off for the Assault AirBike sessions.

Figure 67: Assault AirBike (https://airbikeeurope.com)

Because they require a one-of-a-kind and exceptionally taxing level of exertion, Assault AirBikes, sometimes known as "the Devil's trike," were employed in experiments to create exhaustion in test subjects. For HIIT (High Intensity Interval Training) and metabolic conditioning, it is regarded as the most feared but also the most effective instrument. The CrossFit community has given it this reputation.

In the study of the structure of composite wave forms, such as impact shock waves, spectral analysis is a technique that is frequently utilised. The Fast Fourier Transformation (FFT) is the principal instrument used in spectral analysis. This allows us to calculate the runner's native frequency, which corresponds to the peak of the Power Spectral Density (Figure 65):

$$PSD = \frac{1}{N}\left|\int_{-\infty}^{+\infty} a(t)e^{-j2\pi f t}\, dt\right|^2 \tag{1}$$

where N is the number of points of the recording, $a(t)$ is the acceleration modulus, f is the frequency and t is the time. Power Spectral Density (PSD) provides a convenient method of separating different frequency components in the impact shock wave such as acceleration moments due to impact shock.

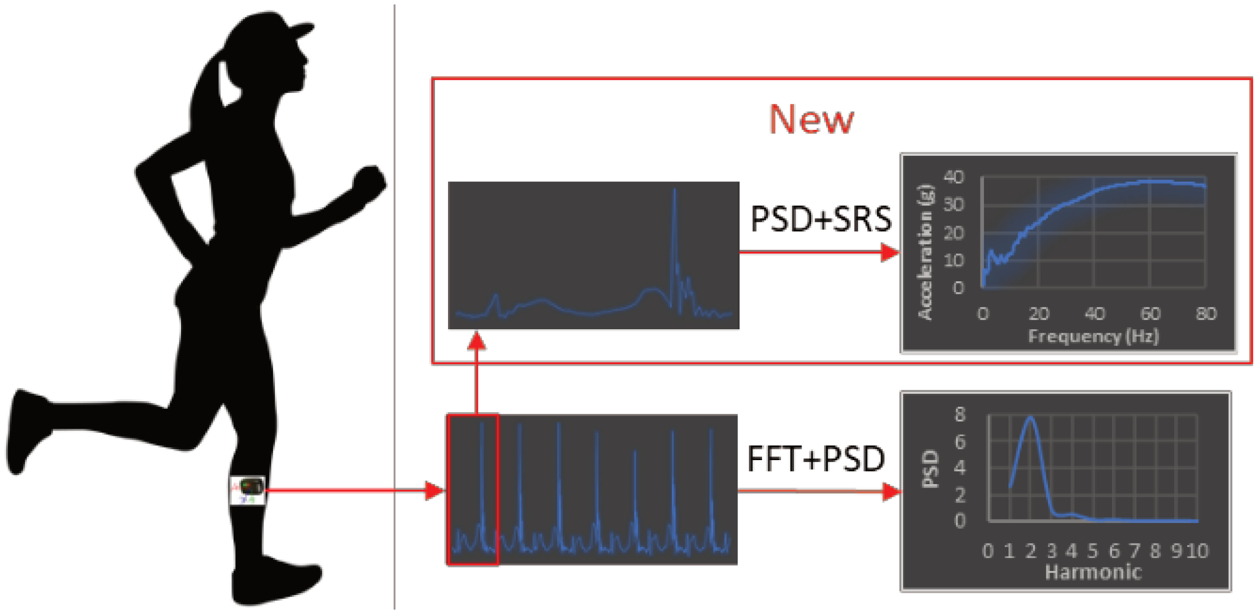


Figure 68 : SRS vs. PSD running analysis.

In this thesis, we present a new methodology for the study of shock events that occur during the suggested experimental process. This methodology was developed by ourselves. Shock Response Spectrum (SRS) is a frequency-based function that is used to represent the degree of vibration that is caused by a shock or a transient event [94]. Our method is predicated on the SRS, which is a frequency-based function. In this particular investigation, the following method, which is comprised of numerous stages, is utilised:

Step 1: The acceleration modulus $a(t)$ is extracted from the recording. Figure 66 illustrates the acceleration modulus results for one athlete.

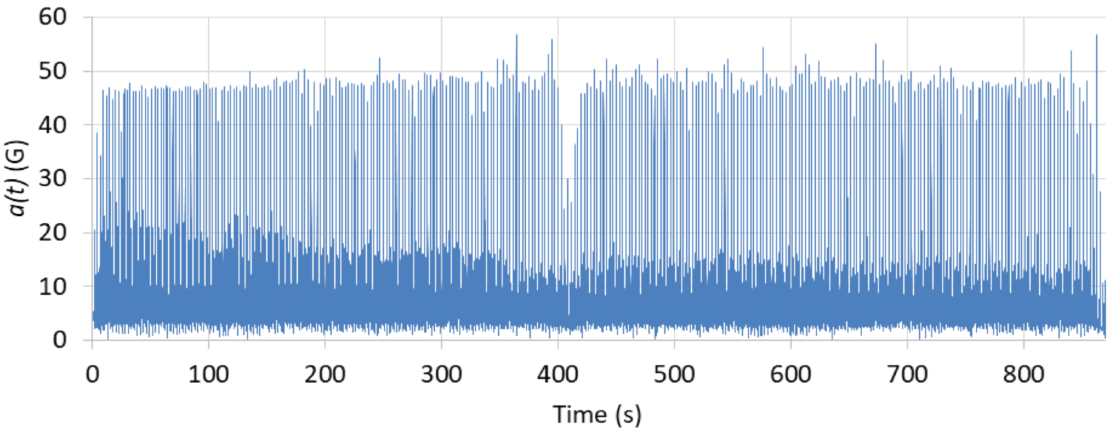


Figure 69 : Experimental acceleration modulus $\boldsymbol{a}(\boldsymbol{t})$ for one athlete.

Step 2: The power spectral density (PSD) given in Equation (1) is then calculated using a Fast Fourier Transform (FFT). This calculation allows us to determine the fundamental frequency of the runner f_0 corresponding to the position of the largest peak of the PSD [93]. The inverse of this frequency gives the time period of the runner's step as: $T = 1/f_0$. The proposed algorithm extracts automatically the "first" step from the entire signal, and thus defines the "pattern" of the runner as shown in Figure 67.

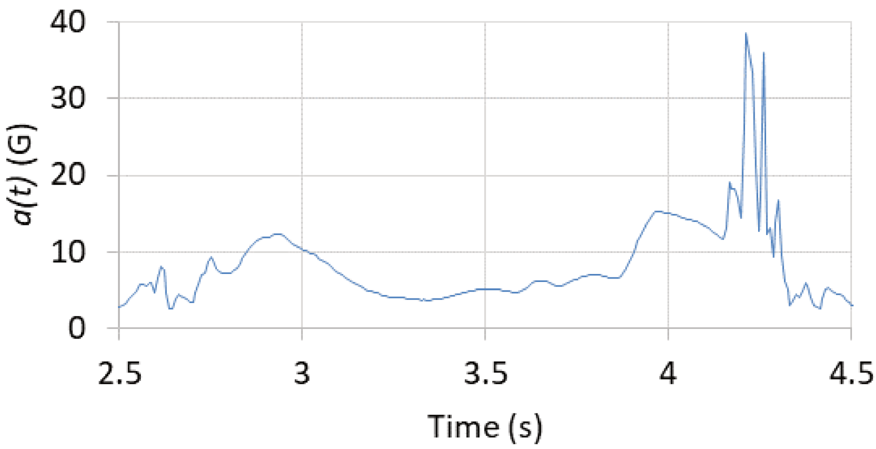


Figure 70 : Experimental pattern of one athlete.

Step 3: We then carry out the cross-correlation $CC(\tau)$ between the runner's pattern and the recording's duration $a(t)$:

$$CC(\tau) = \int_{-\infty}^{+\infty} a(t) pattern(t+\tau) dt \tag{2}$$

We observe that at each step, the convolution is maximum. For each maximum value of $CC(\tau)$ we calculate the SRS of each step and of the entire signal as explained in the next step.

Step 4: The calculation of the SRS is based on the acceleration time history. It applies an acceleration time history as a common base excitation $(\ddot{y})$ to an array of single-degree-of-freedom (SDOF) systems composed of spring (k_i), mass (m_i) and damper (d_i), as depicted in Figure 68 and 69.

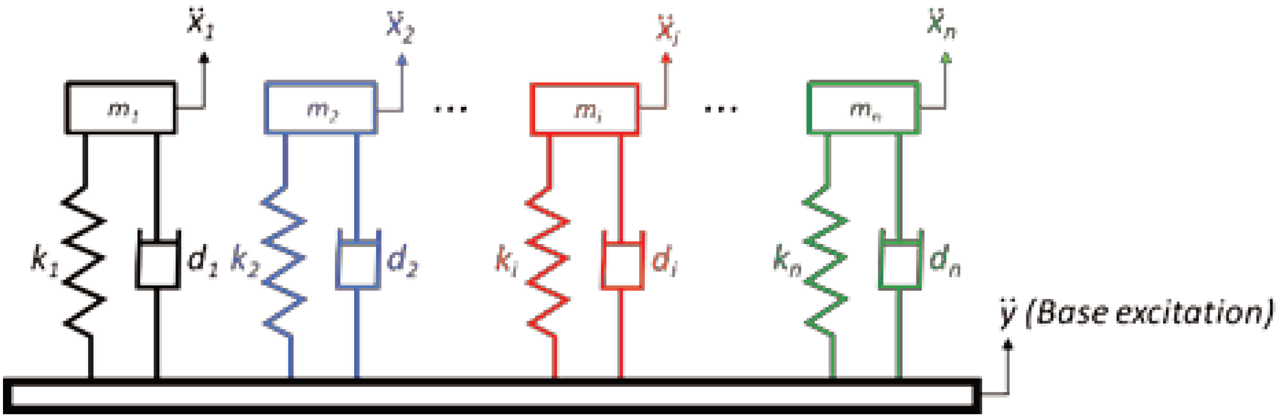


Figure 71 : SRS model.

$\ddot{x}_i$ is the absolue response of each system to the input $\ddot{y}$. This can be determined by applying the Newton's law to a free-body diagram of an individual system, as shown in Figure 69.

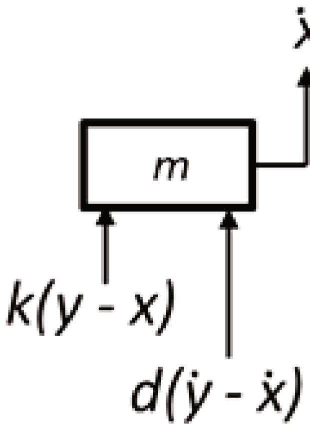


Figure 72 : Free-body diagram of an individual system.

The force balance yields the following governing differential equation of motion:

$$m\ddot{x} + d\dot{x} + kx = d\dot{y} + ky \tag{3}$$

By defining the relative displacement $z = x - y$, Equation (3) can be rewritten as:

$$\ddot{z} + 2\xi\omega\dot{z} + \omega^2 z = -\ddot{y} \tag{4}$$

where $\omega_0 = {}^{k}/_{m}$ is the natural frequency in radians per second and $\xi = {}^{d}/_{(2\omega_0 m)}$ is the damping ratio. Moreover, ξ is usually represented by the amplification factor $Q = {}^{1}/_{(2\xi)}$.

Since the base excitation $\ddot{y}$ is an arbitrary function of time, Equation (4) does not have a closed-form solution.

We have utilised the algorithm for calculation of the SRS that was proposed in order to do the calculation of the SRS for each step as well as for the complete signal. When a system's natural frequency f 0 and the quality factor Q for each potential natural frequency are known, we are able to use SRS to calculate the maximum acceleration that the system will be subjected to.

In this investigation, a relative dampening of 5% was utilised, which produced a value of Q=10. Calculations of SRS can be performed throughout the duration of a full recording as well.

After that, we look at the peaks that occur at the fundamental frequency as well as the harmonic frequencies of the recorded signal. Within the confines of this discussion, SRS combines the ideas of transfer function and reaction to transitory regimes.

Because it takes into account the frequency as well, intra comparison of the SRS allows for a great deal of nuance to be brought to the study. The aggressiveness of a running stride is determined not only by the value of the greatest acceleration, but also by the general shape of the movement; the SRS is the sole analysis method that allows this to be taken into consideration. The process of determining SRS is illustrated in more detail in Figure 70.

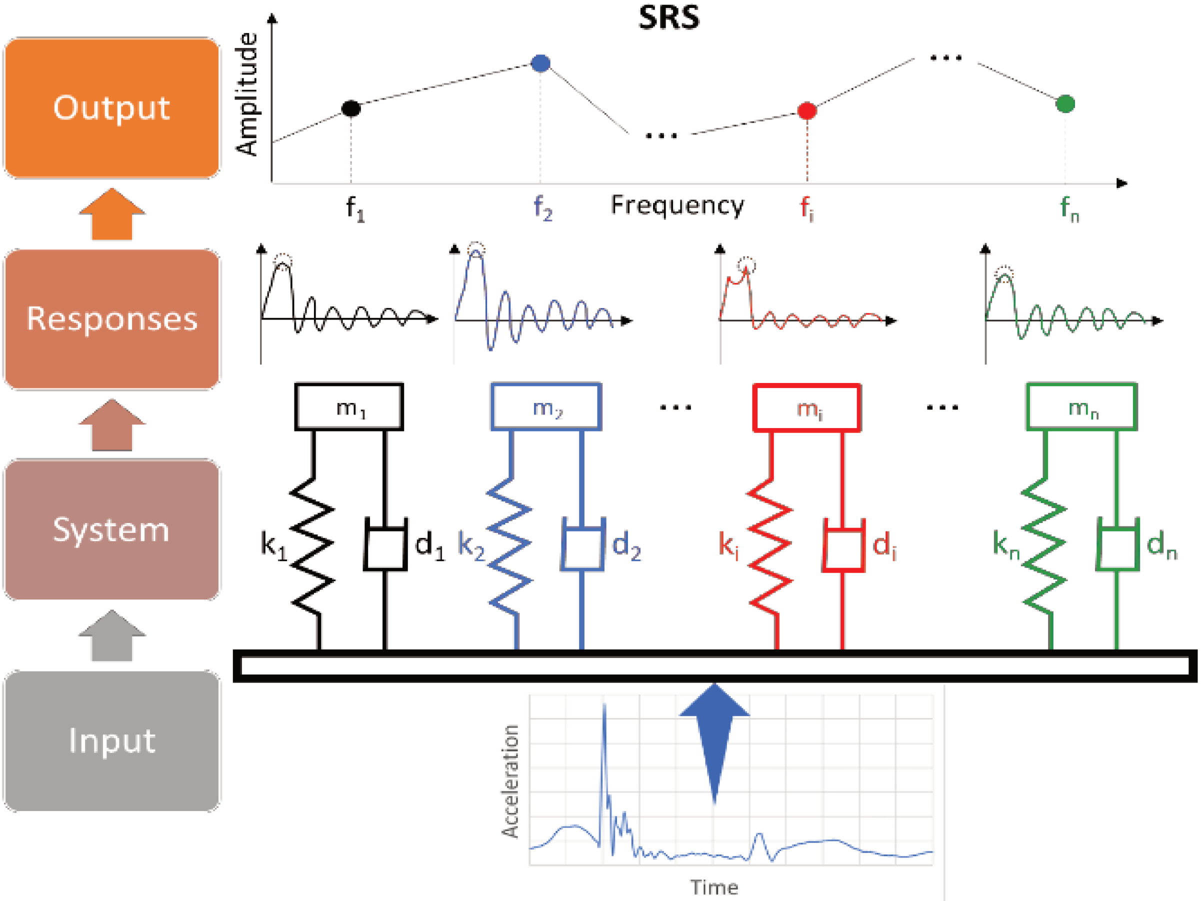


Figure 73 : Numerical workflow for SRS determination (Nicholas Di Vincenzo, 2019)

The capacity to isolate spectral peaks from the rest of the data is the primary benefit of spectral analysis over time-domain analysis of the impact shock wave. This advantage gives spectral analysis an advantage over time-domain analysis. Because the motion, impact, and resonant components of the acceleration signal all have fundamental frequencies that are distinct from one another, the peaks that they form in the power spectrum are not in the same places [99].

One objective of the third study (appendix A3) was to investigate the impact that muscle fatigue brought on by SRS has on the capacity of the human musculoskeletal system to dampen shock waves brought on by the impact of the foot striking the ground. When a person is fatigued, their musculoskeletal system is less able to prevent itself from being overloaded by the shock waves that are generated when they touch the ground with their foot. An increased shock wave amplitude, as measured on the tibial tuberosity, may be an indication that the victim has lost some level of protection.

The results were achieved by following this protocol while the subjects ran on a treadmill that was powered by an electric motor. The collecting of data was made more simpler with such a set-up, although the patterns of locomotion that were found may be different from those seen whilst running over land. The participants in this study were required to maintain a steady running speed for the duration of the experiment, regardless of whether or not they were experiencing signs of exhaustion. The introduction of the association between running fatigue and lower limb injuries was one of the primary aims of the SRS study; nevertheless, it is vital to keep in mind that this may be slightly different when referring to running on overground surfaces: As runners begin to feel the effects of weariness, they may choose to lower their pace as a preventative measure. As a consequence of this, the state of weariness might begin to lift, and if this occurs, the acceleration data might not show any further increase.

According to the findings of this study, there must be some level of fatigue present for there to be an increase in the acceleration data. Because of this, the findings of this study can be generalised to apply to running on land, assuming that fatigue is a factor.

Also, the majority of running injuries in overground running are injuries to the lower extremities, with a predominance of knee injuries. Our findings indicate an increase in acceleration data in the tibial tuberosity, which tends to support this extrapolation because it tends to indicate a correlation between the two. Previous research has demonstrated that the loading rate of the lower limb has a direct and strong correlation with the pace at which a person is running. Furthermore, as can be seen in figure 71, the vertical impact force increases as the speed at which a person is running increases.

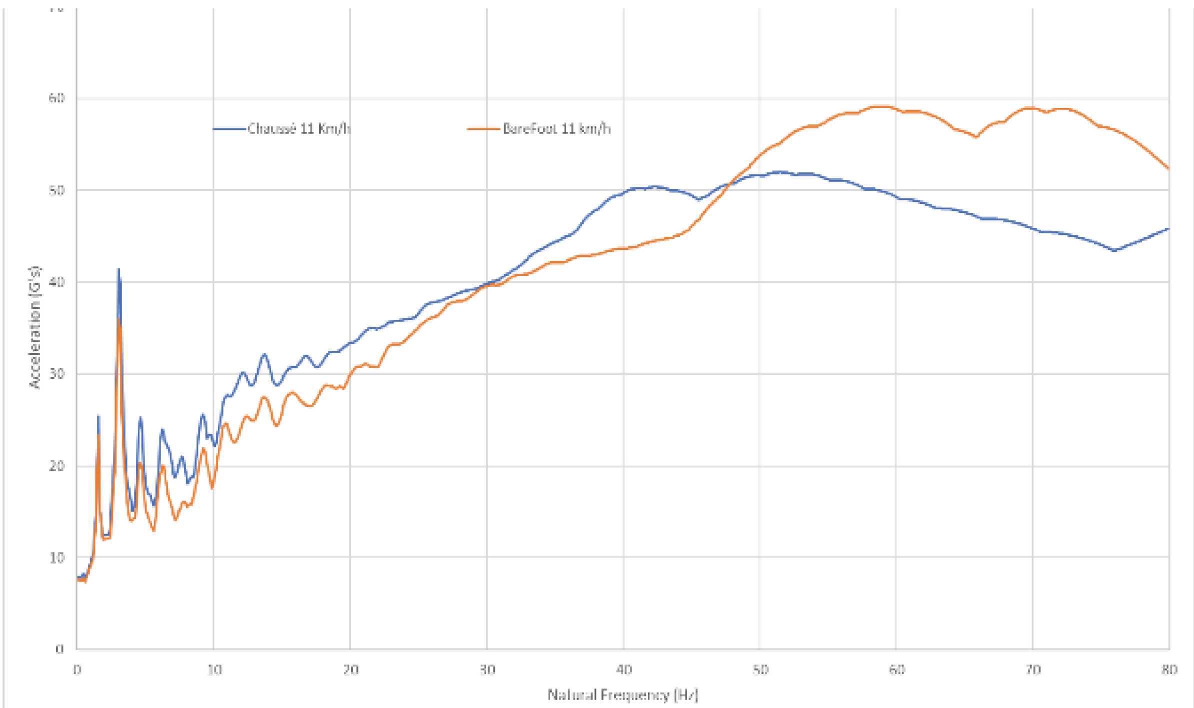


Figure 74: Experimental SRS Comparison shod/barefoot at 11km/H

In our most recent research (appendix A4), one of our primary objectives was to investigate how the effects of fatigue brought on by SRS impacted the capacity of the human musculoskeletal system to dampen the shock waves caused by foot strikes.

Based on the findings of this research, it appears that the various components of the acceleration signal can be differentiated in the frequency domain by means of spectral analysis, as illustrated in Figure 72. This is relevant to the analysis of impact shock that occurs while the subject is running.

Figure 75: Experimental example of SRS results for one CrossFit athlete extracted for three runs on both feet.

The idea behind this concept is that exhaustion reduces the capacity of the human musculoskeletal system to defend itself against overloading brought on by the generation of shock waves by the foot strike. This loss of protection may be seen as an increase in the amplitude of the shock wave. As can be seen in Figure 10, there was a direct association between exhaustion and an increase in the aggressiveness of the SRS, and this was the case for all five of the athletes.

We found that the average SRS peak for all five athletes during the third run was considerably larger than during the first run and the second run (p 0.01) at the same natural frequency as the athlete. This was observed during the third run (figure 73). Our hypothesis that exhaustion leads to a reduction in the capacity of the musculoskeletal system to absorb shock and, as a result, a possibly increased risk of overuse injury has been proven correct by these findings.

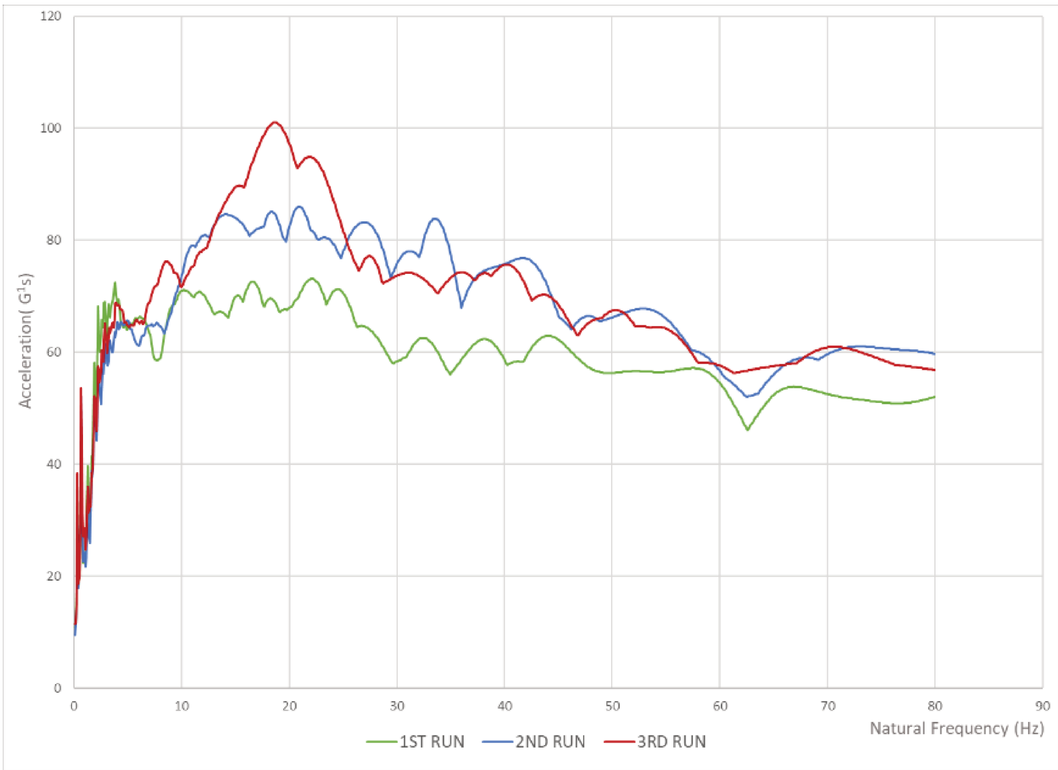


Figure 76: Experimental average SRS peaks for every athlete.

It is possible to speculate that when fatigue sets in, athletes may slow down in order to protect themselves from additional injury. As a direct result of this, the person may have recovered from their state of exhaustion; if this is the case, the acceleration data may not have increased. Based on the findings of the investigation that we carried out, this was not the case.

Previous studies have shown that the loading rate of a person's lower limb has a direct and strong correlation with the speed at which they are running, and that the vertical impact force increases as the speed at which a person is running increases [87]. This correlation is supported by the finding that the vertical impact force increases with the speed at which a person is running.

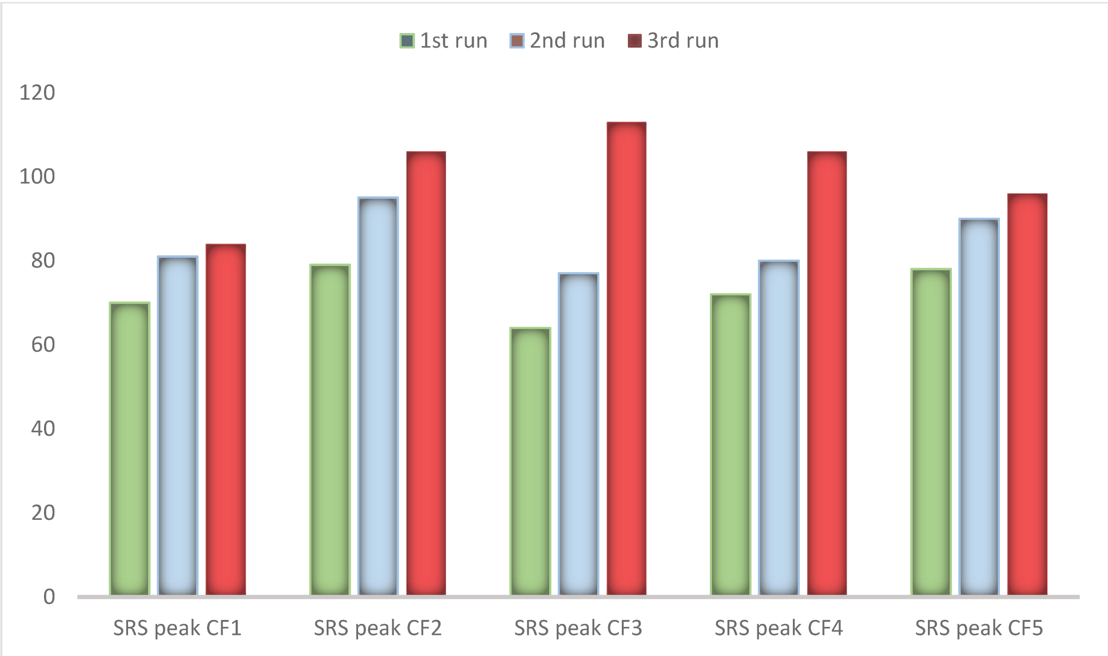


Figure 77: Experimental individual SRS peaks for each CF athletes

In conclusion, when muscles are recruited, the bending force that is exerted on bones is minimised, and the peak dynamic loads that could potentially cause damage to musculoskeletal tissues are controlled. This is because bones are more stable when there is less bending tension placed on them. Previous research has shown that weary muscles are unable to maintain "optimal" running, and they also claimed that exhaustion in runners may lead to adjustments in the mechanics of the landing phase of their stride. These findings were published in the journal Sports Medicine. It was also demonstrated that as muscle fatigue sets in, the amount of mechanical energy that can be transferred between the eccentric and concentric phases declines by a large amount. This was demonstrated by the fact that there was a direct correlation between the two phases. It has been hypothesised that these kinds of modifications play a function in the beginning stages of the injury process [101-102-103].

CHAPTER 4

Discussion and conclusion

In this chapter we will first explore the definition, mechanism and prevalence of overuse injuries and how they can plague a runner’s physical activity. We will then cover the most typical lower limb overuse injuries that occur in runners. We will then talk about the results of our finding and correlate the biomechanics gait analysis, our novel SRS approach and how to potentially prevent overuse injuries through embedded technologies in connected wearable. We will then draw our conclusions and perspectives from our research the current state of knowledge.

1. Overuse injuries

When energy is delivered to the body in quantities or rates that surpass the threshold for human tissue damage, injuries occur. Typically, while discussing sports injuries, one refers to mechanical energy transfer. Typically, these conceptual definitions give lead to the management of injuries that meet particular requirements for time-loss or medical care. In fact, according to a recent consensus statement on injury definitions and data collection procedures in football (soccer), injuries are: "Any physical complaint sustained by a player as a result of a football match or football training, regardless of the need for medical attention or time-loss from football activities." [105].

In the past, several definitions of injury have been given that were occasionally identical; this heterogeneous language can be confusing. For Junge et al. [106], an injury was defined as "any musculoskeletal complaint freshly obtained owing to participation and/or training during the tournament that got medical attention regardless of the absence from competition or training".

Important aspects of this injury definition include: (1) all injuries that received medical attention (not just time loss or reduced performance), (2) newly incurred (exclusion of pre-existing and not fully rehabilitated injuries), (3) injuries occurring during competition or training, and (4) exclusion of illnesses and diseases. This expansive definition of damage makes it easy to evaluate the effects of the entire spectrum of injuries, from small contusions to fractures. This may be relevant for analysing the long-term effects of injuries, as an examination of injury sequences indicates that minor injuries are frequently followed by moderate or severe injuries, and that acute complaints are predictive of subsequent injuries. Therefore, a "all-inclusive" injury definition precludes the physician from deciding which injuries should or should not be included.

The availability of additional information regarding time loss (estimated period of future absence from sport) permits the expression of the incidence of time-loss injuries and the possibility of comparing the results to those of other research employing the same definition. Zemper observed in 2005 [107] that none of the nine research chosen for his review study on track and field injuries used the same definition of an injury, making it difficult to compare data across studies.

He suggested that the most commonly used (and recommended) definition of a reportable injury in sport injury epidemiology was: "an injury sustained during participation in the sport, requiring medical attention at some level (e.g., coach, school nurse, trainer, and physician), and preventing the athlete from normal full participation for the remainder of that competition/training session or one day after the injury."

The concept of time loss prevents the data collecting system from being overwhelmed by minor injuries that do not impede normal participation. It is not recommended to utilise definitions incorporating longer time loss periods (e.g., two days, one week) since they may overlook many of the more minor ailments. In conclusion, the definition provided by Dick et al. in [108] the ISS appears to be a suitable summary. They proposed that a reportable injury be defined as "one that (1) occurred as a result of participation in an organised practise or competition, (2) required medical attention from a team-certified athletic trainer or physician, and (3) resulted in a restriction of the student-participation athlete's or performance for one or more calendar days beyond the date of injury." Athletic trainers were tasked with determining whether an injured athlete would have been able to compete if he or she had been given a day off following an injury incident. "Running-related (training or competition) musculoskeletal pain in the lower limbs that restricts or prevents running (distance, speed, duration, or training) for at least 7 days or 3 consecutive scheduled training sessions, or requires the runner to consult a physician or other health professional."

Running is one of the most popular and accessible sports practised by people all over the world, and it has grown in popularity over the last 50 years. The number of runners and running events has increased significantly over the past decades due to the low cost and ease of implementation with minimal equipment by a wide range of people.

More importantly, running is a good type of exercise for people who want to improve their physical fitness and/or live a better lifestyle because it has been related to lifespan and a reduction in risk factors for cardiovascular disease. Despite these health benefits, runners are prone to running-related musculoskeletal injuries (RRMIs) [109].

These RRMIs are typically induced by the application of relatively light loads over a large number of repeating cycles. Several studies have looked at the proportion of injuries (incidence and prevalence rates) among runners, with incidence rates ranging from 3.2 to 84.9 percent. This wide range may be explained by changes in study designs, injury classifications, subject characteristics, and follow-up periods, all of which might vary between studies.

In epidemiological investigations, incidence and prevalence are fundamentally different, although both are significant. Incidence is a measure of the number of new athletic injuries that occur. It conveys information regarding the possibility of being injured and is typically only available in prospective research. Prevalence describes how common an injury is in the sample population and is typically documented in retrospective investigations. As a result, developing efficient injury prevention programmes may lower injury incidence and, consequently, injury prevalence.

Running is one of the most common activities that leads to lower back and lower extremity overuse problems. Each year, 50 percent of runners suffer an injury that prevents them from running for an extended length of time, and 25 percent of runners are injured at any given time [110].

Overuse injuries account for roughly 70% to 80% of running diseases, primarily affecting the knee, ankle/foot, and shank anatomic locations. According to Francis et al., the most prevalent overuse injury is patellofemoral pain syndrome, however Lopes et al. suggest that the most common RRMI is medial tibial stress syndrome. These two systematic reviews represented in figure 74 used different methodological techniques, which could explain why they did not reach the same conclusion.

Some studies claim that acute injuries when running are uncommon, while others claim that they are prevalent, primarily consisting of ankle sprains and muscular injuries (e.g., quadricep and hamstring strains).

There is still debate over whether an ankle sprain is a prevalent ailment among runners. Although Francis et al. discovered that ankle sprains were not among the top ten most common RRMIs, Lopes et al. [111] discovered that ankle sprains were among the top five most common RRMIs.

This could be due to the fact that Lopes et al. focused on prospective studies (incidence) to identify the most common RRMIs, whereas Francis et al. [112] pooled the number of injuries across all research designs (prevalence) to determine the most common RRMIs. This emphasises the need of distinguishing between incidence (prospective studies) and prevalence (retrospective studies).

RRMIs, regardless of the kind of injury, reduce enjoyment in exercise and are related with unfavourable outcomes such as significant cost implications, temporary or permanent termination of running, and absence from work.

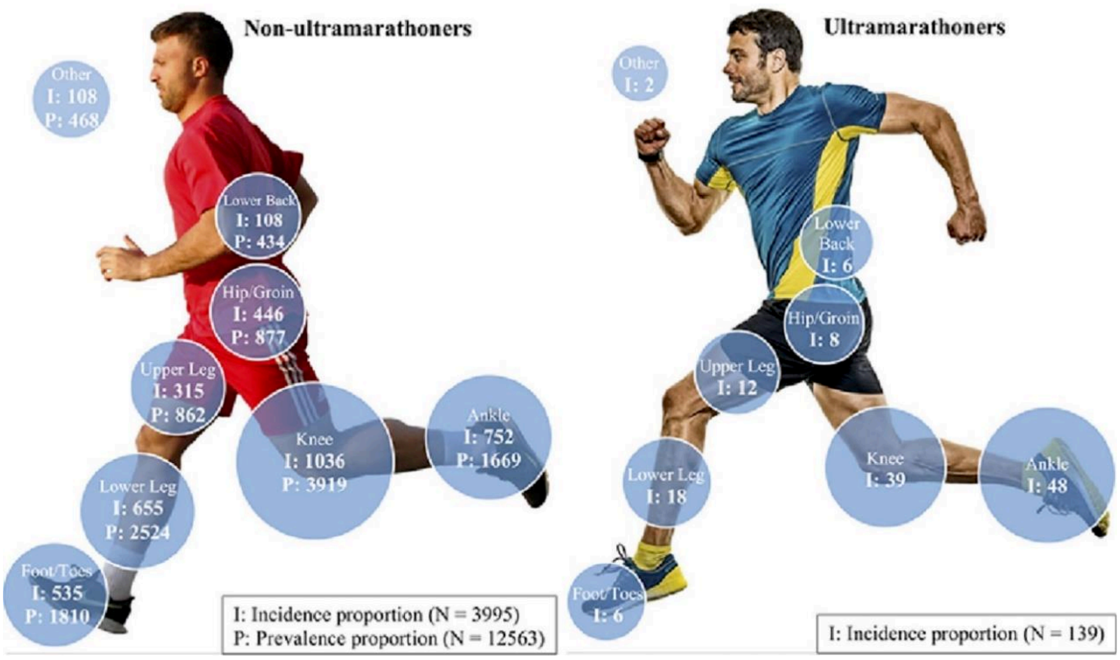


Table II. Incidence rates of running-related musculoskeletal injuries

Study, year	Study design (follow-up period in months)	Population	Musculoskeletal injury definition	Injuries data collection	RRMIs	Incidence [% (n)]
Pileggi et al.,[23] 2010	Prospective cohort (12)	18 amateur runners, running at least 5 ×/wk and runners who aimed to run 50 km/wk	Interference in training: grade I (no interruption); grade II (volume reduction); grade III (interruption for at least 2 wk). Symptoms duration: acute (until 2 wk); subacute (2–6 wk); chronic (>6 wk)	Injuries were reported by the runners by telephone or in person and evaluated by a physician	Patellar tendinopathy	22.7 (5)
					Medial tibial stress syndrome	13.6 (3)
					Iliotibial band syndrome	9.1 (2)
					Achilles tendinopathy	9.1 (2)
					Tibial stress fracture	9.1 (2)
					Retrochanteric bursitis	9.1 (2)
					Calf muscle injury	4.5 (1)
					Hip adductor muscle injury	4.5 (1)
					Iliac crest stress fracture	4.5 (1)
					Infrapatellar bursitis	4.5 (1)
					Plantar fasciitis	4.5 (1)
Jakobsen et al.,[24] 1994	Clinical trial[a] (12)	20 marathon runners	Any injury of the musculoskeletal system that was sustained during running and prevented training or competition	Injuries were reported by the runners in a questionnaire and evaluated by a physician	Medial tibial stress syndrome	20.0 (4)
					Sprain ankle joint	15.0 (3)
					Achilles tendonitis	10.0 (2)
					Muscle fibre rupture	10.0 (2)
					Runner's knee[b]	10.0 (2)
					Plantaris fasciitis	10.0 (2)
					Sprain knee joit	5.0 (1)
					Costal fracture	5.0 (1)
					Other	20.0 (4)
Lysholm and Wiklander,[10] 1987	Prospective cohort (12)	60 runners: 28 long-distance/marathon; 13 middle distance; 19 sprinters	Injuries that markedly hampered running training or competition for at least 1 wk	Injuries were reported by the runners in a running diary and evaluated by a physician	Medial tibial stress syndrome	14.5 (8)
					Hamstrings strain	10.9 (6)
					Ankle sprain	10.9 (6)
					Achilles tendinitis	10.9 (6)

Figure 78 : Systematic review of running-related musculoskeletal injuries in runners (Nicolas Kakouris, 2021)

A detailed understanding of the most common RRMIs is a necessary step in developing effective injury prevention programmes and rehabilitation intervention strategies that can minimise the high incidence and prevalence of RRMIs. Figure 74 shows that 70 percent of all RRMIs are due to overuse. Furthermore, the reported injuries are mostly at or below the knee. This could be because propulsion is generated mostly by the lower leg during normal running, resulting in an increased biomechanical load on these structures.

The knee, ankle, and lower leg account for the majority of injury incidence, while the knee, lower leg, and foot/toes account for the majority of injury prevalence. The most common injuries are Achilles' tendinopathy (10.3 percent), medial tibial stress syndrome (9.4 percent), patellofemoral pain syndrome (6.3 percent), plantar fasciitis (6.1 percent), and ankle sprains (5.8 percent), while patellofemoral pain syndrome (16.7 percent), medial tibial stress syndrome (9.1 percent), plantar fasciitis (7.9 percent), iliotibial band syndrome (7.9 percent), and Achilles' The ankle (34.5%), knee (28.1%), and lower leg (12.9%) were the three most often damaged areas [109].

An overuse injury is one that happens as a result of tissue damage caused by recurrent demand over time, as opposed to an acute injury such as a shoulder dislocation or ankle sprain [113]. In young athletics, these injuries can affect the muscle-tendon unit, bone, bursa, neurovascular systems, and the physis (growth plate). Overuse injuries commonly include shoulder impingement, lateral epicondylitis (tennis elbow), tendinitis, and stress fractures. An overuse injury is generally caused by:

- **Training errors.** Training errors can occur when you take on too much physical activity too quickly. Going too fast, exercising for too long or simply doing too much of one type of activity can strain your muscles and lead to an overuse injury.
- **Technique errors.** Improper technique also can take its toll on your body. If you use poor form as you do a set of strength training exercises, swing a golf club or throw a baseball, for example, you may overload certain muscles and cause an overuse injury.

Errors in training, such as ramping up an activity too quickly or practising for too long without proper rest and recovery, can lead to overuse injuries. These can also arise when performing only one specialised exercise that utilises certain muscles or bones, such as repetitive pull-ups, or when only one sport is done year-round. Poor technique can also contribute to overuse injuries in which the tissue is repeatedly and improperly overloaded. This can be observed during strength training activities such as bench presses and squats, as well as during the actual sporting activity, such as baseball pitching or golf club hitting.

There are several ways in which overuse injuries can potentially be prevented, including:

- Limiting exercise time to allow adequate rest and recovery
- Limiting the number of specific repetitive movements (i.e., the number or repetitions in a specific workout routine or certain sport-specific activities such as pitch counts)
- Making sure you're using the correct technique and proper equipment when starting a new activity
- Aiming for a gradual increase to achieve your workout goals rather than increasing your activity level too quickly

Different definitions of overuse have separated epidemiologic research from clinical practise by imposing limits. In injury monitoring, clinicians reported overuse as a mechanism of injury, a category for diagnosis, or both a mechanism of injury and a category for diagnosis, as documented in the systematic review by Roos and Marshall [114]. The authors recommended that doctors and researchers only use the word overuse when referring to a "gradual onset mechanism with an underlying aetiology of recurrent microtrauma." Clinicians may overlook the underlying pathologic issue or misrepresent the aetiology of an injury by referring to it as overuse. Rather of addressing the symptoms by the cause of damage, clinicians typically utilise conventional diagnostic words for musculoskeletal injuries. In clinical practise, rotator cuff tears can be produced either by an acute mechanism, such as falling on an extended hand, or by an overuse mechanism, such as repetitive throwing. The mechanism of damage is essential for comprehending the aetiology of a pathologic symptom, and it would be confounding for doctors if all rotator cuff cases were recorded as overuse rather than by the actual mechanism.

Table. Working Definitions Related to Injury Surveillance

Term	Working Definition
Injury[8-10]	Any physical complaint sustained by a player that results from a practice or competition, irrespective of the need for medical attention or time loss from activity
Overuse[3]	Mechanism of gradual onset with an underlying pathogenesis of repetitive microtrauma
Recurrent injury[8-10]	Injury of the same type and at the same site as an index (initial) injury which occurs after a player's return to full participation
Injury severity[8-10]	The number of days that have elapsed from the date of injury to the date of the player's return to full participation in training and availability for competition selection: • Slight: 0–1 d • Minimal: 2–3 d • Mild: 4–7 d • Moderate: 8–28 d • Severe: more than 28 d • Career ending • Nonfatal catastrophic • Catastrophic

Figure 79 : Injury classification (Dick, R., Agel, J., & Marshall, S. W. 2007)

Injury-surveillance systems may have produced erroneous results due to the lack of a functioning definition. To bridge the gap between epidemiological research and clinical practise, doctors should utilise the established working definitions (Figure 75) to characterise the present health condition of their patients. In addition, acceptable functional surveillance definitions should be consistent with the categorization and coding methodology of the International Statistical Classification of Diseases and Related Health Problems. Utilizing a precise working definition of overuse will aid in the process of finding modifiable risk factors and causal processes that researchers and clinicians can target in order to design interventions to prevent overuse-related injuries. This will allow academics and doctors study and attempt to comprehend how sport participation affects these persons' long-term health-related quality of life. This transition necessitates that clinician correctly identify and classify injury processes and injury classifications. It fosters improved medical documentation by clinicians and enables researchers to precisely evaluate the risk of sport involvement through injury surveillance programs [115].

2. Lower limb injuries

The knee, ankle, and lower leg account for the highest percentage of injury incidence, whereas the knee, lower leg, and foot/toes account for the highest percentage of injury prevalence. Achilles's tendinopathy (10.3%), medial tibial stress syndrome (9.4%), patellofemoral pain syndrome (6.3%), plantar fasciitis (6.1%), and ankle sprains (5.7%) account for the greatest proportion of injury incidence, whereas patellofemoral pain syndrome (16.7%), medial tibial stress syndrome (9.1%), plantar fasciitis (7.9%), iliotibial band syndrome (7.9%), and Achilles' Ankle (34.5%), knee (28.1%), and lower leg (12.9%) were the three most frequently injured locations [109].

Achille's Tendinopathy

A common overuse ailment, Achilles tendinopathy is produced by recurrent energy storage and release with severe compression [116]. This can result in a sudden injury or, in the worst scenario, a ruptured Achilles tendon. In both instances, a lack of flexibility or a stiff Achilles tendon might increase the likelihood of sustaining these injuries. The word tendinopathy is currently recommended to characterise this group of individuals. Cook and Purdum [117] developed a new approach to tendon discomfort, which they dubbed the Tendon Continuum. The continuum model offered a stage system for tendinopathy based on the variations and distribution of tendon disorganisation.

Three stages are as follows:

- Reactive tendinopathy
- Tendon disrepair
- Degenerative tendinopathy

It has been proposed that the tendon can move up and down this continuum, particularly in the early phases of tendinopathy, by adding or reducing load. Achilles tendinopathy can be classified as either insertional or midportion, depending on its location. The insertional form is positioned at the transition between the Achilles tendon and the bone, whereas the midportion form is located at the tendon body. Surgical specimens reveal a variety of degenerative tendon alterations, including tendon fibre structure and organisation as well as an increase in glycosaminoglycans, which may explain the tendon swelling.

The exact aetiology of tendinitis is yet unknown. Even while tendonitis of the Achilles tendon is frequently associated with athletic participation, the condition is equally common in non-athletes. The most significant cause is severe tendon overload. There may be latent deterioration of the Achilles tendon, but the discomfort only manifests when the tendon is overworked. It is also noticed that trauma does not typically precede the condition.

The Achilles tendon is the body's largest and strongest tendon. The tendon can withstand substantial tensile stresses. It originates at the distal juncture of the gastrocnemius and soleus muscles and inserts at the base of the calcaneus (figure 76). The usual structure of a tendon is comprised of thin, cylindrical cells and extracellular matrix. Tendon cells, specifically tenocytes and tenoblasts, are responsible for the creation of all extracellular matrix components. We find bundles of type I collagen and elastin within the matrix. This type-I collagen is accountable for the tendon's strength. Between the collagen, a ground substance composed of proteoglycans and glycosaminoglycans is located.

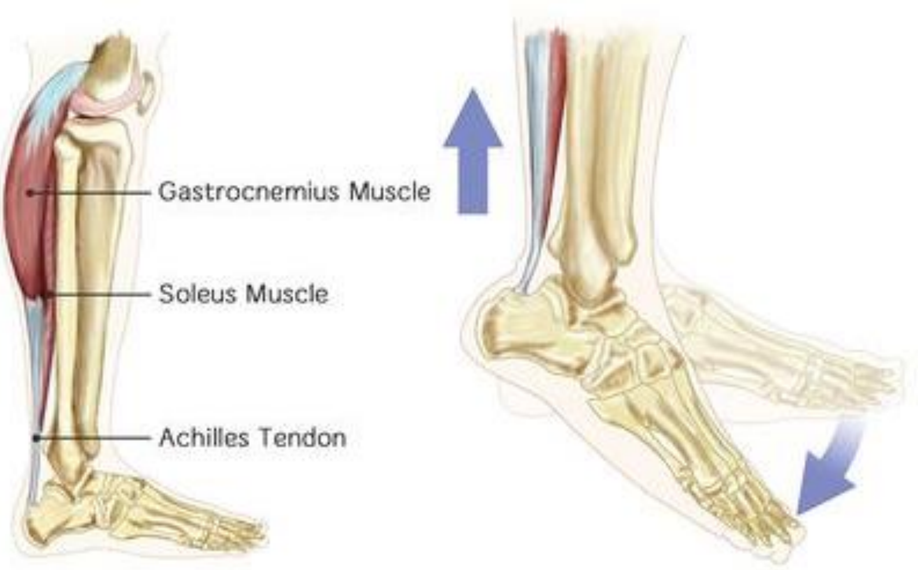


Figure 80 : Achille's tendon (Netter, F. H. 2010. Atlas of Human Anatomy)

The Achilles tendon is surrounded by paratenon, which functions as an elastic sleeve that permits the tendon to move freely within the surrounding tissue. The paratenon is composed of a layer of cells and is responsible for the tendon's blood transport. Under the paratenon, the next layers are created in chronological order: the epitenon, which is a thin membrane, and the endotenon, which surrounds the collagen fibres and forms bundles. The tendon's blood supply is weak along its entire length, as indicated by the tiny number of blood vessels per cross-sectional area, particularly in the region 4-6 cm above the calcaneus. Inadequate vasculature may characterise a sluggish healing rate after trauma.

A reactive tendon is the initial stage on the tendon continuum and is characterised by a non-inflammatory proliferative response in the cell matrix. This is due to excessive compressive or tensile force. Straining the tendon during physical activity is regarded as one of the most significant pathogenic stimuli, and systematic overloading of the Achilles tendon beyond its normal limit can result in microtrauma. Repetitive micro-traumas caused by a non-uniform tension between the gastrocnemius and soleus result in frictional forces between the fibres and aberrant loading concentrations in the Achilles tendon. This may result in tendon sheath inflammation, tendon degeneration, or a combination of both. Without the minimum recuperation time, this can result in tendinopathy.

It has been claimed that decreased arterial blood flow, local hypoxia, decreased metabolic activity, nutrition, and a persistent inflammatory response may contribute to chronic tendon overuse injuries and tendon degeneration.

An ankle misalignment produced by overpronation of the foot is the most prevalent and arguably the most serious misalignment. It has been suggested that increased foot pronation is connected with Achilles tendinopathy.

In acute trauma, external forces predominate, whereas overuse injuries typically have a complex cause. Inflammation and edoema creation characterise the acute phase of Achilles tendinopathy, which is triggered by abrupt overload, blunt trauma, or rapid muscle fatigue. If the therapy of the acute phase fails or if it is neglected, fibrin and adhesions might form on the tendon.

Reactive tendinopathy can develop to tendon degeneration if the tendon is not deloaded and allowed to revert to its normal state. During this phase, there is a continuance of increased protein synthesis, which has been demonstrated to result in collagen separation and cell matrix disorder (figure 77). This is an attempt at tendon healing similar to the first phase, but with increased physiological involvement and breakdown. Degenerative tendinopathy is the ultimate step on the continuum, and it is believed that there is a dismal outlook for the tendon at this point, as the changes are irreversible. Often, tendon degeneration and peri-tendinous adhesions occur together, although this does not entail that one condition causes the other.

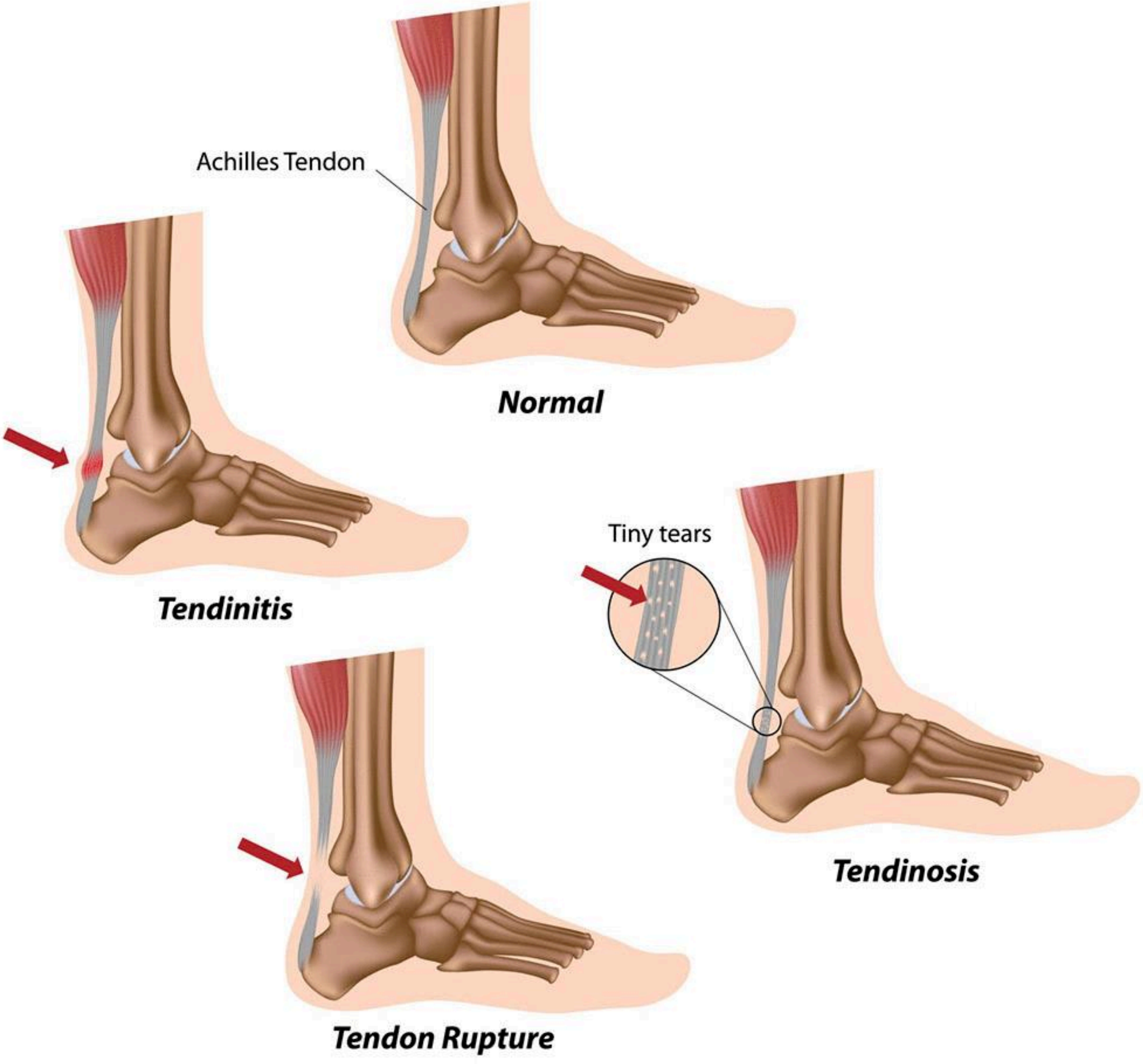


Figure 81 : Achille's tendinopathy classification (https://www.healthdirect.gov.au)

Tendinopathies can be caused by overuse, poor circulation, lack of flexibility, gender, endocrine, and metabolic variables. This recurrent strain (typically eccentric in nature) disrupts the tendon's structure, and collagen fibres migrate together, break the crosslinks, and denature the tissue, causing inflammation. This cumulative microtrauma is hypothesised to damage not just the collagen cross-linking, but also the collagenous matrix and the vascular components that influence the tendon, resulting in tendinopathy.

In addition, current research indicates that older age, a larger android fat mass ratio, and a waist circumference greater than 83 centimetres are connected with an increased risk of Achilles Tendinopathy in men. Additionally, the existence of the COL5A1 gene variation was identified as a potential risk factor. This gene is generally responsible for the creation of tendon protein; however, people with the disorder had significantly different allele frequencies of the COL5A1 BstUI RFLP compared to healthy individuals. In addition to misuse and degradation, it was claimed that Achilles Tendinopathy has a substantial metabolic influence due to its poor anatomical vascularity, correlation with body fat, and genetic factor. In addition, a recent study discovered that individuals with chronic Achilles tendinopathy exhibited both peripheral and central pain sensitivity. Morning discomfort is a characteristic sign of Achilles tendonitis because the tendon must tolerate a full range of motion, including stretching, immediately upon arising in the morning.

Typically, symptoms are restricted to the tendon and its immediate surroundings. Pain and swelling are less common. In the A-P and M-L planes, the tendon may appear to have slight changes in outline, becoming thicker. Tendinopathy of the Achilles tendon with a sensitive zone and intra-tendinous swelling that moves with the tendon and whose sensitivity increases or decreases when the tendon is subjected to pressure would have a high predictive value for tendinosis. Compared to the unaffected side, the damaged side of the tendon has a larger diameter, greater stiffness, and lower strain.

Plantar fasciitis

Plantar fasciitis produces pain at the heel's bottom. The plantar fascia is a thick, web-like ligament connecting the heel to the forefoot [118]. It helps you walk by acting as a shock absorber and supporting the arch of your foot. Plantar fasciitis is among the most prevalent orthopaedic conditions. The plantar fascia ligaments endure significant wear and tear on a regular basis. Too much pressure on the foot might cause ligament injury or rupture. The plantar fascia becomes irritated, causing discomfort and stiffness in the heels. Unknown is the source of plantar fasciitis discomfort.

The illness may include degeneration rather than inflammation of the plantar fascia, according to 2003 research [119]. Because fasciitis means "inflammation of a fascia," plantar fasciosis may be a more appropriate term. Plantar fasciitis is caused by the degeneration of collagen in the plantar fascia's origin, the calcaneal tuberosity of the heel, and the surrounding perifascial tissues. Normal foot biomechanics are significantly influenced by the plantar fascia. The fascia plays a significant role in providing arch support and stress absorption.

Despite the inclusion of "itis" in the diagnosis, this illness is distinguished by the absence of inflammatory cells. There are several causes of plantar heel pain outside the plantar fascia; hence, the phrase "Plantar Heel Pain" should be interpreted broadly when discussing this and related pathology.

The plantar fascia is (figure 78):

- Composed of white longitudinally organised fibrous connective tissue that begins on the periosteum of the medial calcaneal tubercle, where it is thinner but extends into a thicker central portion. • The thicker central portion of the plantar fascia then extends into five bands surrounding the flexor tendons as it passes all five metatarsal heads. Pain in the plantar fascia may be insertional or non-insertional, and may encompass not only the bigger central band, but also the medial and lateral bands.
- Combines with the paratenon of the Achilles tendon, the intrinsic foot muscles, the skin, and the subcutaneous tissue.
- The thickhickcoelastic multilobular fat pad is responsible for absorbing up to 110 percent of body weight during walking and 250 percent of body weight during running, and deforms most during barefoot walking as opposed to shoed walking.

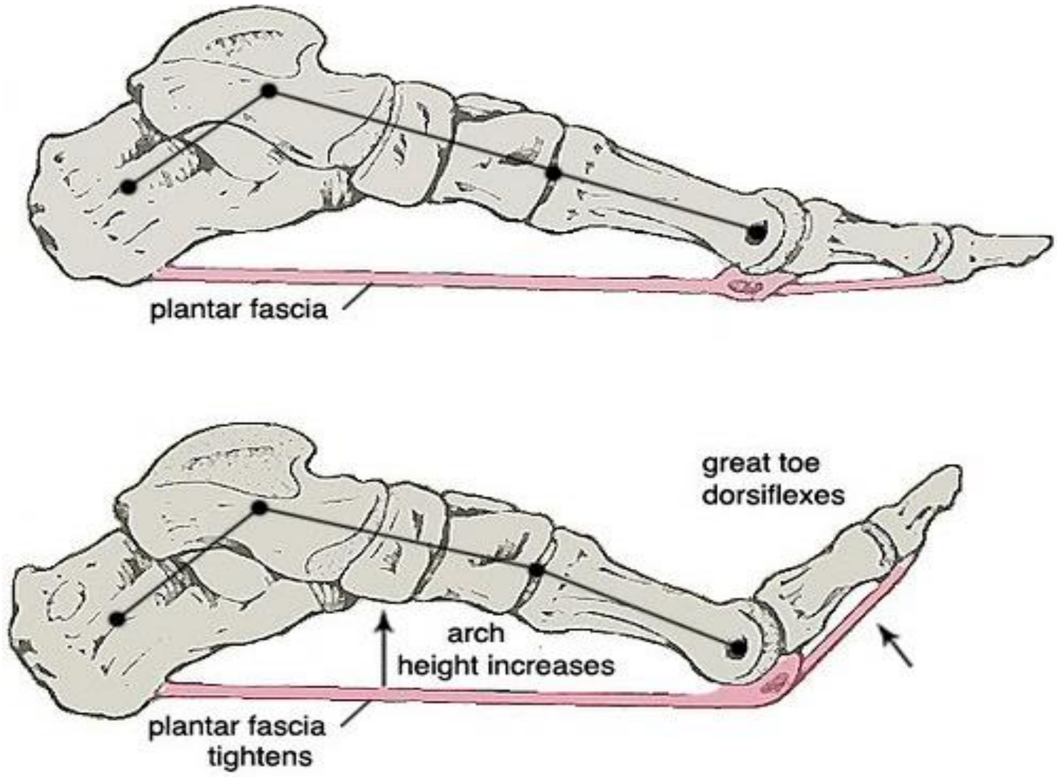


Figure 82 : Plantar fascia function (Hicks JH. The mechanics of the foot, II: the plantar aponeurosis and the arch. J Anat. 1954)

During weight-bearing:

- Tibia loads the foot “truss” and creates tension through the plantar fascia (windlass mechanism).
- The tension created in the plantar fascia adds critical stability to a loaded foot with minimal muscle activity.

This is often an overuse injury that is primarily due to a repetitive strain causing micro-tears of the plantar fascia but can occur as a result of trauma or other multifactorial causes. There are many risk factors which contribute to plantar heel pain including but not limited to:

- Loss of ankle dorsiflexion (talocrural joint, deep or superficial posterior compartment)
- Pes cavus OR pes planus deformities
- Excessive foot pronation dynamically
- Impact/weight-bearing activities such as prolonged standing, running, etc
- Improper shoe fit
- Elevated BMI > kg/m2
- Diabetes Mellitus (and/or other metabolic condition)
- Leg length discrepancy
- Tightness and/or weakness of Gastrocnemius, Soleus, Tendoachilles tendon and intrinsic muscle.

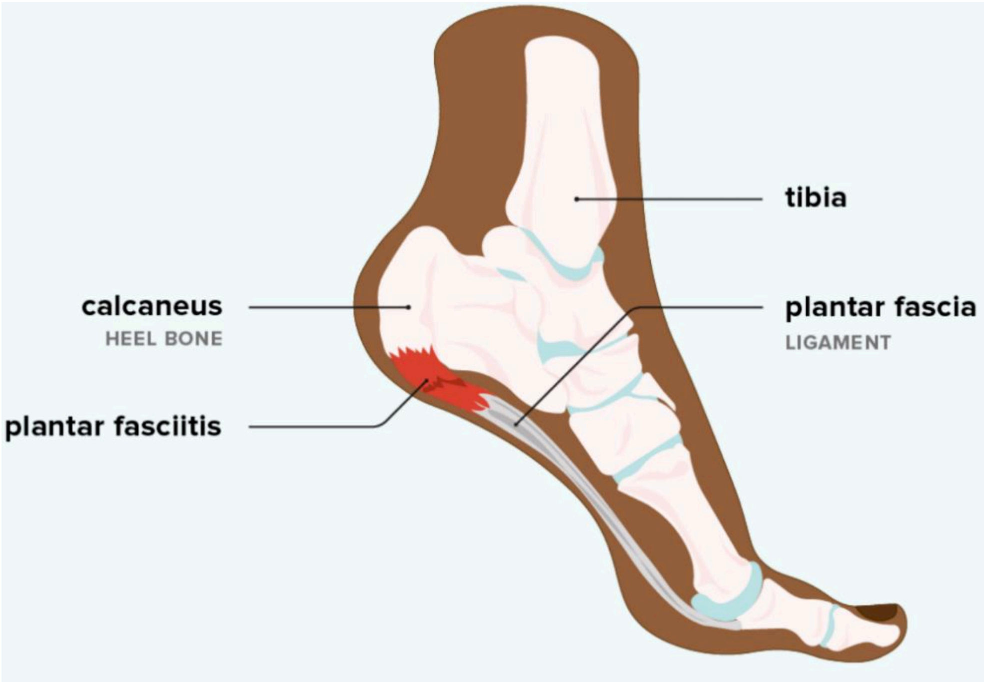


Figure 83 : Plantar fasciitis (https://www.healthline.com)

Plantar fasciitis is the most common cause of heel pain in the outpatient environment (Figure 79). Unknown are the incidence and prevalence of plantar fasciitis by age. It accounts for around 10% of all running-related injuries [120]. (Some research indicates prevalence rates among runners as high as 22 percent), it is believed to affect approximately 10 percent of the overall population:

- 83 percent of these patients are active, employed adults between the ages of 25 and 65.
- Represents 11 to 15 percent of all foot symptoms requiring medical attention. • May manifest bilaterally in one-third of cases.
- The average episode of plantar heel pain lasts longer than six months and affects up to ten to fifteen percent of the population. Approximately 90% of cases are effectively treated with conservative therapy.
- Plantar heel is somewhat more common in females than in males.
- It is estimated that this illness causes up to two million patient visits annually in the United States and accounts for one percent of all visits to orthopaedic clinics.
- Plantar heel pain is the most prevalent foot issue treated in physical therapy clinics, accounting for up to 40 percent of all patients seen in podiatric clinics.

Medial tibial stress syndrome

Medial Tibial Stress Syndrome (MTSS) or Shin-Splint Syndrome is a clinical pain condition defined as exercise-induced pain along the posteromedial tibial border (distal third) caused by repetitive loading stress during running and jumping and provoked on palpation over 5 consecutive centimetres [121]. The American Medical Association (AMA) described shin splints in The Standard Nomenclature of Athletic Injuries as "pain and discomfort in the leg from recurrent jogging on hard surfaces or forcible, excessive use of the foot flexors." "A more accurate word that explains the inflammatory traction event in the tibial aspect of the common leg in runners is the medial periostitis of the tibial traction or simply the medial tibial periostitis" [96].

The pathophysiology of shin splints is more easily understood after examining the relevant cross-sectional anatomy. There are 4 muscle compartments in the leg:

- Anterior: this compartiment contains the tibialis anterior muscle, the extensor hallucis longus, the extensor digitorum longus and the peroneus tertius.

    1. The tibialis anterior dorsiflexes the ankle and inverts the foot.
    2. The extensor hallucis longus extends the great toe
    3. The extensor digitorum longus extends the other toes and assists in eversion as does the peroneus tertius.

- B) Deep posterior: this contains the flexor digitorum longus, the tibialis posterior and the flexor hallucis longus.

    1. The tibialis posterior plantar flexes and inverts the foot.
    2. The others are predominantly toe flexors.

- C) Superficial posterior: this is the gastrocnemius and soleus group; predominatly plantar flexors of the ankle.
- D) Lateral: this compartment contains the peroneus brevis and longus, mainly foot evertors

A malfunction of the anterior and posterior tibialis is frequently involved, and the attachment of these muscles might be the source of pain. MTSS is frequently linked with muscle imbalance and inflexibility, particularly tightening of the triceps surae (gastrocnemius, soleus, and plantaris muscles). Athletes with triceps surae muscle weakness are more susceptible to muscle exhaustion, resulting in altered running mechanics and tibial stress. Clinicians should also evaluate the hamstring and quadriceps muscles for rigidity and imbalance.

The incidence of shin splints ranges from 4 percent to 19 percent in athletic populations and from 4 percent to 35 percent in the military population. The most prevalent musculoskeletal injury in runners (sprinters, medium and long distance runners, and football players) has been identified as shin splints, with an incidence rate ranging from 13.6% to 20% and a prevalence of 9.5%. Additionally, 20 percent of the population of dancers have it, and up to 35 percent of new runners and dancers will develop it.

Shin splints are most prevalent among runners and jumpers who have made training mistakes, especially when they overtrain or run too quickly for their abilities. This injury may also be a result of modifications to the training regimen, such as an increase in distance, intensity, or duration. Running on a hard or uneven surface and unsuitable running shoes (such as a lack of shock absorption) may have contributed to the injury. The most frequently reported intrinsic variables [122] are biomechanical anomalies such as foot arch deformities, hyperpronation of the foot, and uneven leg length.

Women are especially susceptible to stress fractures when this syndrome is present. This is the result of nutritional, hormonal, and biomechanical irregularities. Overweight individuals are more susceptible to developing this syndrome. Before beginning therapy or a training programme, it is essential that obese individuals combine exercise with a diet or attempt to lose weight. These folks, along with those in poor physical condition, should always raise their training intensity gradually. Because cold weather contributes to this symptom, it is crucial to warm up properly (even more than usual). The pathophysiology is unknown, however there are two ideas to consider: periostitis caused by fascial tension or a local bone stress reaction. Internally, a persistent inflammation of the muscle attachment along the posterior medial tibia and bone abnormalities are regarded as the most probable causes of medial tibial stress syndrome.

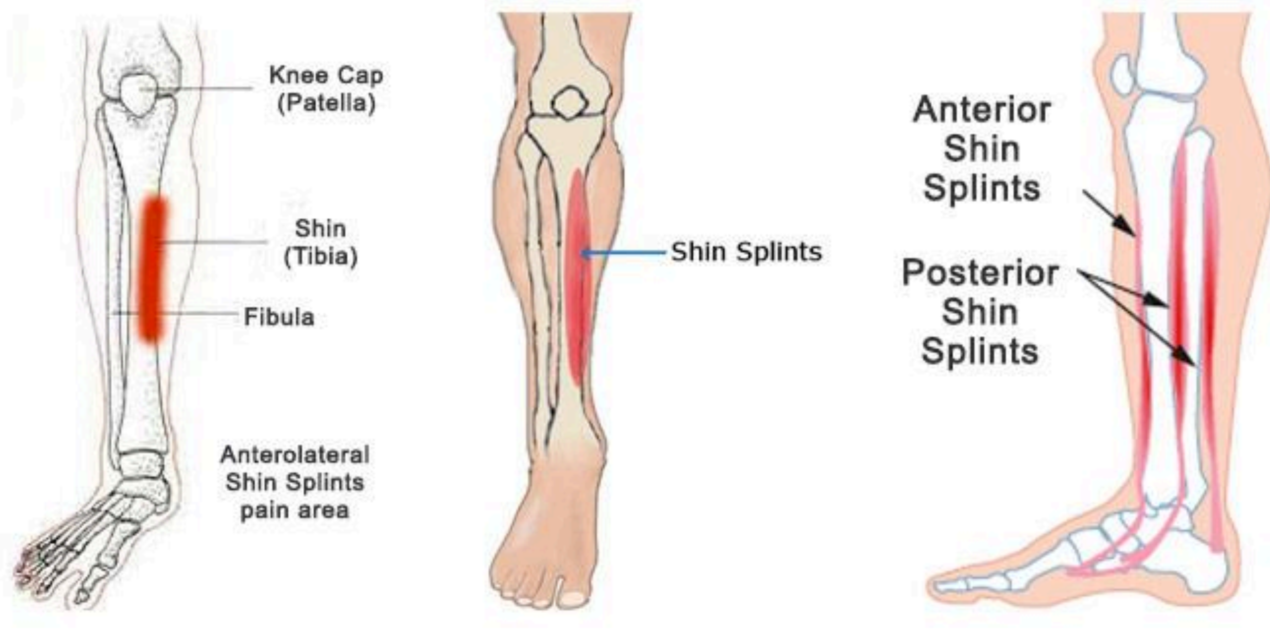


Figure 84 : Medial tibial stress syndrome (N Reshef, 2012)

Pain in the distal two-thirds of the posteromedial tibial border is the primary symptom. The discomfort is non-focal, but spreads "at least 5 cm" and is frequently bilateral. Additionally, it worsens with each contact. Mild swelling and discomfort on palpation may also be present in this uncomfortable location following the inciting action for up to several days.

Initially, the patient only experiences pain at the beginning of the workout, which often subsides throughout exercise and returns after the cool-down time. When shin splints worsen, the discomfort might persist during exercise and persist for hours or days after the triggering activity has ceased. The most frequent complication of shin splints is a stress fracture, which manifests as anterior tibial discomfort. Neurovascular symptoms are seldom attributable to MTSS, and when present, other pathologies, such as chronic exertion compartment syndrome (CECS) or vascular deficits, should be examined as the cause of leg discomfort.

Metatarsal stress fracture

Stress fractures are microscopic hairline fractures that can appear in the foot bones. Overtraining or overuse, poor training routines or surfaces, unsuitable footwear, flatfoot or other foot abnormalities, and osteoporosis can cause these conditions. If left untreated, these microscopic cracks in the bones of the feet can develop to a full fracture. Signs of a stress fracture include pain, edoema, bruising, and redness. 5 to 6 percent of all fractures treated in primary care are metatarsal fractures. This fracture can occur practically anywhere in the foot. It is the most prevalent foot injury. They are approximately five times as common than Lisfranc-dislocations. Equally distributed among men and women and all racial groups. The fracture distribution appears as follows: First metatarsal: 5 percent, Second metatarsal: 12 percent, Third metatarsal: 14 percent, Fourth metatarsal: 13 percent, Fifth metatarsal: 56 percent and Multiple metatarsal fractures: 15,6 percent [123].

Metatarsal injuries are widespread in both acute and chronic situations, and they are the most common location for stress fractures in the human skeleton. The most common metatarsal stress fractures involve the intermediate and distal sections of corpus ossis metatarsalis II or III. Stress fractures at the base of the first or second metatarsals (and sometimes other metatarsals) are uncommon. Athletes are susceptible to metatarsal stress fractures, particularly runners, who account for 20% of all lower extremity stress fractures. Due to the greater forces experienced by the second and third metatarsals while walking and running, stress fractures are most likely to occur in these metatarsals.

Metatarsal fractures can be caused by either direct or indirect assault, and can reveal a wide range of injuries, from simple, isolated fractures of a single metatarsal to crush injuries with serial fractures and severe soft tissue compromise. When a big object falls on the foot of a factory worker, direct trauma is common. When the leg and hindfoot are twisted while the forefoot is stationary, indirect damage ensues. The proportions are as follows: Supination injury: 48%; Fall from height: 26%; Crush injury: 12%.

Athletes, obese adults, and those with osteoporosis, rheumatoid arthritis, or diabetes are at a greater risk for having metatarsal fractures. It is also found in jogging, ballet, gymnastics, and high-impact aerobic exercises. A shoe's ability to absorb shock can avoid metatarsal stress fractures. Repetitive cyclic loading, particularly in the case of a young athlete or military recruit, can result in chronic overloading, predisposing an individual to a stress reaction and, eventually, fracture. It has been demonstrated that fracture pattern and injury severity vary with age and mechanism of injury. This relationship can also be connected with osseous development and age-related exercise levels.

The majority of fractures of the corpus ossis metatarsalis are fatigue fractures caused by persistent stress. As seen in athletes, ballet dancers, and soldiers, it is the result of repetitive force. Walking places greater force on the second and third metatarsals, resulting in increased stress. Therefore, stress fractures and bone remodelling in the second or third metatarsal are prevalent (figure 81). It is also prevalent among military recruits.

The metatarsal can fracture in three locations: the caput, the corpus, and the basis ossis. Thus, we can distinguish numerous distinct fractures: Subcapital fracture, Corpus ossis metatarsalis fracture, and Basis ossis metatarsal fracture.

Running places more force on the second and third metatarsals; consequently, stress fractures and bone remodelling from stress are common in the second and third metatarsals, a condition sometimes referred to as a "marathon runner's fracture" due to its prevalence among long distance runners [124, 125].

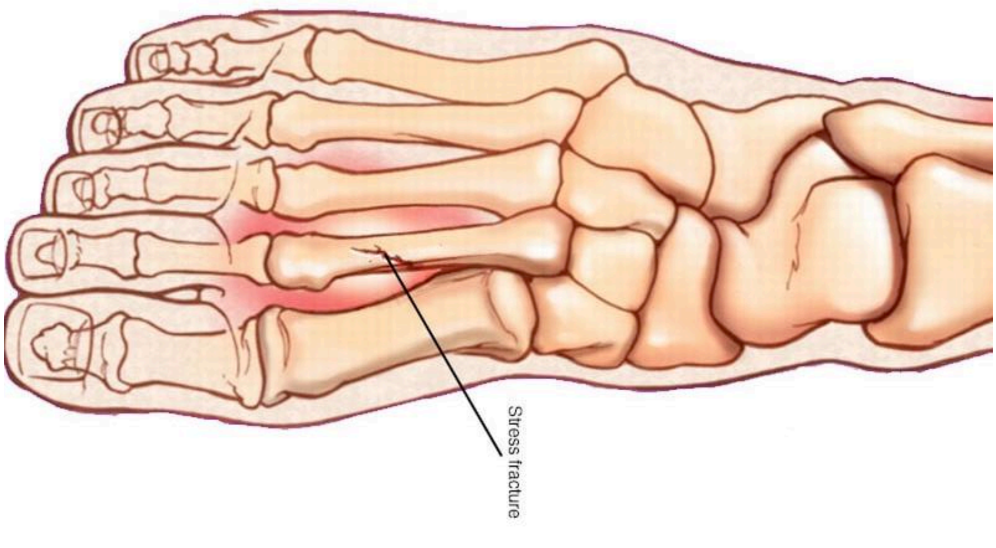


Figure 85 : Marathon runner's fracture (https://www.mayoclinic.org)

Patellofemoral Pain Syndrome

Patellofemoral Pain Syndrome (PFPS) refers to pain originating from the patellofemoral joint or associated soft tissues. It is a chronic condition that tends to aggravate with squatting, sitting, stair climbing, and running. Historically, this condition has been referred to as anterior knee pain, however this term is misleading because the pain can be felt in all areas of the knee (including the popliteal fossa). The most prevalent PFPS in sports medicine is patellar tendinopathy, popularly known as "runner's knee" [125].

Anterior knee soreness is caused by patellar tendinopathy, which is characterised by pain localised to the inferior pole of the patella. Pain is exacerbated by loading and increased with the demand on the knee extensor musculature, especially during activities that involve energy storage and release in the patellar tendon. Patellar tendinopathy primarily affects relatively young (15-30 year old) athletes, particularly men, who participate in activities requiring repetitive loading of the patellar tendon, such as basketball, volleyball, athletic jump events, tennis, and football. Over forty percent of elite volleyball and basketball players have been discovered to have this disease. While several intrinsic risk variables for patellar tendinopathy, such as gender, weight, and body mass index, have been found, training load appears to be the most significant risk factor (i.e. an extrinsic risk factor).

The patella, a sesamoid bone, connects the quadriceps muscles to the inferior pole of the patella via the common quadriceps tendon (figure 82). The patellar ligament then joins the tibial tuberosity to the underside of the patella. The force created by the quadriceps muscles works as a pulley through the patellar to extend the knee. A healthy tendon consists primarily of parallel collagen fibres that are densely packed (86 percent). Type I collagen predominates. Other tendon matrix components include elastin (2%), proteoglycans (1–5%), and inorganic components (0.2 percent).

The collagen in tendons is held together by the proteoglycan components Decorin and Aggrecan, which bind to specific sites on the collagen fibrils. Tenocytes are tendon-specific fibroblast-like cells that generate collagen molecules, which aggregate into collagen fibrils. Tenocytes are closely packed between fibril bundles to create fibres. This signalling enables the cells to sense and respond to mechanical pressure by allowing them to communicate via gap junctions. Within the tendon, blood arteries run parallel to collagen fibres with some branching transverse anastomoses. It is believed that there is no nerve supply to the internal tendon, yet nerve endings and Golgi tendon organs are present at the intersection of the tendon and muscle.

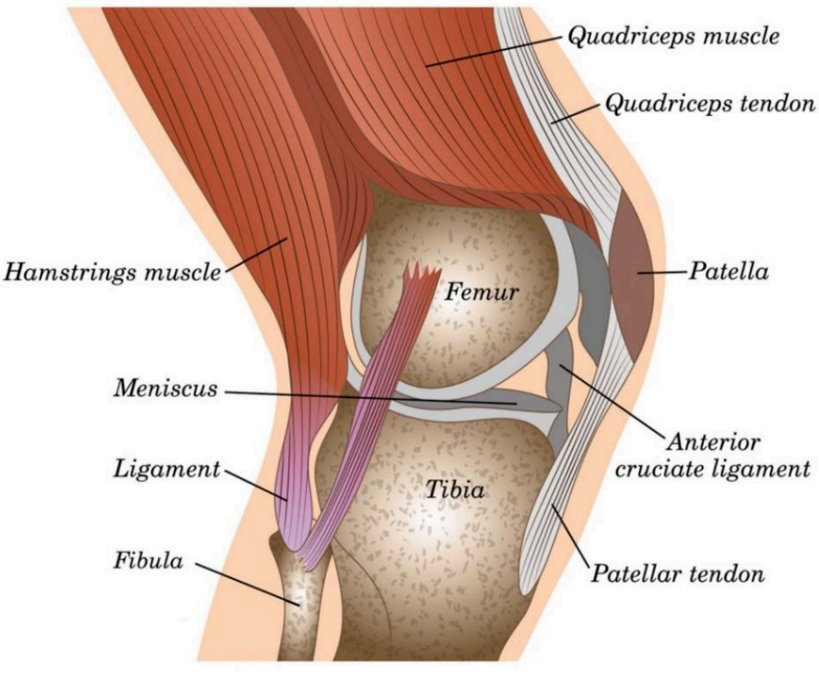


Figure 86 : Knee anatomy (Netter, F. H. 2010. Atlas of Human Anatomy)

Cook and Purdum [117] developed a new approach to tendon discomfort, which they labeled the Tendon Continuum. The continuum model offered a stage system for tendinopathy based on the variations and distribution of tendon disorganisation.

There are three stages as follows:

• Reactive tendinopathy is a non-inflammatory degenerative response in the cell and matrix that develops in response to acute tensile or compressive overload. During this stage, tenocytes expand and protein production rises. This leads in a short-term adaptation in which the tendon thickens, which reduces stress by increasing cross-sectional area or permits adaption to compression. This differs from the normal tendon response to load, which is often a hardening of the tendon. Clinically, reactive tendinopathies are caused by physical exertion that is out of the ordinary. Less frequently following a direct hit, such as landing on the patellar tendon.

• Tendon degeneration: The continuous attempt to mend the tendon after the reactive phase, but with a larger matrix degradation. There is a rise in the number of cells in the matrix, leading to an increase in protein production (proteoglycan and collagen). The increase in proteoglycans causes collagen to separate and become disorganised. It is possible for vascularity and neuronal development to rise. This stage of the pathophysiology is observed clinically in chronically overworked tendons across a range of ages and loading settings.

• Degenerative tendinopathy: regions of cell death as a result of apoptosis, trauma, or tenocyte depletion. Large portions of the matrix are disorganised and filled with capillaries, matrix breakdown products, and very little collagen. There is limited possibility of pathological changes being reversible at this stage.

Patellar tendinopathy (Figure 83) is one of several probable diagnoses for a patient with anterior knee pain. There are thought to be two distinguishing clinical characteristics: Pain confined to the inferior pole of the patella and load-related pain that worsens with increased strain on the knee extensors, especially during activities that involve energy storage and release in the patellar tendon.

The patient may experience pain with extended sitting, crouching, and stair climbing, although these are symptoms of other conditions, such as patellofemoral pain. Pain is seldom experienced during a condition of rest. Instantaneous pain accompanies loading and typically subsides when the weight is withdrawn. The discomfort may diminish with repeated loading. Tendinopathies are dose-dependent, meaning that as the magnitude or rate of load application to the tendon rises, so does the pain. In an examination, pain should increase when advancing from a shallow squat to a deeper squat or from a smaller hop height to a higher hop height. The majority of aggravating activities are loading exercises, such as running or performing squats [126].

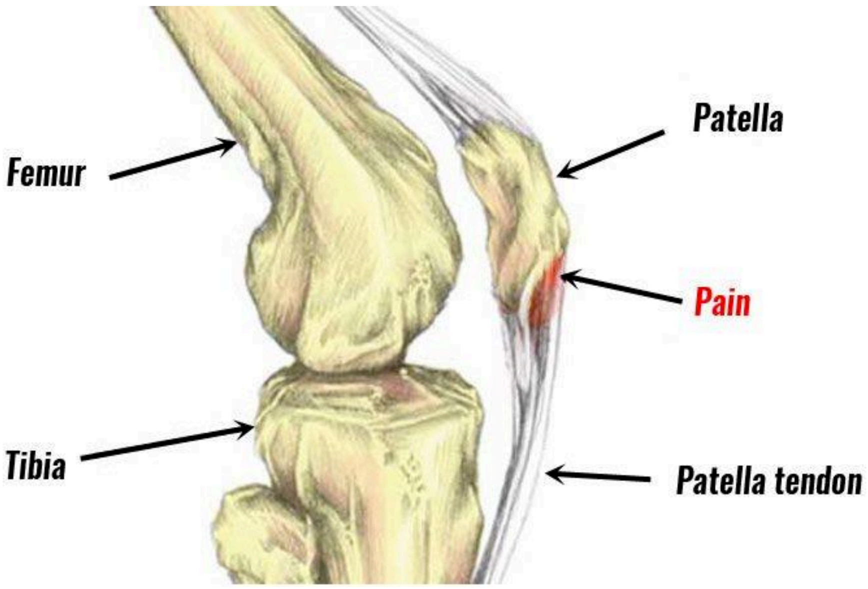


Figure 87 : Patellar tendinopathy (https://www.sportsinjuryclinic.net)

Iliotibial Band Syndrome

Iliotibial band syndrome (ITBS) is a frequent knee injury characterised by pain and/or discomfort in the lateral aspect of the knee, superior to the joint line and inferior to the lateral femoral epicondyle [127]. It is considered a non-traumatic overuse ailment, is frequently observed in runners, and is frequently accompanied by hip abductor muscle weakening. According to the prevalent theory, this syndrome is likely produced by the compression of innervated local adipose tissue. Studies have identified a "impingement zone" at or just below 30 degrees of knee flexion during foot strike and early stance phase of running. During the impingement phase of the running cycle, eccentric contraction of the tensor fascia latae and gluteus maximus muscles leads the leg to decelerate, causing tension (compression) in the iliotibial band (figure 84).

The iliotibial tract is a thick band of fascia that runs from the iliac crest to the knee on the lateral side of the thigh. It consists of dense fibrous connective tissue originating from the m. tensor fasciae latae and the medius gluteus maximus. It descends along the lateral aspect of the thigh, between the layers of the superficial fascia, and inserts at Gerdy's tubercle on the lateral tibial plateau. In its distal section, the iliotibial tract extends the lateral border of the patella and covers the lateral femoral epicondyle.

The absence of bony attachments between the Gerdy tubercle and the lateral femoral epicondyle permits the iliotibial band to migrate anteriorly and posteriorly in response to knee flexion and extension. Histologic and dissectional examinations of the iliotibial band at the lateral femoral epicondyle and gluteus maximus and fascia lata reveal a mechanosensory function operating proximally on the anterolateral knee. This mechanosensory function could influence the interpretation of the ligament versus tendon function of the ITB from the hip to the lateral femoral epicondyle.

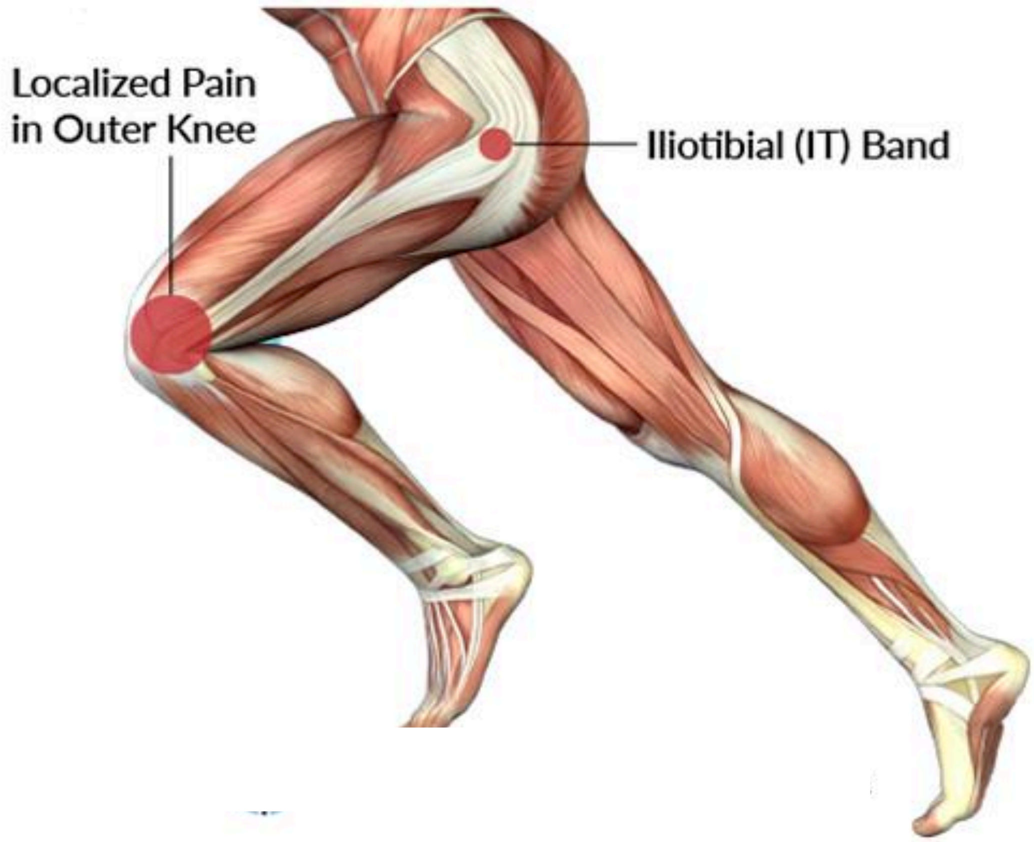


Figure 88 : Ilio tibial band syndrome (https://www.bouldermedicalcenter.com)

In many cases, the subjective evaluation will already provide a good basis for suspecting that this illness is present. A searing pain at the level of (or just underneath) the lateral femoral epicondyle is typically noted, as well as activities that demand repetitive movements involving knee flexion-extension. Different symptoms are taken into consideration while making a diagnosis in patients who have this illness.

ITBS is characterised by a strong pain on the outer aspect of the knee, which is most pronounced when the patient's heel hits the floor, and which may extend into the outer thigh or the calf. This pain is the most common symptom of ITBS. When I run or descend a flight of stairs, the pain is typically at its worst. As the knee bends, there is a possibility of hearing or feeling a cracking sensation caused by the band as it flicks over the bony tubercle.

In addition to that, there can be some swelling on the outer aspect of the knee. We notice, as one of the characteristics, a soreness over the lateral femoral epicondyle that is exacerbated by exercise. When pressure is placed to the lateral femoral picondyle of the patient's knee while the knee is in either the flexed or extended position, the patient may experience a sharp, searing pain on a frequent basis. Inflammatory markers are sometimes present as well. During running, there is discomfort on the lateral part of the knee, and this pain is made worse by running downhill or down stairs. In general, the pain is made worse when running for longer distances.

Lieutenant Commander James Renne, a medical corps officer, is credited with performing the first thorough documentation of cases of ITBS. Renne did documentation on 16 cases of ITBS out of a total of 1000 military recruits. The pain began in the lateral knee the majority of the time after a distance of two miles of jogging or after a distance of ten miles of trekking.

Walking while keeping the knee extended was effective in alleviating the discomfort. Five of the patients exhibited an atypical palpation that was described as "rubbing of a finger over a wet balloon," which was present in all of the patients. The focused tenderness was present over the lateral femoral epicondyle at 30 degrees of flexion in all of the patients. It is estimated that between 16 and 50 percent of women have ITBS, while between 50 and 81 percent of men do. The prevalence of ITBS in women ranges from 16 to 50 percent.

Piriformis syndrome

Piriformis syndrome is a common ailment among runners, in which the piriformis muscle becomes chronically tight and causes pain either locally to its position in the buttock region, or discomfort going down the sciatic nerve into the back of the leg [128]. Occasionally, this sciatic pain can also be felt in the lower leg and foot. A variety of disorders, such as lumbar disc herniations, can produce pain in the piriformis region. Consequently, piriformis syndrome is sometimes initially misdiagnosed. However, if the following symptoms are present, piriformis syndrome may be the cause of your pain or changed sensations, such as numbness or "pins and needles:" Localized pain in the buttock/piriformis region, as well as particular soreness between the sacrum and the top of the femur – the location of the piriformis muscle, buttock, and sciatica - are diagnostic of sciatica. Symptoms can be increased by positioning the hip and leg in a way that causes tension on the piriformis muscle, similar to how prolonged sitting can aggravate pain. The gluteal muscle group is situated near to the hip joint and covers the piriformis muscle (figure 84). It is frequently referred to as a deep buttock muscle. The Piriformis muscle extends from the sacrum (base of the spine) to the outside of the femoral head (thigh bone). Anatomically, its function is to assist in the rotation of your leg outward when your hip is extended and to assist in the rotation of your leg inward and into abduction when your hip is in deep flexion. The sciatic nerve is a thick nerve that begins in the lumbar spine and travels down the glute muscles, down the backs of the legs, and into the toes. There is a degree of anatomical variance in the course of the sciatic nerve across the population. In the majority of people, the sciatic nerve passes directly beneath the piriformis muscle. However, research has found that in 15% to 20% of the population, the nerve really goes through the piriformis muscle's midsection.

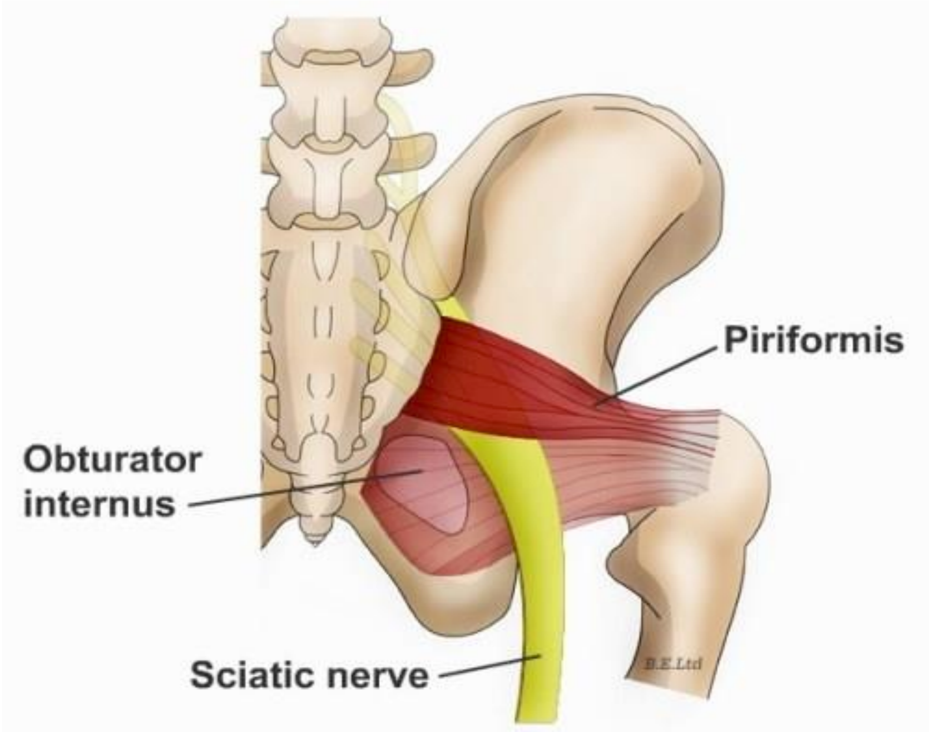


Figure 89 : Piriformis syndrome (https://www.stalbanstherapyclinic.co.uk)

Relentless stress between the sciatic nerve and the piriformis muscle of the hip is the primary cause of piriformis syndrome. This may arise for a variety of causes, such as:

• Piriformis muscular spasm, possibly caused by inflammation of the piriformis muscle or a nearby structure such as the sacroiliac joint. Typically, this is one of the body's defense processes.

• Tightening (and maybe hypertrophy) of the piriformis muscle as a result of increased physical demands on the muscle over time.

In the majority of runners, the piriformis muscle has tightened as a result of the increased pressure placed on it by weak glutes and core muscles. As we run, the gluteal muscles and core muscles should offer stability to the hip, pelvic, and lumbar spine regions. However, runners with weak glutes and core muscles may experience piriformis tightening in an attempt to supply the hip stability that would usually be provided by the gluteal muscles. The traditional "hip drop" or Trendelenburg sign is a prominent biomechanical indicator that a runner is weak through their glutes and/or core from a running technique perspective.

3. Discussion

Running is the favoured form of physical activity for tens of millions of people all over the world, and runners' range in age from teenagers to retirees. The ease with which it can be implemented is a significant contributor to its widespread acceptability. Running, on the other hand, is associated with an increased risk of musculoskeletal injuries, and it is essential to have an understanding of the factors that lead to these injuries in order to successfully prevent them [112]. Shock waves are generated whenever a person's foot makes contact with the ground [96]. One of the primary functions of the human musculoskeletal system is to dampen and spread out these shock waves. The great majority of actions, such as walking and running, are responsible for the generation of these shock waves. When walking, periods of double support during the stance phase of the gait cycle (both feet are simultaneously in contact with the ground) give way to two periods of double float at the beginning and end of the swing phase of gait (neither foot is touching the ground) [46]. This is what differentiates walking from running. [When moving at a faster speed, the point of initial contact will often move forward from the hindfoot to the forefoot.

When you run, your foot will strike the ground repeatedly when you are only using one leg. A sudden peak in the ground reaction force (impact force), rapid deceleration of the lower extremity (impact shock), and the transmission of an acceleration and deceleration wave (impact shock wave) through the body are the distinguishing characteristics of this type of impact [80].

During landings, the body is subjected to an impact shock wave, which must be attenuated by a number of different structures and systems within the body. These structures and systems include bone, synovial fluids, cartilage, soft tissues, joint kinematics, and muscular activity. A passive attenuation of shock is provided by the connective tissues and the bone. Active shock attenuation is achieved through eccentric muscle activity [80], and it is considered that this active mechanism is far more important in the process of shock attenuation than the passive mechanism. It has been hypothesised that, because muscles are thought to play a substantial part in energy and shock absorption during landing, decreased muscular activity as a result of weariness lowers the body's shock absorption capacity and, as a result, increases the risk of injury [97]. According to one definition [59], fatigue can be defined as "any loss in the total neuromuscular system's capacity to generate force," independent of the amount of force that is required to complete a specific task.

It has been found that repetitive impact loads are linked to degenerative joint disorders and athletic overuse injuries, such as stress fractures, shin splints, osteoarthritis, and lower back discomfort. There is extensive evidence linking impact, fatigue, and injury [96–98], despite the fact that the specific mechanisms of impact-related harm are only partially understood and are the subject of much debate.

Our first study (appendix A1) was designed to determine whether or not there is a correlation between the agreement between a single inertial sensor and a standard method for measuring running gait, more specifically foot strike pattern, as well as whether or not this agreement varies with increasing velocity and whether or not the runner is wearing shoes. Specifically, we were interested in whether or not there is a correlation between these two factors. A strong agreement was found between the two methods, as evidenced by the fact that 80.6% of the data were in accordance with one another and that the correlations ranged from extremely high to almost perfect. The positioning and attachment of the MEMS gyroscope can be to blame for the 19.4 percent margin of error. Evidence suggests that accelerometers are sensitive to the site and method of attachment, and studies of faster velocities have found step identification error of up to 7 percent. Despite the fact that biomechanical testing has confirmed the validity and dependability of accelerometers for measuring accelerations within the frequency and amplitude range of human body motion, evidence suggests that accelerometers are sensitive to these factors. This research found a strong correlation between the classification of foot striking patterns and the use of a single inertial sensor that was attached to the tibia and a treadmill that was outfitted with other sensors. It is possible that the error margin can be reduced by using numerous MEMS gyroscopes (one each in the foot, shin, and sacrum).

Our second study (appendix A2) aimed to determine the agreement between a single inertial sensor and a widely accepted method for measuring running gait, more specifically foot strike pattern, as well as whether or not this agreement changes with increasing velocity and whether or not the subject is wearing shoes or going barefoot when using the harmonic components as gait features. Running is an activity that occurs on a regular basis. As a consequence of this, using harmonic components as gait characteristics appears to be the most logical choice. The Fourier decomposition of the time series describing the gait features is used as the foundation for the process of obtaining the fundamental and higher-order harmonics. The magnitude measured at the fundamental frequency is a measure of the total change undergone by the associated feature, whereas the relative phase between different time series indicates the temporal delay between the various features. This can be seen by comparing the relative phase between different time series. The higher harmonics of a feature are evaluated in relation to the fundamental harmonic in order to provide a description of the nonsinusoidal but nonetheless periodic trajectory of a feature.

Although the fundamental harmonic components are responsible for the majority of the information, they are not responsible for the minute variations that occur in the dynamics of other aspects. In order to adequately represent these distinctions, higher harmonics are required. A translation-independent description of a signal consisting of solely the first and second harmonics can be intuitively offered by the amplitude of the fundamental frequency, the magnitude of the second harmonic, and the phase of the second harmonic with respect to the fundamental frequency. Since the sampling rate and amount of noise in gait analysis renders them unstable, we do not look at components that are higher than the second harmonic.

The shod running gait analysis (C8 and C11) and the barefoot running gait analysis (C2 and C3) each have their own unique characteristics that set them apart from one another (B8 and B11). An obvious rise can be seen in the shod analysis of the second harmonic. At a speed of 3.3 metres per second, the energy is likewise substantially more important. The purpose of this study was to investigate whether or not the level of agreement between a single inertial sensor and an established method for measuring running gait, specifically foot strike pattern, varies with increasing velocity and whether or not the runner is wearing shoes. Specifically, the researchers were interested in determining whether or not the level of agreement varies depending on whether or not the runner was wearing shoes.

Throughout the course of the proposed experimental procedure, shock events may take place. In our third and fourth papers (appendices A3 and A4), we describe a new way for analysing these shock events. The Shock Response Spectrum (SRS) [99] is the basis for our method. The SRS is a frequency-based function that represents the magnitude of vibration induced by a shock or other transient event [100]. The primary purpose of this study is to investigate the ability of the human musculoskeletal system to dampen the mechanical stresses brought on by the fatigue effect. This will be done by measuring the Shock Responses Spectrum (SRS) of the foot strike-generated shock waves while the subject is jogging. The majority of previous research has concentrated on shocks and impacts, ground force reactivity, spectral or vertical impact load rate, or any combination of these three. Analysis of running gait has not yet been done utilising SRS as a measurement tool; this has not yet been researched. This innovative approach has the potential to pave the way for an altogether new method of analysing the gait pattern of a runner by making use of smart linked shoes.

These studies were conducted in order to determine whether or not the SRS could be used as an effective tool to measure fatigue, and if it could, how the SRS measurement could be correlated with the effect that fatigue has on impact shock wave attenuation and how human biomechanics are related to shock attenuation while running [99-104]. It was hypothesised that fatigue would lower the shock-absorbing capacity of the musculoskeletal system, which would consequently increase the risk of experiencing an injury caused by overuse.

The primary purpose of the third study (appendix A3) was to investigate the effects of SRS-induced fatigue on the ability of the human musculoskeletal system to attenuate the shock waves that are caused by foot impacts. The capacity of the human musculoskeletal system to protect itself from overloading caused by the shock waves generated by foot strikes is reduced when the individual is fatigued. When measured on the tibial tuberosity, a rise in the amplitude of shock waves may be an indication that the individual has lost some of their protection. This strategy was successful in producing the outcomes while the participants ran on a treadmill with a motor. The collecting of data is made easier by such a system, but the patterns of locomotion that are produced may be different from those produced by running over ground. Runners were asked to maintain a consistent speed throughout the course of this inquiry, despite the fact that they were working to their physical limits.

Due to the fact that one of the key goals of the SRS study was to establish the association between running weariness and lower limb injuries, it is imperative that it be emphasised that this may slightly differ for overground running: When they feel themselves beginning to tyre, runners might consider reducing their pace as a precautionary strategy. In the event that the outcome is not a continuation of the condition of fatigue, the acceleration data might not show any further increase. This analysis came to the conclusion that in order for acceleration data to grow, there must first be a state of fatigue. As a consequence of this, the findings of this study can be extrapolated to apply, if applicable, to jogging on the ground even if fatigue is present. In addition, the majority of running injuries that occur over ground affect the lower extremities, with knee injuries being the most common. This extrapolation is supported to a large extent by the findings that we obtained, which indicated an increase in acceleration data in the tibial tuberosity. Previous studies have shown that the loading rate of the lower limb is directly and strongly connected with the running velocity, and that the vertical impact force increases as the running velocity increases. In addition, the loading rate of the lower limb is directly and strongly connected with the running velocity. When muscles are significantly fatigued, the musculoskeletal system becomes less capable of withstanding shock waves created by foot strikes. This is a conclusion that can be drawn from the available evidence. By gaining an understanding of the influence that SRS has on fatigue and the volume of dynamic stress placed on the human musculoskeletal system, it will be possible to design appropriate training protocols and exercises, hence reducing the amount of damage caused to the musculoskeletal tissues.

For there to be an increase in acceleration data, there must first be a presence of weariness, as demonstrated by the findings of our earlier research (appendix A4). When muscles are significantly fatigued, the musculoskeletal system becomes less capable of withstanding shock waves created by foot strikes. This is a conclusion that can be drawn from the available evidence. [124-125] Research has shown that running is one of the most common activities that can lead to stress fractures, often known as SF. Bones have inherent microcracks, the quantity of which increases with repeated loading, which leads researchers to conclude that these fractures are associated to bone fatigue.

Bone remodelling is the process of repairing fatigue-induced microcracks in bone. When a bone is continuously stressed, leading to repetitive or cyclic strain, it is believed that the subsequent accumulation of microdamage is the threshold of a pathological continuum that manifests clinically as stress responses and SF. This is because it is believed that the accumulation of microdamage is what causes a bone to break down into a pathological state. In the event that the activity is not halted, and the bone is unable to heal itself, a complete bone fracture may result. Notably, the number of loading cycles a bone is able to withstand before it gives in to exhaustion and breaks down diminishes when the strain or strain rate increases [131]. Stress fractures are a clinical manifestation of the gradual accumulation of fatigue damage in bone tissue [132-133-134]. Although it is common knowledge that running has an effect on bone structures due to the mechanical pressure it places on those tissues, the evidence supporting the aetiology of SF [135-135-137] is not as compelling. Despite this, a number of researchers have demonstrated unequivocal links between bone stress injuries and exhaustion. For instance, it is well known that the contraction of neighbouring muscles helps to reduce the amount of tensile strain that is placed on the tensile side of a bending bone [138-139]. This is done in order to protect the bone from stress-related injury. Therefore, it is conceivable that muscles serve as shock absorbers, and that fatigued muscles lose some of their ability to absorb shock, leading to an increase in the loading rate or loading peak experienced by the bones [140, 141, 142]. This line of thought can be supported by the observation that muscle fatigue can have this effect.

According to the results, there was a direct connection between being exhausted and generating shock waves, and the group that was experiencing fatigue experienced a gradual increase in acceleration amplitude (as revealed by the SRS). When the muscles are significantly tired, the human musculoskeletal system becomes less capable of absorbing shock waves caused by single-leg strikes. This is a conclusion that may then be validated by the findings of the investigations that came before it. The presence of this condition is thought to play a role in the onset of injuries, and the findings presented here have significant repercussions for the study of the aetiology of running injuries. Therefore, numerous recommendations to the runner community or coaches may be effective in reducing this stress-related injury risk, particularly those suggested by Brukner and Khan's [105] multifactoral model. Runners and coaches should focus on minimising the negative effects of stress on their bodies.

To begin, it could be useful to make certain that the bulk of one's training and exercise is carried out so as to avoid reaching an extreme state of fatigue and in accordance with the concept of managing one's load. Extrinsic factors such, for instance, progressive load progression [143], training surfaces, and footwear adaptations [105], need to be taken into consideration. Understanding the effect of SRS on fatigue and the degree of dynamic loading on the human musculoskeletal system would allow for the establishment of appropriate training protocols and may contribute to the reduction of damages to the musculoskeletal tissues. The ability of a runner's lower limb muscles to maintain their strength in the face of fatigue is an essential component in the fight against stress-related injuries. This is completely consistent with the objectives of the current research project. A fatigue-related imbalance between the plantar flexors and dorsiflexors may hinder the protective activity of these muscles on the skeletal structures of the lower leg, according to previous studies [136], which are consistent with the results of the current study, which found that the current study's findings were consistent with those of the previous studies. In this situation, it is possible that the calf muscles' ability to protect the bone from the risk of stress damage may be impaired as a result of fatigue-related degeneration of the muscles.

Last but not least, running injuries are frequently brought on by fatigue as well as improper technique, both of which are reflected in the runner's kinematics [144, 145]. Several researchers have proposed a gait retraining technique that either involves a slight increase in step rate or a transition from a strike on the rearfoot to a strike on the forefoot. This technique has been deemed effective for reducing impact forces and vertical load rate, and as a result, it has been shown to prevent running-related bone stress injuries [146]. Even in this day and age, a customised approach is still very much in demand and is unquestionably attainable. The vast majority of athletes are unable to get their hands on optical motion capture systems, which are used in modern research on kinematics in sports. Recent developments in microelectromechanical systems (MEMS) have led to an increase in the use of inertial sensors in the research of wearable running gait analysis [147]. This is due to a number of factors, including the fact that inertial sensors are relatively inexpensive and simple to use. Inertial sensors have the potential to be utilised in the solution to the issue of gait recognition, in which the evaluated gait can be regarded as a biometric signature. This is due to the fact that every person possesses a unique walking or jogging style. Therefore, gait identification based on inertial sensors has the potential to play a key role in a wide variety of applications that are relevant to health. We illustrated the possibilities of wearable technology for the evaluation of kinematic properties by using the example of running as our primary example. We came to the conclusion that wearable technology presents a wide variety of athletes with opportunity to improve their techniques and reduce their chances of becoming injured. Inertial sensors are now being included into smart devices, which are present at every step. As a result, inertial sensor-based gait recognition has become a very intriguing and emerging field of study that will yield a great deal of information that is really exciting.

In spite of the fact that we are unable to generalise our findings to the entire population due to the small sample size, this research is qualitative and prospective, and it investigates a topic that has not been researched before. Since the use of SRS as a measurement in running gait analysis has never been addressed before, we do not have sufficient data that is equivalent to the design of our study to generate a standard sample size. In spite of this, the results of this study offer insightful knowledge on the application of SRS as a biomechanical risk factor in runners. As more people participate in research on this subject at a greater scale, we will have a better understanding of the injury risk that runners face. It is common knowledge that the limited number of participants in this study makes it difficult to draw definitive conclusions; this is especially true if the findings of this study contradict those of earlier studies. The findings of our investigation were in line with findings from previous research as well as with our hypothesis. In subsequent studies, we will collect data from a larger pool of athletes in order to validate the results of our earlier investigations. In addition to this, studies on elite runners will be conducted similarly, and the findings will be compared. Our most obvious assumption is that runners who are at the very top of their game will be able to suppress SRS and/or keep a considerably higher SRS threshold for longer periods of time.

The findings of these trials are positive, and they demonstrate the one-of-a-kind capability of wearable devices to evaluate real-time running economy and foot mechanics out in the field. Inertial sensors that are worn on the feet can create a significant quantity of data with each stride, which can provide athletes with a number of gait variables. It is possible that this possibility will make it possible for athletes, researchers, and coaches to determine the biomechanical factors that influence the energetic cost of running. This will minimise the risk of injury or excessive fatigue while simultaneously providing the athlete with real-time feedback on performance. Additionally, the combination of these new improvements in performance with artificial intelligence (AI) and machine learning methodologies ought to make it possible to forecast gait patterns during running as well as the danger of injury. To comprehend individual differences in gait biomechanics, it is important to note that the algorithms that underpin the machine learning technique should forecast the subject-specific gait patterns rather than making group-based comparisons. The new technique that was taken here suggests that the accelerometer-based SRS model that was utilised may be able to predict injuries connected to running. The current inquiry is a preliminary application of unique methodologies that serves as a proof-of-concept demonstration. Even though we found some encouraging results, there are still a lot of methodological tweaks that might be made to make the approaches taken in this study even more effective. These encouraging first results are evidence that additional research in this area is required and should be carried out. The methodologies now being used should serve as the foundation for subsequent research, which should then collect data from bigger and more diverse samples of the population. The determination of thresholds for running-related injuries may result from the calculation of crucial time periods over which load builds to induce damage and/or the calculation of rolling probabilities of injury based on loading. Both of these calculations can be done. These results would assist in the refinement of injury prediction models and give the evidence necessary for the development of adaptive feedback and training prescriptions that take into consideration the mechanics of individual runners.

4. General conclusion and perspective

The past ten years have seen explosive growth in the research and development of wearable technologies. Now, wearable technology encompasses a huge variety of subcategories, many of which have seen great development in investment over the past few years. The size, effectiveness, and affordability of wearable technological aids for athletes are all on the rise. Businesses that specialise in sports technology have been pushed to innovate and produce "invisible" solutions as a result of the need to collect data outside of the lab rather than inside of it [1,2]. As efforts are made to incorporate wearable technology into the day-to-day activities of professional athletes, it would appear that a new era of devices is on its way to becoming a reality. In addition to becoming more compact, technology will eventually be incorporated into articles of clothing, footwear, and safety gear. Because of this, sports scientists will be able to improve not just the number and quality of data collected, but also the level of participation from athletes. This method of collecting quantitative data offers the genuine value of an athlete's performance without asking the athlete to provide any kind of subjective evaluation on their own performance. As a result of the demand for data from coaches and medical professionals in recent years, various training and competitive arenas, including the track field, have adopted the use of wearable technology. On the track, movement velocity is routinely measured so that coaches can modify training based on daily athlete preparedness rather than relying on outdated, prescriptive methods. This is done so that coaches can better serve their athletes.

One of the most important facets of the discipline of sports science is known as sports biomechanics. This area of study involves doing in-depth research on the fundamental actions performed in sports, with the dual objectives of reducing the likelihood of athletes suffering injuries and improving their overall performance. Researchers examining the physical components of performance often use sports biomechanics analysis in an effort to better understand the connection between kinetics and physical performance. When it comes to performance, biomechanical analysis can have a considerable impact, especially when it comes to the creation of models for optimal mobility and load analysis. This is done to evaluate not only the safest or most effective method to move or perform a skill, but also the influence that a changed environment has on the way the body moves or reacts. Specifically, this is used to evaluate the most efficient way to move or perform a skill. Because the legs play such an important role in movement, lower limb biomechanics is frequently the focus of attention in the field of sport biomechanics. In recent years, scientists have become increasingly interested in gaining a better understanding of weariness and the consequences it has on the human body. For the purpose of determining the forces that operate upon the body and the ways in which their kinetics and kinematics interact with one another, researchers use biomechanical evaluation equipment such as force plates and inertial sensors. Muscle mechanics are an important factor to consider while evaluating athletes for their dynamic motions. A person's injury risk and the amount of time it takes them to recuperate can be estimated based on how their muscles react to different forces. The granular metrics of displacement, velocity, and rotational speed can all be measured with the help of inertial sensors. These evaluations instruct athletes on the appropriate techniques for load-bearing, which contribute to the maintenance of their overall health.

In the last decade, the area of sports science known as sports movement analysis has been an increasingly attractive area of study. Since the beginning of time, coaches have built their careers on the ability to subjectively evaluate movement in order to enhance a talent or technique. However, up until very recently, formal and objective sport movement analysis was only possible in the lab. The level of precision that can be achieved with high-speed cameras, such as those produced by VICON motion systems, is quite astonishing. However, the bulk of professional sports teams are unable to afford such technology due to their typically excessively high cost. Cost-effective alternatives, such as the functional movement screen (FMS), have become increasingly popular in elite sports as a result of time and financial restrictions. The topic of sport movement analysis has practitioners as well as academics enthralled as they search for solutions to the question of how meaningful data may increase athlete performance or reduce injury risk. Inertial measurement units, sometimes known as IMUs, were developed as a means of attempting to address these concerns. These internal components, when linked to the athlete, acquire accurate data that is then communicated to the dashboard of the device where it is displayed. These data can then be utilised to generate a statistical profile of the athlete's kinetic performance by assessing the athlete's bone load, acceleration, step intensity, and other performance-related variables. IMUs have the ability to monitor kinetics, which can provide a biomechanical picture of the athlete. Kinetics is the study of forces involved with motion. The athlete and others who have suffered injuries comparable to theirs can benefit from having access to this data, as it can be used to inform recovery routines. The highly technical sports like tennis, football, and basketball all require optimal movement throughout the entirety of training and competition. Fatigue and strain have a negative impact not only on performance but also on health. The majority of the same motions that are used in endurance or high-velocity sports are typically performed hundreds of times when training. Therefore, proper mechanics are absolutely necessary for an athlete to have in order to maximise their performance and avoid difficulties that are related to tiredness. Every endurance athlete have a unique physical make-up that is made of strengths, weaknesses, and asymmetries that impact the movement style that is most natural to them. It is not unusual for athletes to differ in their biomechanics; nonetheless, coaches should be able to evaluate an athlete's biomechanics and suggest drills and adjustments that will assist a player in becoming more proficient. This is an area in which technological advancement is changing how sports movement analysis is performed. Movement analysis in sports is essential for lowering the risk of injury and improving overall performance.

Accelerometers are ubiquitous. They can be found in our mobile phones and laptops, as well as our autos and wearable technology. Accelerometers are capable of detecting changes in acceleration at a frequency of up to one thousand hertz, which provides us with one thousand data points per second. Accelerometers are supported by a number of other devices within the system due to the specialised requirements of the majority of our wearable technology [1,120]. Accelerometers would not be able to deliver critical information to coaches and practitioners if they did not also have gyroscopes. Gyroscopes determine orientation by making use of the gravitational pull of the earth. Magnetometers can perform the functions of compasses in addition to assisting accelerometers with determining direction and orientation. These three sensors are frequently referred to as an inertial measurement unit when they are discussed combined. As a result of the proliferation of global positioning systems (GPS) over the course of the previous decade, accelerometers are commonly disregarded. The sample rate of modern GPS systems is just 15-20Hz, which is a fraction of the sample rate of the accelerometer. This allows the modern GPS system to accurately record metrics such as total distance. When practitioners become more experienced in the use of such systems, they frequently start to notice the flaws in this kind of data and turn to the accelerometer for more profound insights as they advance in their use of the system. Because of this, practitioners are forced to look more closely at accelerations, decelerations, and the peculiarities of their stride. The evaluation of an athlete's stride characteristics is becoming increasingly common as practitioners adopt "invisible monitoring" to detect athlete tiredness while the athlete is competing. Collisions can also be detected by the accelerometer, which can be beneficial in sports such as American football and rugby, in which coaches aim to periodize the amount of direct contact their athletes face each week. In recent years, concussions have received a significant amount of attention in the news, which has led to the retirement of a number of well-known athletes. In an effort to measure the impact of collisions, a number of governing bodies and associations are looking to businesses that specialise in sports technology to integrate accelerometers into protective equipment. It is hoped that this would result in sports scientists and performance coaches having a better understanding of the demands placed on their athletes, which will allow for the protection of those athletes to be ensured.

Given that the vast majority of runners complete their workouts outdoors, the utilisation of inertial measurement units (IMUs) to analyse running biomechanical patterns in the natural environment of the runner is an amazing idea. On the other hand, guidelines for the application of IMUs in longitudinal running monitoring have not yet been formulated. When recording running biomechanics in natural environments and over the course of multiple training runs, one of the challenges that arises is the ability to determine when observed changes in running patterns are due to intrinsic factors (such as fatigue or the development of an injury) as opposed to external factors that are a part of a runner's natural environment. Despite the limited research that has been conducted in this area, it has been hypothesised that only "a few days" worth of data need to be gathered in order to establish a gait pattern that is "fairly stable." On the other hand, neither the accurate measurement of stability over a period of many days nor the number of runs necessary to achieve stability have been established. Additionally, the stability of the running pattern may be affected by the individual variables or combination of variables that are utilised to express the running pattern. According to the findings of this study, distinct running patterns for individual runners can be identified by using individual biomechanical variables or combinations of biomechanical factors.

Runners are prone to injury for a number of reasons, the most common of which are fatigue and improper technique, both of which are represented in the runner's kinematics. Research on sports kinetics and kinematics conducted in the modern day makes use of motion analysis technologies that are out of reach for the majority of participants. Wearable sensors have a great deal of potential in terms of both their applicability and their cost-effectiveness when it comes to analysing the kinetics and kinematics of runners. The results of our study show that wearable sensors have the potential to be used for analysing the kinetics and kinematics of runners. Several studies that used inertial measurement units (IMU) as a performance evaluation tool, a training aid, or a fatigue monitoring device are discussed in this article. We retrieved a variety of gait measures in order to conduct health and performance assessments. Wearable sensors are a useful tool for monitoring running technique, and they can be used by runners of varying levels of expertise.

Even in this day and age, a personalised approach is still very much in demand and is unquestionably attainable. The vast majority of athletes are unable to get their hands on optical motion capture systems, which are used in modern research on kinematics in sports. Recent developments in microelectromechanical systems (MEMS) have resulted in the widespread usage of inertial sensors in the research of wearable running gait analysis [40]. This is due to a variety of factors, including the fact that inertial sensors are very inexpensive and simple to use. Inertial sensors have the potential to be utilised in the solution to the issue of gait recognition, in which the evaluated gait can be regarded as a biometric signature. This is due to the fact that every person possesses a unique walking or jogging style. Therefore, gait identification based on inertial sensors has the potential to play a key role in a wide variety of applications that are relevant to health. We illustrated the possibilities of wearable technology for the evaluation of kinematic properties by using the example of running as our primary example. We came to the conclusion that wearable technology presents a wide variety of athletes with opportunity to improve their techniques and reduce their chances of becoming injured. Inertial sensors are now being included into smart devices, which are present at every step. As a result, inertial sensor-based gait recognition has become a very intriguing and emerging field of study that will yield a great deal of information that is really exciting.

The running gait of healthy and injured individuals has traditionally been evaluated in the context of a laboratory setting; however, the findings of these studies may not be applicable to actual running. The vast majority of runners get their workouts outside and almost never run in a controlled environment or while being watched closely. Laboratory studies often only examine 5–15 steps of a subject's run due to constraints in data storage and analysis, but the average human takes roughly 2,000 steps each mile. This is because to the limitations in data storage and processing. In order to account for the fact that there is a chance that the mechanics of running will vary over longer distances, it may be beneficial to examine a large number of steps. Both medical professionals who work with runners and researchers who study a variety of runners can benefit from the fact that gait biomechanics can be measured by wearable sensors [148]. Wearable sensors have the potential to offer continuous measurements of gait mechanics spanning thousands of steps during the average training run of a distance runner, hence enabling more precise monitoring of many facets of gait biomechanics.

MEMS accelerometers are the sort of sensor that has been the most widely utilised and authorised for detecting physical activity outside of traditional laboratories. Despite this, running biomechanics have received relatively little attention through the use of wearable sensors. The test-retest reliability of the accelerometers employed in gait assessment was found to be good to outstanding when compared to the gold standard motion analysis system (interclass correlation coefficient (ICC) = 0.70–0.97).

The RunScribe, which was developed by Scribe Labs, Inc. and is based in Half Moon Bay, California, in the United States, is a small, wearable gadget that can be mounted to a runner's shoe in order to analyse the biomechanics of their running gait. This wearable device, in comparison to standard pedometers and accelerometers, records kinematic, kinetic, and spatiotemporal data for each individual step. This represents a substantial advancement in the technology that underpins the device. Because it is attached to the runner's shoe in such a way that it cannot be felt, the sensor provides another advantage. In contrast to the typical laboratory arrangement, in which markers or sensors are attached to each participant, these subjects are free to run around and experiment as they like thanks to this location.

To meet the growing demand for healthcare services that are becoming progressively more expensive due to the skyrocketing costs of in-lab diagnostic equipment, there is a growing desire for wearable healthcare solutions that are low-cost, unobtrusive, and reliable. This is because there is a growing demand for healthcare services. In recent years, significant strides have been made in the creation of diminutive sensors and actuators, as well as in wireless communication, computer, and information technologies. These developments have permitted the development of intelligent healthcare devices that are also affordable. Because of the rapid growth of micro-electromechanical systems (MEMS) technology, it is now possible to create microscopic sensors and actuators with outstanding sensitivity that are inexpensive and do not take up much space. Wearable technologies that incorporate pressure sensors have the potential to greatly expand the range of applications for health monitoring and to improve interactions between humans and technology. In addition, inertial measurement devices based on MEMS, such as accelerometers and gyroscopes, are increasingly being incorporated into a wide range of applications, such as fitness trackers, mobile phones, and automobiles. In point of fact, it is anticipated that in the era of internet-of-things (IoT) and internet-of-everything (IoE), such small actuators and sensors would play a vital role in the functioning of smart systems and smart devices. Therefore, wearable pressure sensors and inertial measurement units (IMUs), when combined with the high-speed communication and computing technologies of modern smart devices, have the potential to be utilised for real-time monitoring of plantar pressure, activity, and running gait pattern, as well as for quantitatively assessing the athletic endeavours of individuals.

Running gait analysis will undoubtedly progress toward using intelligent footwear in the near future [149]. The footwear of the future is just now starting to become accessible on the market after years of research and development. The footwear of the next generation is a product that is more difficult to conceive of, create, and manufacture than other wearable technologies already available. Smart shoes will be able to broadcast critical variables to your smartphone. In order to do so, they will need to be equipped with battery-powered pressure sensors, MEMS gyroscopes and accelerometers, and application connection, all while maintaining their comfort and lightweight design. The following are some of the technology components that are generally included in current linked shoes that are available on the market:

- a gyroscope that calculates the direction your shoe is rotated, as well as rotation as your foot moves while running

- an accelerometer that measures movement and velocity (in several dimensions, depending on capability)

- optional pressure sensors to assess impact and weight distribution, which might assist detect potential injury risks in your running style.

- GPS (optional) for location and route tracking

- altimeter (optional) to track height changes, which is vital if you do trail running or even if you merely frequently run on steep terrain

- a CPU that can inform you how to enhance your running style by calculating movement patterns from sensor data

- connection, typically Bluetooth, for data transmission to your smartphone.

In many respects, we now have shoes that are similar to those that will be worn in the future. Wearable technology has the ability to monitor, gather, and analyse data, which can improve athletic performance as well as the quality of life for those who are physically challenged. On the other hand, when smart insoles and shoes become more widely used and popular, their capabilities will definitely advance alongside this trend. A knowledgeable assistant will provide you with coaching comments in real time and direct you as you complete your run. Data collected from jogging could one day be used to improve the performance of a pair of running shoes that have been 3D printed.

It is possible that in the future, as technology and materials continue to advance, it will be possible to automatically and dynamically adapt the sole and cushioning to running situations. For example, increasing traction when running on unstable terrain or softening the insole when running on asphalt could become possible. The application of monitoring technology to all types of footwear and the incorporation of the data obtained from that tracking into other types of health and fitness data could constitute a more comprehensive approach. You will be better able to identify patterns and connections between your day-to-day behaviours and your athletic performance, general fitness, and specific challenges, as well as perhaps your physical, mental, and overall health and well-being if you do this. It is hoped that intelligent running shoes will improve running economy; there is a correlation between cadence and running economy (i.e., efficiency).

As a result, it is in the best interest of every runner to occasionally alter their cadence in order to assess whether or not doing so makes running easier. However, it has never been easy to determine how to get started with anything. Connected footwear makes the process easier by providing a personalised selection of options to choose from. Even if the selection might not be perfect for everyone, it is still a good place to begin looking.

Because your shoe is providing you with this information, you are able to make adjustments to both your stride and your cadence. The coaching feedback provided by the application details how the adjustments you've made have contributed to an improvement in your technique over long distance runs and intervals.

A fascinating function of smart shoes is the ability to determine the angle at which your foot strikes the ground [150]. In addition to making suggestions to increase your cadence, shorten your stride, or take additional rest days before an interval session, smart shoes will have the ability to measure the angle at which your foot strikes the ground. [Citation needed]

New sensors can tell you exactly what type of running shoe is best for you by analysing your weight, height, and weekly mileage in addition to your foot strike data. They can also tell you when you may need a new pair of running shoes by tracking your mileage to determine whether or not the shoe has exceeded its intended lifespan. This information can be combined with foot strike data.

In conclusion, smart connected shoes will also help identify fatigue without having to rely on satellite for GPS metrics like pace and distance, which is certainly helpful in situations where your signal may fail or suffer interference from other sources. However, one of the most intriguing capabilities of smart shoes is their potential capacity to help runners gauge their fatigue and injury risk, as depicted in figure 86 [151-152].

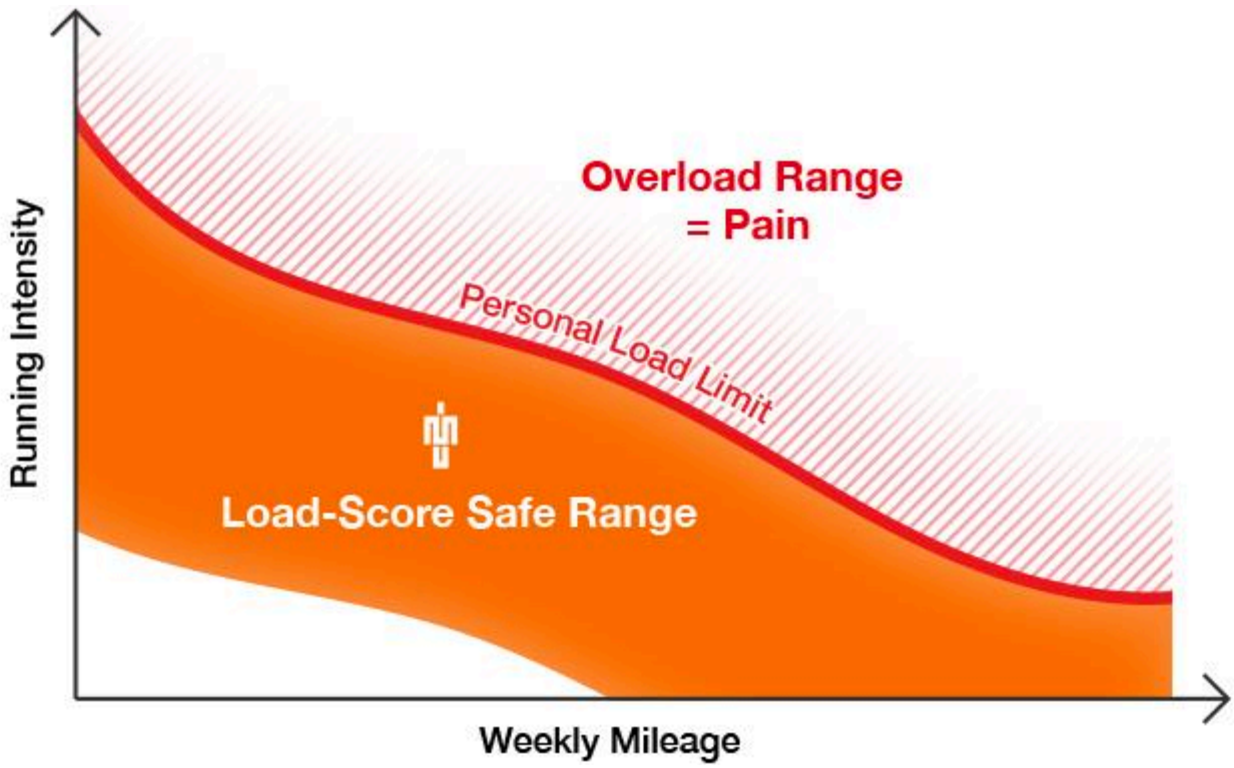


Figure 90 : Theoretical injury threshold (https://imeasureu.com)

CHAPTER 5
Résumé en français

1. Introduction

Cette thèse explore le sujet de la reconnaissance de la biomécanique humaine en course à pied et de la classification des schémas de foulées qui pourraient conduire à des blessures de surmenage. Nous avons conçu un protocole expérimental et un nouvel outil de mesure que nous avons appliqué à la biomécanique de la course. De plus, nous avons développé un algorithme interne qui nous permet de mesurer plusieurs types de caractéristiques de séquence de foulées qui peuvent être utilisées pour extraire des paramètres discriminants, telles qu'une signature biométrique de foulée pouvant amener une blessure par surmenage microtraumatique. Ces méthodes d'intégration de données et de temps sont conçues pour répondre à la question suivante : « Pourquoi me suis-je blessé aujourd'hui ? » Ces caractéristiques de foulée sont testées sur des données d'accélération/gyroscope que nous avons collectées pour simuler des scénarios de course en extérieur réalistes. L'approche portant sur les facteurs de risque de blessures liés à la course à pied visait à déterminer :

- Comment la fatigue liée à la course peut affecter la capacité du système musculosquelettique humain à atténuer les contraintes mécaniques résultant de la course

- Comment trouver une nouvelle approche et méthodologie pour mesurer correctement le risque lié aux impacts et aux chocs pendant la course

- Comment les ondes de choc associées à la foulée entraînent-t-elle des blessures et comment mesurons-nous le seuil de charge avant la blessure d'un coureur.

La course à pied est l'exercice de choix pour des millions de personnes partout dans le monde et à tous les âges. L'une des principales raisons de sa popularité tient à sa simplicité. Cependant, la course à pied comporte également un risque accru de blessures musculosquelettiques et il est nécessaire de comprendre l'étiologie des blessures afin de les prévenir efficacement.

J'exerce la podologie du sport depuis plus d'une décennie à Paris à partir d’examens cliniques, d’analyses quantifiées de la marche/foulée et je fournie des orthèses plantaires sur mesure pour mes patients. Une très grande partie de mon activité professionnelle est notamment consacrée à la prise en charge des coureurs blessés. Je peux honnêtement dire que j'ai vu un très large éventail de coureurs, du coureur récréatif occasionnel à l'ultra-coureur d'élite, qui sont venus à moi dans toutes les formes, tailles et âges.

Dès le début de ma carrière, j'ai eu la chance de pouvoir investir dans des outils d'analyse du mouvement de qualité. Nos outils d'analyse vont des tapis roulants instrumentés (gold standard) aux systèmes de balayage optique léger basés sur la vidéo-raster-stéréographie (VRS), aux systèmes optométriques, aux systèmes d'analyse vidéo et jusqu'aux appareils portables connectés pour les mesures en extérieur telles que les centrales inertielles (IMU) ou systèmes micro-électromécaniques (MEMs).

L'accès à de telles technologies m'a permis de collecter de nombreuses données sur un large éventail de coureurs tout au long de ma carrière. Très vite, j'ai réalisé que ma seule formation en podologie ne suffirait pas pour que je puisse décrypter et analyser correctement ces banques de données et les exploiter à bon escient. J'utilise également les dernières technologies CAO/FAO pour fabriquer les semelles orthopédiques telles que le fraisage numérique 3D et l'impression 3D afin d'offrir la meilleure qualité de traitement aux patients. La méthode de fabrication CAO/FAO permet la reproductibilité et la répétabilité du traitement des orthèses plantaires.

L'apogée de ma thèse a été de proposer une nouvelle méthodologie pour l'analyse des impacts/choc survenant au cours de la procédure expérimentale proposée en course à pied. Notre approche est basée sur le spectre de réponse aux chocs (SRS), qui est une fonction basée sur la fréquence utilisée pour indiquer l'amplitude des vibrations dues à un choc ou à un événement transitoire.

L'objectif principal est d'analyser la capacité du système musculosquelettique humain à atténuer les contraintes mécaniques résultant de l'effet de fatigue par le spectre de réponse aux chocs (SRS) des ondes de choc générées par la pose du pied pendant la course. L'utilisation du SRS comme mesure dans l'analyse de la foulée en temps réel n'a jamais été étudiée à ce jour. La course implique des impacts et chocs répétés entre le pied et le sol. De tels impacts sont caractérisés par un pic transitoire de la force de réaction au sol (force d'impact), une décélération rapide du membre inférieur (choc d'impact) et l'initiation d'une onde d'accélération et de décélération (onde de choc) qui se propage à travers le corps. La charge mécanique produite par des impacts répétés a été liées à des maladies articulaires dégénératives et à des blessures de surmenage sportif, notamment des fractures de stress, des périostites tibiales, de l'arthrose et des douleurs lombaires. Bien que les mécanismes exacts des blessures liées aux chocs soient relativement mal connus, les preuves parfois controversées associant les impacts de la foulée et les blessures sont bien documentées.

Cette approche innovante pourrait ouvrir la voie à une toute nouvelle façon d'évaluer la foulée d'un coureur à l'aide de chaussures connectées intelligentes. Le développement des objets connectés et plus précisément des «Smart Wearables» (SW) entraînent une explosion quantitative de données amenant de nouvelles informations analyser et à interpréter. Divers experts/grandes institutions (comme le MIT aux États-Unis), administrations et spécialistes sur le terrain des considèrent le phénomène « Big Data » comme l'un des grands défis informatiques de la décennie 2020-2030 et en ont fait une de leurs nouvelles priorités de recherche et développement. En podologie la mesure de pression plantaire est utilisée aujourd'hui dans de nombreuses situations cliniques : suivi post-opératoire, design d'orthèses plantaires, rééducation à la marche, aide aux choix préopératoires, suivi du patient diabétique et évaluation de la chirurgie du pied. En parallèle elle est aussi utilisée en laboratoire pour l'analyse des phénomènes qui gouvernent la marche et la posture humaine.

De nombreux outils tels que les centrales inertielles permettent de faciliter la collecte de données et les analyses en situation réelle. Ce type de dispositif permet le suivi de patient ou de sportif et le monitoring de paramètres de la course ouvrant un vaste champ de possible dans la prévention des pathologies, l'amélioration de la performance, la rééducation des patients et la mise en place d’orthèses. L’analyse quantifiée de la foulée pathologique est un défi majeur pour la mise en place de dispositif préventif et la médecine du sport prédictive.

Au cours des deux dernières décennies, l'analyse quantifiée de la marche/foulée est passée d'une discipline purement académique à un outil de mesure précieux entre les mains des médecins et des thérapeutes. Tout au long de l'histoire, les hommes se sont intéressés aux mouvements liés à la marche. Les peintures et sculptures de la Grèce classique et de Rome démontrent que les artistes de cette époque possédaient une compréhension de la forme et de l'alignement des membres au cours de différentes activités. Au cours de la Renaissance, une telle compréhension a été avancée par la dissection humaine et par des tentatives pour comprendre les rudiments de la biomécanique, en particulier par des personnalités telles que Léonard de Vinci, Galilée, Newton et surtout Borelli. Au XIXe siècle, les premières investigations biomécaniques formelles ont été faites par les frères Weber en Allemagne. Depuis lors, les avancées dans 4 domaines scientifiques différents ont contribué au développement de l'analyse quantifiée de la marche. Ces 4 domaines sont la cinématique, la cinétique, l'électromyographie et les mathématiques appliquées à l’ingénierie mécanique. Garrison (1929), Bresler et Frankel (1950) et Steindler (1953) ont fourni des bases solides pour les rudiments de l'analyse quantifiée de la marche.

Depuis plus d'un siècle, les cliniciens en orthopédie, en physiothérapie et en podologie, tentent d'utiliser des méthodes tirées de la biomécanique humaine pour résoudre les problèmes fonctionnels du pied et de l’ensemble de l’appareil locomoteur. Dans les années 1960 et 1970, Merton Root et ses collègues podologues ont joué un rôle central en fournissant aux praticiens de la santé du pied, et aux podologues en particulier, une base cohérente pour l'évaluation et le traitement biomécanique des pathologies du pied et des membres inférieurs. Au fur et à mesure que la communauté podologique s'impliquait davantage dans la biomécanique, les chercheurs de la communauté internationale de la biomécanique ont commencé à s'intéresser davantage aux résultats cliniques de leurs études. En ce qui concerne la recherche sur la biomécanique du pied et des membres inférieurs, ce n'est que dans la première moitié du XXe siècle que la littérature médicale suggère qu'il existait une relation entre la mécanique du pied et la pathologie du pied et que la mécanique du pied humain était peut-être beaucoup plus compliquée que ce que l'on pensait auparavant.

Outre les podologues, des chercheurs en biomécanique bien connus ont également proposé des théories concernant la fonction possible du pied et des membres inférieurs. La ''Preferred Movement Pathway Theory'' de Benno Nigg (1999, 2001) propose que les orthèses plantaires ne fonctionnent pas en réalignant le squelette mais fonctionnent plutôt en modifiant les signaux d'entrée dans la plante du pied qui, à leur tour, modifient le '' réglage musculaire '' ' du pied et du membre inférieur lors d'activités sportives. L'approche des systèmes dynamiques de Joseph Hamill et ses travaux sur la biomécanique des membres inférieurs (Hamill et al. 1988, 1999, 2008) ont été fréquemment cités dans les discussions podologiques, tout comme les travaux de Peter Cavanagh sur le pied diabétique (Cavanagh 2000, 2006). Il existe d'autres exemples de chercheurs et de cliniciens en biomécanique combinant leurs compétences individuelles pour effectuer des recherches qui ont encore enrichi la base de connaissances sur la biomécanique du pied et des membres inférieurs.

Enfin, il est impossible d'ignorer l'énorme contribution que les marques de sport ont apportée au monde de la biomécanique à travers leur innovation constante dans le domaine des chaussures de course. L'un des premiers exemples de chaussures de course remonte à 1865. Trouvée dans un musée à Northampton, en Angleterre, une ville réputée pour sa fabrication de chaussures à l'époque. Un autre cordonnier de la première heure était aussi une société anglaise appelée JW Foster and Sons, maintenant mieux connue sous le nom de Reebok. Fondée en 1890 par Joseph William Foster, un coureur passionné qui voulait concevoir des chaussures qui pourraient l'aider à aller plus vite, l'entreprise fabriquait des pointes en cuir portées par des athlètes britanniques, dont le champion olympique du 100 mètres de 1924, Harold Abrahams.

Au milieu des années 1800, un nouveau procédé appelé vulcanisation a été développé. Utilisant la chaleur pour fusionner le caoutchouc et le tissu, cela a conduit à l'invention des premiers tennis, des chaussures légères en toile à semelle en caoutchouc portées spécialement pour le sport. Connues sous le nom de baskets parce que la semelle en caoutchouc permettait au porteur de se promener sans être entendu, elles sont devenues populaires auprès des athlètes après la Première Guerre mondiale lorsque des entreprises telles que Keds et Converse ont commencé à les vendre comme chaussures de sport. Dans les années 1920, les frères Adi et Rudolf Dassler ont lancé une entreprise de chaussures de sport dans la petite ville allemande d'Herzogenaurach, spécialisée dans les chaussures d'athlétisme. La politique, la guerre et leurs femmes qui ne s'entendaient pas ont provoqué une rupture entre les deux frères. Dans les années 1940, ils installent des boutiques rivales dans la même ville de part et d'autre du fleuve. Rudolf a lancé Puma en 1948. Adi a ouvert Adidas (les deux sociétés sont toujours basées aujourd’hui à Herzogenaurach).

Dans les années 1950, Bill Bowerman, entraîneur en chef d’athlétisme à l'Université de l'Oregon, voulait inventer des chaussures de course plus légères et plus rapides pour ses athlètes. La plupart portaient des pointes en cuir, semblables à celles que Roger Bannister portait lorsqu'il franchit le mile de quatre minutes en 1954. Bien que Bowerman ait apporté ses créations à plusieurs fabricants, personne n'investira. Dans les années 1960, alors que les courses de longue distance devenaient plus populaires, Bowerman et l'un de ses anciens étudiants athlètes, Phil Knight, ont lancé une société appelée Blue Ribbon Sports qui importait des chaussures de sport japonaises nommées Onitsuka, mieux connues aujourd'hui sous le nom d'ASICS. Ils ont acheté des chaussures basées sur les créations de Bowerman auprès d'entreprises japonaises de chaussures de course et les ont vendues à l'arrière de camionnettes lors de courses. Leur chaussure la plus populaire était la Cortez. Grâce à une semelle intermédiaire en caoutchouc spongieux, elle a été l'une des premières à offrir un amorti contre l'impact de la route. Lorsque Bowerman et Knight ont créé leur propre entreprise de fabrication en mai 1971, la Cortez est devenue leur chaussure phare. Cette société s'appelle Nike.

Les années 1970 ont vu l'application de la science du sport aux chaussures de course. Les podologues, désormais impliqués dans la recherche et la conception, ont identifié différentes allures de course ainsi que des chaussures adaptées à chaque type. L'une des innovations les plus durables dans la technologie des chaussures est apparue. L'acétate de vinyle éthylène, ou EVA, encore utilisé dans la plupart des chaussures aujourd'hui, est une mousse infusée d'air qui offre un amorti et absorbe les chocs. Brooks, le premier à utiliser l'EVA, l'a incorporé dans les chaussures Villanova en 1975.

Au cours de la même décennie, Bowerman de Nike a développé un nouveau type de semelle de traction plus légère pour les chaussures de piste. À la fin des années 70 et au début des années 80, la course à pied était en plein essor. Entre 1971 et 1981, il y a eu une augmentation de 1 800 % du nombre de marathoniens et les chaussures de course sont devenues plus populaires. Brooks a lancé la première chaussure pour essayer de contrôler la pronation (la rotation du pied) en 1976. La pronation étant censée causer des blessures, la Brooks Vantage avait un coin à l'intérieur pour aider le pied du coureur à s'incliner légèrement vers l'extérieur. Dans les années 1980, les chaussures conçues pour la stabilité, avec la combinaison d'un bon soutien et d'un amorti suffisant, faisaient fureur. En 1987, Nike a créé une icône en rendant sa technologie de bulle d'amorti au talon, utilisée pour la première fois en 1979, visible sur la Nike Air Max.

Au début des années 2000, un nouveau courant de pensé illustrer dans l'article dans la revue Nature du professeur Daniel E. Lieberman de Harvard, a suggéré que la course pieds nus peut être confortable, pour certaines personnes, même sur les surfaces les plus dures. Le monde de la course est devenu préoccupé par cette idée. Après avoir remarqué que les athlètes de Stanford s'entraînaient pieds nus, en 2001, Nike a commencé à travailler sur la Nike Free, une chaussure conçue pour recréer la sensation de course pieds nus en réduisant le poids de la chaussure et en utilisant une semelle qui permet aux coureurs de se sentir plus connectés au sol.

En 2006, les FiveFingers de Vibram - des chaussures minimalistes semblables à des gants avec des compartiments individuels pour chaque orteil et des semelles en caoutchouc extrêmement fines - sont sorties, et le minimalisme, l'idée que moins de chaussures, c'est plus, était le nouveau mot à la mode. Ce principe pieds nus, selon lequel les chaussures doivent fonctionner avec le mouvement du pied plutôt que contre lui, a été cimenté lorsque Christopher McDougall a publié le livre "Born to Run", après avoir vécu avec les Tarahumara, une tribu mexicaine indigène d'ultra-coureurs prolifiques qui portaient seules des sandales fines pour réaliser des exploits d'endurance extraordinaires, comme parcourir des centaines de kilomètres en une seule course. La course pieds nus n'a peut-être pas été la réponse à la prévention des blessures, mais le mouvement a laissé une empreinte. Ces dernières années, les chaussures ont continué à devenir plus légères, plus confortables et plus personnalisées.

En 2013, Adidas a présenté sa technologie Boost, un système d'amorti conçu avec le leader de la chimie BASF. L'idée était de créer quelque chose de "meilleur que l'EVA". La semelle intercalaire en polyuréthane thermoplastique, composée de milliers de capsules à retour d'énergie, fait sans doute l'affaire. Il se comprime sous pression pour absorber les chocs et rebondit instantanément pour fournir un retour d'énergie à chaque foulée.

De nos jours, la course pour concevoir la paire qui franchit la barrière du marathon de deux heures est fermement lancée. Plus tôt cette année, le documentaire de Nike "Breaking2" a relaté la tentative de la marque de passer en dessous de deux avec certains des meilleurs athlètes du monde courant dans la Nike Zoom Vaporfly Elite. Conçu spécifiquement pour l'objectif de deux heures, le Vaporfly comprend une plaque en fibre de carbone dans la semelle intermédiaire conçue pour fournir un meilleur retour d'énergie et aider à propulser les athlètes vers l'avant.

2. Anatomie fonctionnelle et biomécanique humaine

L’appareil locomoteur humain est le système qui permet aux humains de se déplacer en utilisant les systèmes musculaire et squelettique. Le système musculosquelettique fournit forme, soutien, stabilité et mouvement au corps. L'ensemble de ce système est constitué des muscles du squelette, du cartilage, des tendons, des ligaments, des articulations et d'autres tissus conjonctifs qui soutiennent et lient les tissus et les organes. Les principales fonctions du système musculosquelettique consistent à soutenir le corps, à permettre le mouvement et à protéger les organes vitaux.

Ce système décrit comment les os sont interconnectés via les articulations et comment les fibres musculaires se connectent à ces os et articulations via les tendons et les ligaments. Les os fournissent le soutien et la stabilité au corps, le mouvement des os et le mouvement sont possibles grâce à la mobilité articulaire. Les muscles maintiennent les os en place et jouent également un rôle dans le mouvement des os et des articulations, le muscle se contracte pour déplacer l'os attaché à l'articulation. Les articulations sont principalement constituées de cartilage pour empêcher les extrémités osseuses de se frotter directement les unes contre les autres, créant ainsi de l'arthrite.

Un squelette humain adulte est composé de 206 os : 22 os crâniens et faciaux, 6 os de l'oreille, 1 os de la gorge, 4 os de l'épaule, 25 os de la poitrine, 26 os vertébraux, 6 os du bras et de l'avant-bras, 54 os de la main, 2 os du bassin, 8 os de la jambe et 52 os du pied.

Le mouvement s'effectue dans 3 plans différents : le plan sagittal, le plan coronal et le plan transversal. Les mouvements sont classés d'après les plans anatomiques dans lesquels ils se produisent, bien que le mouvement soit le plus souvent une combinaison de différents mouvements se produisant simultanément dans plusieurs plans.

Les articulations sont divisées en trois classes : les synarthroses (articulation immobile entre les os, par exemple les sutures du crâne), les amphiarthroses (une articulation légèrement mobile composée de tissu conjonctif fibreux dense, par exemple la syndesmose entre les extrémités distales du tibia et du péroné) et diarthroses (une articulation librement mobile entre des os recouverts de cartilage articulaire et reliés par des ligaments bordés par une membrane synoviale, par exemple toutes les articulations synoviales).

La partie musculaire de l’appareil locomoteur est composée de trois types de muscles : cardiaque, squelettique et lisse. Les muscles lisses sont utilisés pour contrôler le flux de substances dans le corps et ne sont pas contrôlés consciemment, les muscles cardiaques se trouvent dans le cœur et font exclusivement circuler le sang. Seuls les muscles squelettiques peuvent induire le mouvement, ils sont attachés aux os et disposés en groupes opposés autour des articulations en utilisant différents types de contractions pour générer des forces permettant la démarche et la posture. La contraction musculaire volontaire est contrôlée par le système nerveux central.

Classification des contractions musculaires volontaires :

- Contraction concentrique : la force générée par le muscle est suffisante pour vaincre la force extérieure et crée le mouvement ; le muscle se raccourcit en se contractant. Par exemple : la phase de pointe de la marche est induite par la contraction concentrique du muscle triceps sural

- Contraction excentrique : la force générée est insuffisante pour vaincre la charge externe sur le muscle et les fibres musculaires s'allongent en se contractant. Une contraction excentrique est utilisée comme moyen de décélération d'une partie du corps. Par exemple : la phase de contact du talon de la marche induit une contraction excentrique du muscle triceps sural pour ralentir la phase de chargement.

- Contraction isométrique : le muscle reste de la même longueur et aucun mouvement n'en résulte. Par exemple : la théorie de Pauwels sur la biomécanique de la hanche permettant une position unipodale et le maintien de l'équilibre.

- Contraction isotonique : la tension dans le muscle reste constante malgré une modification de la longueur du muscle. Ces contractions peuvent être excentriques ou concentriques.

Les muscles peuvent être classés en fonction des fonctions qu'ils remplissent dans de tels schémas, à savoir en tant que moteurs principaux, antagonistes, muscles de fixation et synergistes. Une catégorie spéciale comprend ceux qui ont une action paradoxale ou excentrique, dans laquelle les muscles s'allongent en se contractant. Ce faisant, ils accomplissent un travail négatif. Un muscle peut être un moteur principal dans un modèle, un antagoniste dans un autre ou un synergiste dans un troisième.

La biomécanique applique les principes de la physique aux mouvements humains. Certaines articulations fonctionnent comme des leviers, d'autres comme des poulies et d'autres encore comme un mécanisme roue-essieu. La plupart des mouvements utilisent le principe des leviers. Un levier consiste en une "barre" rigide qui pivote autour d'un point d'appui fixe :

- Point d'appui ou axe : le point autour duquel tourne le levier

- Charge ou résistance : la force appliquée par le système de levier (en biomécanique c'est le plus souvent le poids du corps)

- Effort ou force : la force appliquée par l'utilisateur du système de levier

Dans le corps humain, le point d'appui est l'axe de l'articulation, les os sont les leviers, les muscles squelettiques créent généralement le mouvement et la résistance peut être le poids d'une partie du corps, le poids d'un objet sur lequel on agit, la tension d'un antagoniste le muscle. Les leviers sont classés en première, deuxième et troisième classe, selon les relations entre le point d'appui, l'effort et la résistance.

La marche est le terme utilisé pour décrire la locomotion humaine ou la façon dont nous marchons. Fait intéressant, chaque individu a un modèle de démarche unique. La démarche d'une personne peut être grandement affectée par des blessures, des maladies et des chaussures mal adaptées. La marche utilise une séquence répétitive de mouvements des membres pour déplacer simultanément le corps vers l'avant tout en maintenant la stabilité de la position. Au fur et à mesure que le corps avance, un membre sert de source mobile de support tandis que l'autre membre avance vers un nouveau site de support, puis les membres inversent leurs rôles. Pour le transfert du poids du corps d'un membre à l'autre, les deux pieds sont en contact avec le sol. Cette série d'événements est répétée par chaque membre avec une réciproque jusqu'à ce que la destination de la personne soit atteinte. C'est un mécanisme qui dépend de l'action étroitement intégrée des sujets, des os, des muscles, des articulations et du système nerveux. Les prérequis de la démarche sont : la capacité à soutenir ou à adopter une position verticale, la capacité à maintenir l'équilibre en position verticale en situation statique et dynamique et la capacité à développer ou à créer un nouveau pas en avant.

Pendant la marche, le corps se divise fonctionnellement en 2 unités : passagère et locomotrice. L'unité passagère n'est responsable que de sa propre intégrité posturale, la mécanique de la marche normale est si efficace que les exigences imposées à l'unité passagère sont réduites au strict minimum, ce qui en fait pratiquement une entité passive portée par le système locomoteur. L'alignement de l'unité passagère sur les membres est un déterminant majeur de l'action musculaire au sein du système locomoteur. La tête, les bras et le tronc sont l'unité passagère. L'unité locomotrice se compose de deux membres reliés par le bassin. Cela fait du bassin un élément à la fois du passager et des unités locomotrices, avec deux sites de jonction très mobiles, la colonne lombaire et les articulations de la hanche. La structure complexe du système locomoteur humain permet la locomotion bipède et offre une stabilité et une démarche stable.

La démarcation entre la marche et la course se produit lorsque des périodes de double appui pendant la phase d'appui du cycle de marche (les deux pieds sont simultanément en contact avec le sol) cèdent la place à deux périodes de double flottement au début et à la fin de la phase oscillante de la marche (aucun pied ne touche le sol). Bien que la course "au sol" de nos jours suggère que certains types de coureurs peuvent avoir une courte phase de double appui. Les démarches de course au sol se produisent le plus souvent dans la plage de vitesses (~ 2 - 3 m s-1) à laquelle ni la course ni la marche ne sont « confortables » ou énergétiquement efficaces. Généralement, à mesure que la vitesse augmente, le contact initial passe de l'arrière-pied à l'avant-pied.

La course est souvent décrite par la vitesse car cela a un sens empirique, car il s'agit de la norme de performance d'un athlète. Par définition, la course se distingue de la marche par une phase de flottaison au cours de laquelle la personne prend une phase d'envolée. Le cycle de foulée est également divisé en une phase d'appui et une phase d'oscillation. La phase d'appui est subdivisée en phases d'absorption et de propulsion, et la phase d'oscillation en phases d'oscillation initiale et terminale.

Le début et la fin de chaque phase oscillante ont une période de double flottement, où aucun membre n'est en contact avec le sol. Par conséquent, par définition, la phase d'appui doit représenter moins de 50 % du cycle de course pour permettre cette période aérienne.

L'augmentation de la vitesse est initialement obtenue en augmentant la longueur du pas, puis en augmentant la cadence. Avec une augmentation de la vitesse, il y a une réduction de la phase d'appui et une augmentation du temps de balancement, ce qui entraîne également une augmentation de la phase de double flottement. À mesure que la vitesse augmente de la course au sprint, la longueur et la cadence des pas sont encore plus grandes. De plus, le sprinteur entre initialement en contact avec le sol avec ses orteils plutôt qu'avec le médio-pied ou l'arrière-pied.

Le modèle de pose du pied est défini comme une analyse biomécanique de la façon dont le pied touche le sol. Une interprétation courante de cette définition utilise la distinction de trois modèles de pose du pied : pose de l'avant-pied (FFS), pose du médio-pied (MFS), pose de l'arrière-pied (RFS).

Outre les classifications de pose de pied, une autre approche est fréquemment utilisée pour décrire les modèles de foulée : l'indice de pose du pied (FSI) ne se concentre pas principalement sur le contact visible du pied avec le sol, mais plutôt sur la mesure des forces qui se produisent pendant la pose du pied. En utilisant cette approche, le modèle de pose peut être déterminé par l'emplacement du centre de pression (COP) par rapport au pied lors du contact avec le sol.

Il existe également la classification de foulée en fonction de l'angle d'attaque que fait le pied par rapport à la perpendiculaire du sol :

- La foulée universelle présente par une pronation « normale ». Le talon se pose sur la partie centro-latérale de la chaussure et par la suite le pied va dérouler vers l'avant et l'intérieur pour finir par une phase de propulsive située sur l'avant pied, idéalement sur le 1er métatarsien bien qu'il soit fréquent d'observer que la phase d'appui se termine sur les têtes métatarsiennes moyennes.
- La foulée pronatrice présente un excès de valgus. Dans ce cas, le talon se pose sur la partie latérale de la chaussure et la pronation amène le pied en interne jusqu'à obtenir une phase propulsive anormale sur l'avant pied.
- La foulée supinatrice sous-entend qu'il y a un manque de pronation lors du déroulé de la foulée, c'est-à-dire que le pied ne vient pas suffisamment s'incliner vers l'intérieur. La foulée va majoritairement se concentrer sur la partie latérale de la chaussure.

3. Analyse quantifiée de la foulée. Matériel et méthodes

L'analyse quantifiée de la marche/foulée est l'étude systématique de la locomotion humaine, utilisant l'œil et le cerveau des observateurs, complétée par des instruments de mesure des mouvements corporels, de la mécanique corporelle et de l'activité musculaire. L'analyse de la marche est utilisée pour évaluer et traiter les personnes atteintes d'affections affectant leur capacité à marcher. Il est également couramment utilisé dans la biomécanique du sport pour aider les athlètes à courir plus efficacement et pour identifier les problèmes liés à la posture ou au mouvement chez les personnes blessées. L'étude comprend la quantification (introduction et analyse des paramètres mesurables des allures), ainsi que l'interprétation des résultats à partir de son schéma de marche. Alors que la course à pied continue de gagner en popularité, l'intérêt pour la recherche et l'évaluation de la foulée augmente également. Les progrès récents et la prolifération de la technologie utilisée dans l'analyse de la foulée rendent cette capacité plus largement accessible à un plus large éventail de professionnels. L'analyse de la foulée n'est plus limitée aux laboratoires d'analyse et de recherche cliniques bien financés.

De nombreux outils ont été développés pour aider à l'évaluation de la foulée. Ceux-ci incluent les systèmes de capture de mouvement plus traditionnels utilisés pour décrire le mouvement du corps, les plateformes de force qui quantifient les forces agissant sur le corps et l'électromyographie (EMG) utilisée pour estimer le niveau d'activité musculaire pendant le mouvement. Plus récemment, des capteurs embarqués plus petits ont été développés et utilisés avec succès pour mesurer les paramètres de marche en cours d'exécution. Ceux-ci incluent des accéléromètres de systèmes micro-électro-mécaniques (MEMS), des électro-goniomètres, des gyroscopes et des semelles à capteurs de pression. Ces outils ont été utilisés avec succès pour étudier les performances des chaussures et des orthèses plantaires, les facteurs de risque de blessure, les performances de course, les effets de la fatigue et les adaptations de la foulée à diverses techniques de course.

ANALYSE DE MOUVEMENT

L'analyse de la marche/foulée nécessite souvent la quantification du mouvement de segments corporels individuels dans un espace à deux ou trois dimensions. La méthode la plus courante pour collecter ces informations consiste à utiliser la technologie de capture de mouvement, dans laquelle des marqueurs sont apposés sur le sujet et suivis tout au long du mouvement d'intérêt. Ces systèmes utilisent généralement des marqueurs passifs qui réfléchissent la lumière ambiante ou infrarouge, des marqueurs actifs qui émettent de la lumière (diodes électroluminescentes) ou des systèmes électromagnétiques qui sont capables de détecter la position et l'orientation du récepteur placé sur un segment du corps. Grâce à des techniques de numérisation manuelles ou automatiques, l'emplacement des coordonnées (bidimensionnelles ou tridimensionnelles) des marqueurs peut être déterminé. A partir de ces données de position, la vitesse et l'accélération peuvent ensuite être calculées en prenant la dérivée temporelle de la position et de la vitesse, respectivement. Les comparaisons directes faites entre les systèmes optiques et électromagnétiques indiquent que les mesures obtenues à partir de l'un ou l'autre système sont comparables et précises. Cependant, chaque système a ses limites. Les systèmes optiques sont sujets à l'occlusion des marqueurs s'il existe un nombre important de marqueurs ou un nombre insuffisant de caméras. Les systèmes électromagnétiques sont sensibles aux interférences magnétiques provenant d'objets métalliques situés à l'intérieur ou à proximité du volume de capture et leur précision est affectée par la distance entre l'émetteur et les récepteurs.

ANALYSE DES FORCES

Les plateformes de force sont couramment utilisées pour mesurer les forces de contact entre le pied et le sol pendant la phase d'appui de la marche/foulée. Ces informations peuvent être utilisées pour quantifier les forces d'impact, les taux de chargement et les forces de propulsion et de rupture, et suivre les changements du centre de pression (CoP) au fil du temps. Cependant, en raison de leur taille relativement petite, ils imposent des contraintes sur le placement du pied, ce qui peut amener les sujets à adopter une stratégie de « ciblage » pendant la course, altérant la biomécanique naturelle de la marche/foulée. Cette stratégie de ciblage peut entraîner une variabilité accrue de la longueur des pas et entraîne souvent l'exclusion de l'essai, ce qui entraîne des périodes de collecte de données prolongées et une fatigue du sujet. Bien que l'influence du ciblage montre peu d'effet sur les forces de réaction au sol pendant la marche, des vitesses d'approche plus élevées associées à la course et des changements plus importants dans la longueur relative de la foulée pour entrer en contact avec la plate-forme de force peuvent entraîner des différences plus importantes observées dans les forces de réaction au sol à ces vitesses plus élevées. Récemment, le développement de tapis roulants instrumentés a permis la collecte rapide des forces de réaction au sol sur des cycles de marche/foulée répétés, permettant une vitesse de marche hautement contrôlée, tout en éliminant les erreurs potentielles introduites par les stratégies de ciblage.

ANALYSE DES PRESSIONS

L'utilisation de capteurs de pression intégrés à la chaussure offre une alternative légère, portable et facile à utiliser pour analyser la foulée. Contrairement aux plates-formes de force, elles sont capables de quantifier la répartition de la force sur la surface plantaire du pied, fournissant des informations plus détaillées sur la charge du pied pendant la marche que les mesures de force seules. Parce que ce dispositif est placé dans la chaussure, les charges agissant sur la surface du pied peuvent être mesurées directement, contrairement à la force agissant sur le bas de la chaussure avec une plate-forme de force standard. Pour cette raison, les capteurs de pression intégrés à la chaussure sont couramment utilisés pour quantifier l'effet de la conception de la chaussure sur la charge du pied et ont été utilisés pour comparer la charge du pied entre des types de chaussures similaires et entre des chaussures de conceptions de semelle intermédiaire différentes, ainsi que des changements dans les capacités d'absorption d'impact d'une chaussure sur des cycles d'impact répétés.

Les capteurs de pression intégrés à la chaussure permettent également de mesurer les forces verticales subies par le pied lors d'une course prolongée et de détecter les paramètres de marche typiques nécessaires à l'analyse de la marche, tels que l'attaque du talon et le décollement des orteils nécessaires pour définir la phase d'appui de la marche. Bien que les plates-formes de force soient considérées comme la méthode de référence par laquelle ces mesures sont généralement collectées, comme mentionné précédemment, elles sont limitées dans le nombre d'étapes pouvant être échantillonnées et sont généralement limitées à une utilisation en laboratoire. Les capteurs de pression intégrés à la chaussure donnent au chercheur ou au clinicien la possibilité de collecter des données à partir de foulées répétées dans un environnement qui facilite une course normale.

ÉLECTROMYOGRAPHIE

L'électromyographie (EMG) est une technique couramment utilisée pour mesurer les niveaux d'activité musculaire pendant la marche ou la course à pied. En règle générale, le moment de l'activation musculaire et l'intensité relative sont les principales mesures d'intérêt et peuvent être recueillies grâce à l'utilisation d'électrodes de surface. Cette technique peut être utilisée pour détecter un comportement de marche anormal et évaluer le contrôle neuromusculaire d'un coureur. L'activation musculaire normale pendant les phases de foulée et de position de la course et du sprint a été étudié, ainsi que la façon dont l'activation musculaire et les schémas de synchronisation sont influencés par les changements de vitesse de marche et de course.

En plus de fournir des informations sur les niveaux d'activation musculaire et le moment, le contenu fréquentiel du signal EMG peut être analysé pour déterminer la fatigue musculaire relative, qui peut être utilisée dans la détection précoce de blessures potentielles en cours d'exécution. Les paramètres EMG se sont avérés hautement reproductibles entre les cycles de foulée lorsqu'ils sont comparés à différentes techniques de course (vitesse de course et fréquence de foulée), mais ont tendance à être moins reproductibles lorsqu'ils sont comparés entre les muscles individuels, les muscles distaux des jambes montrant plus de reproductibilité que les muscles proximaux des jambes pendant course.

SYSTÈMES MICRO-ÉLECTROMÉCANIQUES (MEMS)

L'utilisation de capteurs fixes tels que les accéléromètres deviennent rapidement une alternative viable aux techniques d'analyse de la marche/foulée plus traditionnelles pour une utilisation dans l'évaluation du mouvement humain. Les accéléromètres sont des capteurs inertiels qui fournissent une mesure directe de l'accélération sur un ou plusieurs axes, réduisant efficacement l'erreur associée à la différenciation des données de déplacement et de vitesse dérivées de sources telles que les systèmes de capture de mouvement.

Des systèmes basés sur un accéléromètre ont été utilisés avec succès pour quantifier le choc subi par le membre inférieur pendant la marche et la course, évaluer l'effet des chaussures et des semelles orthopédiques sur le choc tibial pendant la course, évaluer l'atténuation des chocs entre les segments du corps pendant la course et étudier les effets de la fatigue lors de l'exécution des modèles de marche. La capacité de certains accéléromètres à répondre à la fois à l'accélération gravitationnelle ainsi qu'à l'accélération causée par le mouvement leur permet également d'être utilisés pour la mesure de l'orientation des segments dans des conditions statiques. L'utilité des accéléromètres dans l'analyse de la marche peut être encore améliorée grâce à l'utilisation simultanée de gyroscopes et d'électro-goniomètres.

Les accéléromètres, lorsqu'ils sont combinés avec des gyroscopes, se sont avérés produire un angle d'articulation, une vitesse angulaire et une accélération angulaire similaires dérivés de systèmes de capture de mouvement dans des conditions dynamiques, et ont été utilisés efficacement pour estimer la vitesse de marche et les angles d'inclinaison de la surface. L'avantage le plus attrayant est peut-être la capacité des accéléromètres à être utilisés dans l'estimation des paramètres spatio-temporels de la marche, ce qui, jusqu'à récemment, nécessitait l'utilisation d'une plaque de force, de systèmes d'analyse de mouvement ou de pédales.

En raison de leur légèreté et de leur portabilité, les accéléromètres sont capables d'enregistrer des données qui peuvent être collectées en continu sur de nombreux cycles de foulée pendant une période prolongée. Cette technologie a été utilisée efficacement pour détecter les altérations des schémas de course après l'apparition de la fatigue chez les coureurs de demi-fond lors d'une course sur piste, sans modifier les schémas de course du coureur. Bien que les tests mécaniques aient confirmé la validité et la fiabilité des accéléromètres dans la mesure des accélérations dans la plage de fréquence et d'amplitude des mouvements du corps humain, les preuves indiquent qu'ils sont sensibles au site et à la méthode de fixation, les accéléromètres montés sur la peau entraînant une augmentation significative accélérations maximales par rapport aux accéléromètres montés sur os. Indépendamment des limites associées aux accéléromètres montés sur la peau, ils ont été utilisés avec succès pour distinguer les coureurs ayant des antécédents de fractures de fatigue tibiales et les coureurs asymptomatiques.

La fatigue est un état physiologique causé par des restrictions physiques, physiologiques ou psychologiques. Dans le sport, nous utilisons les expressions anglaises suivantes pour différencier l'épuisement naturel et justifiable de la fatigue qui résulte d'une activité excessive et qui s'accumule :

• Functional overreaching (FOR) = Surmenage/ Dépassement fonctionnel de courte durée

• Nonfunctional overreaching (NFOR) : Surmenage/ Dépassement non-fonctionnel extrême

• Overtraining syndrome (OTS) : Syndrome de surentraînement (nécessite une récupération de plusieurs mois).

La fatigue neuromusculaire est une forme d'épuisement qui affecte à la fois les systèmes neurologique et musculaire. Elle affecte la capacité à produire une performance physique ou mentale. Il peut être actif à divers niveaux de production de performance. C'est la fatigue globale qui aura un effet sur la performance, bien que cela puisse commencer à différents niveaux du système et être induit par une variété de variables. Il existe deux principaux types de fatigue neuromusculaire, en fonction de la cause de la diminution de la production de force :

• La fatigue centrale : la fatigue qui trouve son origine au niveau du cerveau (supraspinale) et au niveau de la moelle épinière (spinale)

• La fatigue périphérique : celle qui est originaire d'altérations au niveau musculaire. Elle peut être due à la variation de l'excitabilité musculaire, du couplage excitation/contraction ou de la contractilité musculaire.

Comme son nom l'indique, la fatigue neuromusculaire peut avoir une cause neurologique ou musculaire. En laboratoire, de nombreuses méthodes sont utilisées pour déterminer l'origine d'une substance. La contraction volontaire maximale (CMV) est utilisée pour quantifier la fatigue totale générée par l'exercice. Il s'agit d'une courte contraction maximale d'un muscle ou d'un groupe de muscles avec stimulation visuelle et rétroaction. Le but de cette mesure est de déterminer la force de sortie maximale du sujet. Le test est ensuite répété après une tâche pour évaluer si cette tâche a généré de la fatigue ou non. Il s'agit d'un épuisement général dû au fait que la contraction maximale volontaire est une performance à part entière, incluant ainsi à la fois des éléments neurologiques et musculaires. Ainsi, la fatigue neuromusculaire est fréquemment caractérisée dans la littérature comme une diminution de la force maximale volontaire.

L'hypothèse que la fatigue liée à la course altère les fonctions biomécaniques et neuromusculaires telles que le temps de réaction, la coordination des mouvements et la précision du contrôle moteur, la capacité de génération de force musculaire et les performances de course. Plusieurs variables biomécaniques englobent de nombreux phénomènes neuromusculaires et mécaniques caractérisant simultanément le système de course : longueur et fréquence des foulées, temps de contact et de vol, force maximale verticale, déplacement ou raideur du centre de masse. L'examen de ces paramètres est d'un intérêt primordial lors de l'étude de la façon dont la fatigue peut affecter la biomécanique de course lors d'une course de fond. L'étude de la fatigue lors d'activités telles que la course à pied est compliquée car la nature exacte de la fatigue et les muscles exacts qui subissent la fatigue sont inconnus. Il se peut que les muscles critiques actifs pendant la course fonctionnent à des capacités différentes les uns par rapport aux autres pendant la course sous fatigue. Il existe plusieurs mécanismes physiologiques et neurologiques liés à l'incapacité du muscle à générer une sortie de force attendue.

En course à pied, une inversion constante de la génération et de l'absorption d'énergie se produit au niveau des articulations de la hanche et du genou. Pendant la partie d'absorption de la position et les phases de balancement initial et terminal, la hanche génère de l'énergie tandis que le genou ipsilatéral absorbe de l'énergie. Par exemple, dans la phase d'oscillation initiale, la hanche fléchit et génère de la puissance avec une contraction concentrique du droit fémoral tandis que le genou fléchit avec absorption de puissance avec la contraction excentrique du droit fémoral distal. Pendant la phase de balancement terminale, les muscles ischio-jambiers remplissent une fonction similaire avec une contraction concentrique à la hanche et une contraction excentrique au genou. Ainsi, pendant la course, les muscles de la double articulation (le droit fémoral et les ischio-jambiers), qui traversent les deux articulations, se contractent concentriquement à une extrémité et excentriquement à l'extrémité opposée, permettant un transfert d'énergie efficace entre les articulations.

Dans la course humaine, les mécanismes élastiques sont utilisés de deux manières principales. Premièrement, les mécanismes élastiques réduisent le travail que les muscles doivent faire et économisent ainsi de l'énergie. Pendant la course, un athlète monte et descend, gagnant et perdant de l'énergie potentielle gravitationnelle. De plus, le centre de masse du corps accélère et décélère, de sorte que l'athlète gagne et perd alternativement de l'énergie cinétique externe.

Deuxièmement, les structures élastiques servent de ressorts de suspension. Les propriétés élastiques de certaines structures des membres inférieurs (principalement le pied) modèrent la force d'impact, à l'instar des ressorts d'une voiture qui empêchent le conducteur de subir un choc violent lors de la conduite sur une bosse. Ce deuxième mécanisme évite une décélération rapide du pied et d'éventuels dommages anatomiques lors de la phase d'atterrissage de la foulée. Les muscles et les tendons sont les matériaux généralement considérés comme des réserves d'énergie de déformation lors de la course à pied. Alexander et Bennet-Clark ont souligné qu'un muscle et son tendon transmettent des forces égales, mais que des contraintes élastiques plus importantes étaient susceptibles de se produire dans le tendon. En cas de fibres musculaires courtes et de tendons longs (par exemple triceps sural), le tendon stockerait la majeure partie de l'énergie de déformation.

D'un point de vue mécanique, la course à pied est un mouvement humain typique où les structures musculo-tendineuses du membre inférieur stockent et restituent alternativement de l'énergie élastique pendant le cycle étirement-raccourcissement. Ainsi, les membres inférieurs peuvent être considérés comme des ressorts chargés par le poids et l'inertie de la masse corporelle. Ce paradigme fait référence au « modèle ressort-masse » et est de plus en plus utilisé ces dernières années pour décrire le comportement (régulation de la raideur) du système musculo-squelettique des membres inférieurs lors des phases de rebonds et d’envolées de la course. Dans ce modèle, la rigidité du ressort de la jambe représente la rigidité globale moyenne du système musculo-squelettique intégré pendant la phase de contact avec le sol (appelée « raideur de la jambe ») et est définie comme le rapport de la force maximale à la force maximale compression des jambes au milieu de la phase d'appui. De plus, la rigidité verticale est utilisée pour modéliser le mouvement vertical du centre de masse (COM) pendant le contact est défini comme le rapport de la force maximale au déplacement maximal vertical vers le bas du COM lorsqu'il atteint son point le plus bas, (c'est-à-dire à milieu de la phase d'appui). Pendant la course, le ressort de jambe est comprimé pendant la première moitié de la phase de contact avec le sol et s'allonge pendant la seconde moitié de la phase de contact avec le sol.

Pendant que nous courons, chaque pied entre en contact avec le sol environ 80 à 100 fois par minute en moyenne. Cela se traduit par un taux de foulée/cadence de 160-200 pas/minute, la cadence varie d'une personne à l'autre et dans une certaine mesure avec la vitesse de course. Chaque fois que vous atterrissez, votre pied touche le sol avec une certaine force, qui est contrebalancée par une force égale et opposée appliquée par le sol sur votre pied. Cette force égale et opposée est connue sous le nom de force de réaction au sol ou GRF en abrégé. La force de réaction au sol se décline en un certain nombre de composants, généralement divisés en antérieur-postérieur (dans la direction dans laquelle vous vous déplacez), horizontal (d'un côté à l'autre) et vertical (de haut en bas). Parmi ceux-ci, la GRF verticale est la plus grande en magnitude.

La collision avec le sol à chaque foulée crée une onde de choc qui se transmet dans tout le corps. L'atténuation des chocs est le processus d'absorption de l'énergie d'impact, réduisant ainsi l'énergie d'impact entre le pied et la tête. Comprendre les facteurs qui affectent l'atténuation de l'impact pendant la course est important, car les forces de réaction au sol de deux à trois fois le poids corporel agissent sur le corps à travers le pied à chaque pose du pied et environ 5000 poses de pied se produisent pendant une course typique de 30 minutes. L'énergie des ondes de choc est absorbée par des composants tels que les chaussures de course, les surfaces, les muscles, les os et d'autres tissus structurels. Les actions articulaires telles que la flexion de la cheville, du genou et de la hanche servent également à absorber l'énergie des ondes de choc (soulignant le rôle important que jouent les muscles dans l'absorption de l'énergie d'impact. Étant donné que l'atténuation des chocs est réalisée par la contraction musculaire et les capacités d'absorption d'énergie des structures anatomiques telles que les os et le coussinet adipeux calcanéen, il se peut que certaines structures anatomiques soient soumises à des contraintes plus importantes lors de l'impact si les muscles sont fatigués, où la fatigue musculaire est définie comme «l'incapacité à maintenir la force ou la puissance requise ou attendue». Nordin et Frankel ont émis l'hypothèse que les blessures par surmenage osseux sont liées à la fatigue musculaire, soit en raison d'une perte de capacité d'absorption des chocs des muscles, soit en raison d'un changement dans le schéma neuro-moteur de mouvement pour compenser le changement dans la capacité musculaire. Une chose reste claire : les blessures de course se développent en raison d'interactions complexes entre de nombreuses variables, quel que soit le modèle de foulée. Un examen plus approfondi des variables liées à l'impact peut révéler que les articulations ou les tissus susceptibles de se blesser peuvent différer selon les modèles de foulée. Les événements entourant la collision pied-sol pendant la course sont la principale source du d'impact qui se transmet à travers la jambe et le reste du corps L'impact doit être atténué pour éviter la perturbation des systèmes vestibulaire et visuel à la suite d'une accélération excessive de la tête. L'atténuation se produit principalement par l'absorption d'énergie des muscles actifs, les modifications de la géométrie des articulations et la déformation des tissus passifs. Le corps réagit à des amplitudes d'impact plus importantes en augmentant l'atténuation grâce à une combinaison de mécanismes actifs et passifs. Le recours à certains mécanismes d'atténuation des chocs peut dépendre du contenu fréquentiel du choc d'impact La capacité et le degré d'atténuation seront dictés par le contenu fréquentiel de l'impact et les mécanismes disponibles pour l'atténuation. Une capacité réduite d'atténuation par certains tissus ou mécanismes peut entraîner une plus grande dépendance vis-à-vis d'autres tissus ou mécanismes et pourrait potentiellement entraîner la surcharge d'un tissu.

Le but de notre 1ère était de comparer le modèle de pose du pied de course sur tapis roulant à l'aide d'une plate-forme de mesure de force (incluse directement dans le tapis roulant instrumenté) et de gyroscopes pour d'éventuelles similitudes/différences dans l'analyse de la marche et la cinématique des articulations des jambes. Nous avons émis l'hypothèse que malgré l'utilisation de différents outils pour analyser les modèles de course, nous serions en mesure d'atteindre le même résultat final et la même conclusion.

Pour cette étude comparative des données qualitatives et quantitatives, nous avons utilisé le modèle Support Vector Machine utilisant la méthode de Kernel. Ce type d'algorithmes est particulièrement utilisé pour l'analyse des modèles, elle nous permet d'étudier et de comparer des types généraux de relations au sein des modèles de pose du pied. Nous avons utilisé la classification de pose du pied comme suit : pose de l'arrière-pied (RFS), pose du milieu du pied (MFS) et pose de l'avant-pied (FFS).

Nous avons trouvé une bonne corrélation pour la plupart des données de 80,6 %, associé à un faible biais et démontrant un fort accord entre les deux méthodes. La marge d'erreur de 19,4 % pourrait être liée à l'emplacement du gyroscope MEMS et à la manière dont il était fixé. Bien que les tests biomécaniques aient confirmé la validité et la fiabilité des accéléromètres dans la mesure des accélérations dans la plage de fréquence et d'amplitude du mouvement du corps humain, les preuves indiquent qu'ils sont sensibles au site et à la méthode de fixation, également des études de vitesse plus rapides ont trouvé jusqu'à 7% erreur dans l'identification de l'étape.

Généralement, les pics des signaux de temps d'accélération des accéléromètres montés sur la peau ont été utilisés pour mesurer l'amplitude des ondes de choc d'impact. Alors que la composante dominante de ces pics est l'onde de choc, ils comprennent également des composantes d'accélération dues à l'action musculaire et du bruit dû à la résonance dans la fixation conforme de l'accéléromètre au corps. Ainsi, les analyses dans le domaine temporel de l'onde de choc d'impact sont limitées en précision. Bien que l'analyse spectrale ou fréquentielle de l'onde de choc offre la possibilité d'une étude plus détaillée et de déterminer directement la transmissibilité du choc dans le corps humain, les caractéristiques spectrales de l'onde de choc d'impact qui se propage lors d'activités sportives sont largement inconnues.

Pour notre 2ème étude nous nous sommes concentrés sur la décomposition harmonique des signaux d'analyse de foulée. La course à pied est une activité essentiellement périodique. Par conséquent, il semble naturel d'utiliser les composantes harmoniques comme caractéristiques de la foulée. Nous utilisons la décomposition de Fourier de la série chronologique des caractéristiques des données de foulée comme base à partir de laquelle extraire les harmoniques fondamentales et d'ordre supérieur. Intuitivement, l'amplitude mesurée à la fréquence fondamentale est une mesure du changement global subi par la caractéristique correspondante, et la phase relative entre différentes séries temporelles est une indication du délai entre les différentes caractéristiques. Les harmoniques supérieures mesurées par rapport à l'harmonique fondamentale décrivent la trajectoire non sinusoïdale mais toujours périodique que subit une caractéristique. Un cycle de course complet, ou une foulée, est composé de deux pas, le pas gauche et le pas droit. Par conséquent, la période fondamentale de chaque série de données consiste en la moitié d'un cycle de fonctionnement, c'est-à-dire soit le pas de gauche, soit le pas de droite.

La majorité des humains prennent un peu moins d'une seconde pour une foulée complète. Pour distinguer la fonction de décomposition harmonique des caractéristiques d'apparence moyenne, nous supprimons la moyenne de toutes les composantes, fixant ainsi la composante DC de la transformée de Fourier à 0. En fait, les composantes moyennes des caractéristiques d'apparence moyenne de la foulée sont les composantes harmoniques zéro. L'analyse harmonique des séries d'accélération n'est appliquée qu'aux caractéristiques extraites des chaussures et de la jambe. Certains des spectres de puissance semblent avoir des pics dominants, tandis que d'autres manquent de tels pics. La plupart des spectres montrent une fréquence fondamentale, certains ont même une amplitude importante dans la deuxième harmonique, mais peu montrent une troisième harmonique évidente. De plus, il existe plusieurs raisons pour lesquelles nous ne pouvons que raisonnablement nous attendre à récupérer la première et la deuxième harmonique. Premièrement, les harmoniques les plus élevées ont une amplitude plus faible et sont donc plus sensibles au bruit. Deuxièmement, parce que nos sujets n'ont pas de foulées de course parfaitement périodiques, la localisation de la fréquence fondamentale contient des erreurs, qui sont amplifiées aux harmoniques les plus élevées, augmentant ainsi encore la quantité de bruit dans les estimations d'amplitude et de phase aux harmoniques les plus élevées.

Alors que les composantes harmoniques fondamentales capturent la majorité des informations de la série d'accélération, des harmoniques plus élevées sont nécessaires pour capturer ces variations. Intuitivement, l'amplitude de la fréquence fondamentale ainsi que l'amplitude de la deuxième harmonique et la phase de la deuxième harmonique par rapport à la fréquence fondamentale fournissent une description indépendante de la traduction d'un signal qui ne contient que des première et deuxième harmoniques. Nous ne regardons pas au-delà de la deuxième harmonique car le taux d'échantillonnage et la quantité de bruit dans la série d'accélération rendent instables les composantes harmoniques supérieures. Une inspection visuelle des données d'analyse clinique de la foulée montre clairement que la plupart des séries chronologiques de paramètres de marche ne sont pas de pures sinusoïdes. Cependant, il est beaucoup moins clair si les harmoniques supérieures, en particulier la deuxième harmonique, peuvent être facilement récupérées à partir des caractéristiques des séries d'accélération qui étaient elles-mêmes dérivées des données d'accélération.

On peut considérer un enregistrement accélérométrique de course à pied comme un signal périodique et l'analyser sous l'aspect de la transformation de Fourier. Cependant à bien y regarder, chaque foulée génère son signale que l'on peut considérer comme un choc individuel. Dans ce contexte l'enregistrement peut être découpé en une série d'évènements élémentaires, tous semblables mais tous différents, qui seront analysés comme des événements transitoires intenses : des chocs.

L’outil d’analyse des chocs utilisés dans l’industrie s’appelle le spectre de réponse au choc. Il permet de comparer les chocs entre eux et surtout leur interaction avec la structure de réception, ici le corps humain. Cet outil permet également d’appréhender les phénomènes de fatigue sur une structure soumise à des chocs répétitifs si on y intègre des lois de fatigue associée. Le spectre de réponse aux chocs (SRS) est calculé à partir du signal d'un accéléromètre. Le signal d'accélération est utilisé pour l'excitation primaire d'une série de systèmes à un seul degré de liberté (SDOF) avec des fréquences naturelles personnalisables. Le spectre est formé par les maxima, maxima ou minima absolus des réponses de ces systèmes. Le spectre de réponse aux chocs a été introduit à l'origine pour analyser le potentiel de dommages des impulsions mécaniques, mais il peut également être utilisé pour analyser le potentiel de dommages des vibrations aléatoires stationnaire.

$\ddot{Y}$ ... Signal d'entrée

Mi ... Masse du ième système

Ci ... Coefficient d'amortissement du ième système

Ki ... Rigidité du ième système

fni ... Fréquence naturelle du ième système

$\ddot{X}_i$ ... Réponse en accélération du ième système

Toute forme d'onde transitoire peut être présentée comme un SRS, mais la relation n'est pas unique ; de nombreuses formes d'ondes transitoires différentes peuvent produire le même SRS (quelque chose dont on peut tirer parti grâce à un processus appelé "Synthèse de choc"). En raison du suivi uniquement de l'accélération instantanée maximale, le SRS ne contient pas toutes les informations de la forme d'onde transitoire à partir de laquelle il a été créé. Différents rapports d'amortissement produisent différents SRS pour la même forme d'onde de choc. Un amortissement nul produira une réponse maximale. Un amortissement très élevé produit un SRS très ennuyeux : une ligne horizontale. Le niveau d'amortissement est démontré par le "facteur de qualité", Q que l'on peut aussi penser à la transmissibilité en cas de vibration sinusoïdale.

Un système subit une accélération qui est mesurée (Entrée), un système constitué d'une série de systèmes masse-ressort-amortisseur répond à l'accélération transitoire. L'accélération en fonction du temps de réponse de chaque système masse-ressort est enregistrée, en tant que sortie, la réponse de niveau de crête de chaque système masse-ressort-amortisseur est tracée en fonction des fréquences naturelles correspondantes des systèmes qui est le spectre de réponse aux chocs (SRS).

Le signal d’accélération est enregistré sur les 3 axes simultanément, la norme a(t) du signal est calculée afin de ne travailler que sur la longueur du vecteur accélération.

On calcule ensuite la densité spectrale (PSD Density Spectral Frequency) de puissance grâce à une transformation de Fourier rapide (Fast Fourier Transform FFT). Ce calcul nous permet alors de déterminer la fréquence fondamentale du coureur f_{fond} qui correspond à la position du pic le plus grand de la PSD.

$$PSD = \frac{1}{N}\left|\int_{-\infty}^{+\infty} a(t)e^{-j2\pi\upsilon t}\,dt\right|^2$$

avec N nombre de points de l’enregistrement, a le module de l’accélération, v la fréquence et t le temps.

On effectue ensuite la cross-correlation CC(?) entre ce signal pattern du coureur et l’ensemble de l’enregistrement a(t). On constate qu’à chaque pas, la convolution est maximale. En effet la convolution est une méthode permettant d’estimer la ressemblance entre deux signaux. A chaque pic dans le signal de convolution, cela veut dire que le signal émit par le coureur est similaire au pattern. Cette méthode complexe de prime abord permet de déterminer quant à lieu un pas même si le coureur n'est pas complétement régulier.

$$CC(\tau) = \int_{-\infty}^{+\infty} a(t)pattern(t+\tau)dt$$

Pour chaque valeur maximale de CC(t) on peut alors extraire un évènement similaire au pattern, on « extrait » ainsi chaque pas élémentaire du coureur. Une fois l'extraction effectuée, nous calculons le SRS de chaque pas et du signal au complet. L'algorithme utilisé est celui de [Lalanne 2009].

Le SRS est une méthode d'analyse des régimes transitoires (chocs) développé par l'industrie militaire pour déterminer la façon dont les équipements réagissent à des chocs répétitifs. De là à transposer cette méthode sur l'analyse des signaux d'un coureur à pied il n'y a qu'un pas.

Le SRS consiste à calculer l'accélération maximale que va subir un système à un degré de liberté dont on connait la fréquence propre f0 et le coefficient de qualité Q pour chaque fréquence propre possible. Généralement ces courbes sont tracées avec un coefficient de qualité Q=10. La méthode de détermination du SRS consiste soit à calculer la réponse temporelle du système à la sollicitation ce qui aboutit à des temps de calcul prohibitifs, soit à utiliser l'algorithme développé par [Lalanne 2009] qui permet d'atteindre les résultats de façon extrêmement rapide. On peut alors comparer les événements individuellement et par gamme de fréquence. De façon générale plus le SRS est élevé pour une fréquence donnée et plus l'événement est agressif. On peut alors imaginer en abscisse des SRS positionner les fréquences des éléments caractéristiques du corps humain.

Cette méthode permet ainsi de déterminer dans une course les pas agressifs pour une structure particulière (genoux, mollet…. A condition d'en connaitre la fréquence naturelle. Le SRS peut également être calculé sur l'intégralité de l'enregistrement. On constate alors l'apparition de pics aux fréquences fondamentales et harmoniques du signal enregistré. Dans ce contexte le SRS combine à la fois la notion de fonction de transfert et de réponse aux régimes transitoires. Comparer les SRS entre eux apporte beaucoup plus finesse à l'analyse dans la mesure où la fréquence est également présente. L'agressivité d'un mouvement n'est pas due qu'à la valeur du maximum d'accélération, mais aussi à la forme générale du mouvement, seul le SRS permet de prendre cela en compte dans l'analyse.

Le but de notre 3ème étude était de déterminer les effets de l'augmentation des niveaux de choc d'impact sur les caractéristiques spectrales du corps pendant la course sur tapis roulant. Une nouvelle méthodologie pour l'analyse des événements de choc survenant au cours de la procédure expérimentale proposée. Notre approche est basée sur le spectre de réponse aux chocs (SRS), qui est une fonction basée sur la fréquence utilisée pour indiquer l'amplitude des vibrations dues à un choc ou à un événement transitoire. L'objectif principal est d'analyser la capacité du système musculo-squelettique humain à atténuer les contraintes mécaniques résultant de l'effet de fatigue par Shock Responses Spectrum (SRS) des ondes de choc générées par la pose du pied pendant la course. SRS est une fonction basée sur la fréquence utilisée pour indiquer l'amplitude des vibrations dues à un choc ou à un événement transitoire. C'est une méthode d'analyse des systèmes transitoires (chocs) développée par l'industrie militaire pour déterminer comment les équipements réagissent aux sollicitations répétitives. A partir de là, la transposition de cette méthode à l'exécution de l'analyse de la foulée il n'y a qu'un pas.

L'un des objectifs de la présente étude était d'analyser l'effet de la fatigue par SRS sur la capacité du système musculo-squelettique humain à atténuer les ondes de choc générées par la foulée. La fatigue entrave la capacité du système musculo-squelettique humain à se protéger contre la surcharge due aux ondes de choc générées par les foulées. La perte de protection peut se manifester par une augmentation de l'amplitude de l'onde de choc mesurée sur la tubérosité tibiale. Les résultats ont été obtenus grâce à ce protocole alors que les sujets couraient sur un tapis roulant motorisé. Bien qu'une telle configuration simplifie l'acquisition de données, les modèles de locomotion obtenus peuvent différer des modèles de course au-dessus du sol. Les coureurs de la présente étude étaient contraints de courir à une vitesse constante, qu'ils soient fatigués ou non. Étant donné que l'un des principaux objectifs de l'étude du SRS était d'introduire la corrélation entre la fatigue de la course et les blessures aux membres inférieurs, il est important de noter que cela peut légèrement différer dans la course au-dessus du sol : lorsque la fatigue commence, les coureurs peuvent ralentir comme moyen de protection. Le résultat pourrait être de s'éloigner de l'état de fatigue, auquel cas les données d'accélération pourraient ne pas augmenter. Selon les résultats de cette étude, pour que les données d'accélération augmentent, la fatigue doit être présente. Ainsi, les résultats de cette étude peuvent être extrapolés à la course en surface si la fatigue prévaut effectivement. De plus, la plupart des blessures de course en surface sont des blessures des membres inférieurs, avec une prédominance des blessures au genou et nos résultats indiquant une augmentation des données d'accélération dans la tubérosité tibiale tendent donc à soutenir cette extrapolation. Des études antérieures ont montré que le taux de chargement du membre inférieur est directement et fortement corrélé à la vitesse de course, et que la force d'impact verticale augmente avec l'augmentation de la vitesse de course. L'activation des muscles réduit la contrainte de flexion sur les os et atténue les charges dynamiques maximales qui peuvent endommager les tissus musculo-squelettiques. Des études antérieures ont suggéré que les muscles fatigués ne peuvent pas supporter une course «optimale» et elles ont également suggéré que la fatigue du coureur peut entraîner une modification de la mécanique de la phase d'atterrissage. Il a également été constaté que le transfert d'énergie mécanique entre les phases excentrique et concentrique est considérablement réduit lors de la fatigue musculaire. De tels changements peuvent être impliqués dans le développement de blessures.

Nous pouvons conclure que le système musculo-squelettique devient moins capable de gérer les ondes de choc induites par les foulées lorsque les muscles sont considérablement fatigués. Comprendre l'influence du SRS sur la fatigue et sur l'ampleur de la charge dynamique sur le système musculo-squelettique humain permettra le développement de procédures d'entraînement et d'exercices appropriés, et réduira les dommages aux tissus musculo-squelettiques.

Dans notre 4ème article l'objectif principal est d'analyser la capacité du système musculo-squelettique humain à atténuer les contraintes mécaniques résultant de l'effet de fatigue par Shock Responses Spectrum (SRS) des ondes de choc générées par la frappe du pied pendant la course. La plupart des études précédentes se sont concentrées sur les chocs/impacts, la réaction des forces au sol ou le taux de charge d'impact spectral ou vertical. L'utilisation du SRS comme mesure dans l'analyse de la marche en cours d'exécution n'a jamais été étudiée à ce jour. Cette approche innovante pourrait ouvrir la voie à une toute nouvelle façon d'évaluer la foulée d'un coureur à l'aide de chaussures connectées intelligentes. Le but de cette étude était de déterminer l'effet de la fatigue sur l'atténuation des ondes de choc d'impact et d'évaluer la relation entre la biomécanique humaine et l'atténuation des chocs pendant la course. Il a été émis l'hypothèse que la fatigue entraînerait une diminution de la capacité d'atténuation des chocs du système musculo-squelettique, impliquant ainsi potentiellement un risque plus élevé de blessure due au surmenage.

L'hypothèse est que la fatigue entrave la capacité du système musculo-squelettique humain à se protéger de la surcharge due aux ondes de choc générées par les foulées, la perte de protection peut se manifester par une amplitude accrue des ondes de choc. Pour les cinq athlètes, il y avait une corrélation directe entre la fatigue et une augmentation de l'agressivité du SRS. Nous avons remarqué que pour les cinq athlètes pour la 3ème manche, le pic SRS moyen était significativement plus élevé que pour la 1ère manche et la

2ème manche ($p < 0,01$) à la même fréquence naturelle de l'athlète. Cela confirme notre hypothèse selon laquelle la fatigue entraîne une diminution de la capacité d'atténuation des chocs du système musculo-squelettique, impliquant ainsi potentiellement un risque plus élevé de blessure due au surmenage.

Les résultats obtenus ont montré que l'amplitude d'accélération augmentait régulièrement avec le groupe de fatigue et qu'il y avait une association claire entre la fatigue et les ondes de choc (comme révélé par le SRS). Nous pouvons alors confirmer les conclusions des études susmentionnées, à savoir que le système musculo-squelettique humain devient moins capable d'absorber les ondes de choc induites par l'impact d'une seule jambe lorsque les muscles sont considérablement fatigués. Cette condition peut favoriser le développement de blessures et les résultats actuels ont une implication significative sur l'étiologie des blessures de course. Selon les résultats présentés dans cette étude, pour que les données d'accélération augmentent, il faut que la fatigue soit présente. Nous pouvons conclure que le système musculo-squelettique devient moins capable de gérer les ondes de choc induites par les coups de pied lorsque les muscles sont considérablement fatigués.

L'une des fonctions importantes du système musculo-squelettique humain est d'atténuer et de dissiper les ondes de choc initiées par le contact du pied au sol. Ces ondes de choc sont déclenchées par la plupart des types de mouvement, comme la marche et la course. La course implique des impacts répétés d'une seule jambe entre le pied et la surface. De tels impacts sont caractérisés par un pic transitoire de la force de réaction au sol (force d'impact), une décélération rapide du membre inférieur (choc d'impact) et l'initiation d'une onde d'accélération et de décélération (onde de choc d'impact) qui se propage à travers le corps.

4. Résultats, discussion et conclusion

Les blessures surviennent lorsque l'énergie est transférée au corps en quantités ou à des taux qui dépassent le seuil de tolérance à la lésion des tissus humains. Dans les blessures sportives, on parle généralement de transfert d'énergie mécanique. Ces définitions conceptuelles cèdent généralement la place à une prise en charge des blessures répondant à certains critères d'arrêt d'activité ou de traitement médical. En effet, une récente déclaration consensuelle sur les définitions des blessures et les procédures de collecte de données dans le football suggère que les blessures sont : « Toute affection physique subie par un joueur à la suite d'un match de football ou d'un entraînement de football, indépendamment de la nécessité de soins médicaux ou perte de temps des activités de football.'' Plusieurs définitions de la blessure ont été proposées dans le passé, parfois similaires ; cette hétérogénéité dans la terminologie peut prêter à confusion. Pour Junge et al. une blessure était définie comme « toute affection musculo-squelettique nouvellement survenue en raison de la compétition et/ou de l'entraînement pendant le tournoi et ayant reçu des soins médicaux, quelles que soient les conséquences liées à l'absence de la compétition ou de l'entraînement ».

La course à pied est l'une des activités sportives les plus populaires et les plus accessibles dans le monde entier et elle est devenue de plus en plus populaire au cours des 50 dernières années. Le nombre de coureurs et d'événements de course à pied a considérablement augmenté au cours des dernières décennies car ce sport est peu coûteux et peut être facilement mis en œuvre avec un équipement minimal par une variété de personnes.

Plus important encore, la course à pied est une excellente forme d'exercice pour les personnes qui cherchent à atteindre une forme physique et/ou un mode de vie plus sain, car elle a été associée à la longévité et à la réduction des facteurs de risque de maladies cardiovasculaires. Malgré ces avantages pour la santé, les blessures musculo-squelettiques liées à la course (Running Related Musculoskeletal Injuries) sont courantes chez les coureurs. Ces RRMI sont généralement causées par l'application de charges relativement faibles sur de nombreux cycles répétitifs. Diverses études ont examiné la proportion de blessures (taux d'incidence et de prévalence) chez les coureurs, avec des taux d'incidence variant entre 3,2 % et 84,9 %. Cette grande variation peut s'expliquer par les différences dans la conception des études, les définitions des blessures, les caractéristiques des sujets et les périodes de suivi, qui peuvent toutes différer d'une étude à l'autre.

L'incidence et la prévalence sont fondamentalement différentes, mais les deux sont importantes dans les études épidémiologiques. L'incidence est l'indication du nombre d'occurrences de nouvelles blessures sportives. Il transmet des informations sur le risque de blessure et n'est généralement disponible que dans des études prospectives. La prévalence indique l'étendue de la blessure dans l'échantillon de population et est généralement rapportée dans des études rétrospectives. Par conséquent, l'élaboration de programmes efficaces de prévention des blessures peut réduire l'incidence des blessures et, par conséquent, la prévalence des blessures. La course à pied est l'une des activités les plus répandues qui provoque des blessures de surmenage du bas du dos et des membres inférieurs. En règle générale, 50 % des coureurs subissent chaque année une blessure qui les empêche de courir pendant un certain temps, et 25 % des coureurs se blessent à un moment donné. Environ 70 à 80 % des troubles de la course sont dus à des blessures de surmenage, impliquant principalement les sites anatomiques du genou, de la cheville/du pied et du tibia.

Une blessure de surmenage est un terme utilisé pour décrire une blessure qui survient à la suite d'une lésion tissulaire résultant d'une demande répétitive sur une période de temps plutôt qu'une blessure aiguë telle qu'une luxation de l'épaule ou une entorse de la cheville. Quelques exemples courants de blessures dues au surmenage comprennent le conflit à l'épaule, l'épicondylite latérale (tennis elbow), la tendinite et les fractures de stress. Une blessure due au surmenage provient généralement de :

• Erreurs d'entraînement. Des erreurs d'entraînement peuvent survenir lorsque vous faites trop d'activité physique trop rapidement. Aller trop vite, faire de l'exercice trop longtemps ou simplement faire trop d'un type d'activité peut fatiguer vos muscles et entraîner une blessure due au surmenage.

• Erreurs techniques ou « technopathie ». Une mauvaise technique peut également avoir des conséquences néfastes sur votre corps. Si vous utilisez une mauvaise forme lorsque vous faites une série d'exercices de musculation, balancez un club de golf ou lancez une balle de baseball, par exemple, vous risquez de surcharger certains muscles et de provoquer une blessure due au surmenage.

Des blessures de surmenage peuvent survenir à la suite d'erreurs d'entraînement telles que l'accélération trop rapide d'une activité ou l'exercice trop long sans donner suffisamment de temps pour se reposer et récupérer. Ceux-ci peuvent également se produire lors de la pratique d'un seul exercice spécifique dans lequel seuls certains muscles ou os sont utilisés, tels que des tractions répétitives, ou avec une spécialisation sportive où un seul sport est pratiqué toute l'année. Une mauvaise technique peut également jouer un rôle dans les blessures de surmenage dans lesquelles le tissu peut être surchargé de manière répétitive de manière inappropriée.

Différentes définitions du surmenage ont créé des limites qui séparent la recherche épidémiologique et la pratique clinique. Comme décrit dans la revue systématique par Roos et Marshall, les cliniciens ont rapporté le terme surmenage comme un mécanisme de blessure, une catégorie de diagnostic, ou à la fois un mécanisme de blessure et une catégorie de diagnostic dans la surveillance des blessures. Les auteurs ont recommandé aux cliniciens et aux chercheurs d'utiliser le terme de surmenage uniquement lorsqu'ils se réfèrent à un « mécanisme d'apparition progressive avec une pathogenèse sous-jacente de microtraumatismes répétés ».

Une compréhension approfondie des RRMI les plus fréquentes est une étape essentielle dans l'élaboration de programmes efficaces de prévention des blessures et de stratégies d'intervention de réadaptation qui peuvent réduire l'incidence et la prévalence élevées des RRMI, respectivement. 70 % de tous les RRMI sont liés à un surmenage. De plus, les blessures signalées se situent principalement au niveau ou en dessous du genou. Cela peut être dû au fait que, lors d'une course normale, la propulsion est principalement générée par le bas de la jambe, ce qui entraîne une augmentation de la charge biomécanique sur ces structures.

Le genou, la cheville et la partie inférieure de la jambe représentent la proportion la plus élevée d'incidence des blessures, tandis que le genou, la partie inférieure de la jambe et les pieds/orteils présentaient la proportion la plus élevée de prévalence des blessures. La tendinopathie d'Achille (10,3 %), le syndrome de stress tibial médial (9,4 %), le syndrome de douleur fémoro-patellaire (6,3 %), la fasciite plantaire (6,1 %) et les entorses de la cheville (5,8 %) représentent la proportion la plus élevée d'incidence des blessures, tandis que la douleur fémoro-patellaire (16,7 %), le syndrome de stress tibial médial (9,1 %), la fasciite plantaire (7,9 %), le syndrome de la bandelette iliotibiale (7,9 %) et la tendinopathie d'Achille (6,6 %) ont la plus forte proportion de prévalence de blessures. La cheville (34,5 %), le genou (28,1 %) et le bas de la jambe (12,9 %) étaient les 3 sites les plus fréquemment blessés.

Alors que la course à pied continue de gagner en popularité, l'intérêt pour la recherche et l'évaluation de la foulée augmente également. L'augmentation spectaculaire du nombre de coureurs récréatifs et compétitifs au cours des dernières années a des implications évidentes pour les professionnels de la santé. L'une des principales raisons de sa popularité tient à sa simplicité. Cependant, la course à pied comporte également un risque accru de blessures musculo-squelettiques et il est nécessaire de comprendre l'étiologie des blessures afin de les prévenir efficacement.

Le but de notre 1ère était de comparer le modèle de pose du pied de course sur tapis roulant à l'aide d'une plate-forme de mesure de force (incluse directement dans le tapis roulant instrumenté) et de gyroscopes pour d'éventuelles similitudes/différences dans l'analyse de la marche et la cinématique des articulations des jambes. Nous avons émis l'hypothèse que malgré l'utilisation de différents outils pour analyser les modèles de course, nous serions en mesure d'atteindre le même résultat final et la même conclusion. En conclusion, cette étude a trouvé une forte corrélation entre un seul capteur inertiel positionné sur le tibia et un tapis roulant instrumenté pour la classification du modèle de pose du pied. L'utilisation de plusieurs gyroscopes MEMS (pied, tibia et sacrum) pourrait probablement réduire la marge d'erreur.

Pour notre 2ème publication, nous utilisons les composantes harmoniques comme caractéristiques de la foulée en utilisant la décomposition de Fourier de la série chronologique des caractéristiques de la foulée comme base à partir de laquelle extraire les harmoniques fondamentales et d'ordre supérieur. Intuitivement, l'amplitude mesurée à la fréquence fondamentale est une mesure du changement global subi par la caractéristique correspondante, et la phase relative entre différentes séries temporelles est une indication du délai entre les différentes caractéristiques. Le but de notre 2ème recherche était de déterminer la corrélation entre un seul capteur inertiel et une méthode acceptée pour mesurer la foulée spécifiquement le modèle de frappe du pied et si cette corrélation varie avec l'augmentation de la vitesse et si les pieds nus sont des pieds.

Pour les 3ème et 4ème publications : une nouvelle méthodologie pour l'analyse des événements de choc survenant au cours de la procédure expérimentale proposée. Notre approche est basée sur le spectre de réponse aux chocs (SRS), qui est une fonction basée sur la fréquence utilisée pour indiquer l'amplitude des vibrations dues à un choc ou à un événement transitoire. L'objectif principal est d'analyser la capacité du système musculo-squelettique humain à atténuer les contraintes mécaniques résultant de l'effet de fatigue par Shock Responses Spectrum (SRS) des ondes de choc générées par la pose du pied pendant la course. SRS est une fonction basée sur la fréquence utilisée pour indiquer l'amplitude des vibrations dues à un choc ou à un événement transitoire. C'est une méthode d'analyse des systèmes transitoires (chocs) développée par l'industrie militaire pour déterminer comment les équipements réagissent aux sollicitations répétitives. A partir de là, la transposition de cette méthode à l'exécution de l'analyse de la foulée il n'y a qu'un pas.

La course implique des impacts répétés entre le pied et la surface. De tels impacts sont caractérisés par un pic transitoire de la force de réaction au sol (force d'impact), une décélération rapide du membre inférieur (choc d'impact) et l'initiation d'une onde d'accélération et de décélération (onde de choc d'impact) qui se propage à travers le corps. Les charges produites par des impacts répétés ont été liées à des maladies articulaires dégénératives et à des blessures de surmenage sportif, notamment des fractures de stress, des périostites tibiales, de l'arthrose et des douleurs lombaires. Bien que les mécanismes exacts des blessures liées à l'impact soient relativement inconnus, les preuves parfois controversées reliant l'impact et les blessures soient bien documentées. L'une des fonctions importantes du système musculo-squelettique humain est d'atténuer et de dissiper les ondes de choc initiées par le contact du pied au sol. Ces ondes de choc sont déclenchées par la plupart des types de mouvement, comme la marche et la course. La course implique des impacts répétés d'une seule jambe entre le pied et la surface. De tels impacts sont caractérisés par un pic transitoire de la force de réaction au sol (force d'impact), une décélération rapide du membre inférieur (choc d'impact) et l'initiation d'une onde d'accélération et de décélération (onde de choc d'impact) qui se propage à travers le corps.

Les blessures de surmenage en course à pied sont souvent provoquées par la fatigue ou une mauvaise technique, qui se reflètent toutes deux dans la cinématique du coureur. La recherche de pointe sur la cinétique et la cinématique dans le sport utilise des systèmes d'analyse de mouvement qui sont inaccessibles à la plupart des athlètes. Le potentiel des capteurs embarqués pour l'analyse cinétique et cinématique des coureurs est extrêmement pertinent et rentable. Tout au long de nos recherches, nous avons démontré le potentiel des capteurs portables pour l'analyse cinétique et cinématique des coureurs. Nous présentons plusieurs études utilisant des centrales inertielles (IMU) pour l'évaluation du niveau de performance et surveillance de la fatigue. Nous avons extrait de nombreux paramètres de foulée pour les évaluations de performance et de santé. Les capteurs embarqués constituent un outil précieux pour les coureurs, des débutants aux experts, pour l'évaluation de la technique de course

Une approche individualisée d'un athlète est d'un grand intérêt et disponible de nos jours. En effet, la recherche de pointe sur la cinématique dans le sport utilise des systèmes optiques de capture de mouvement inaccessibles à la plupart des athlètes. Avec le développement récent des systèmes microélectromécaniques (MEMS), les centrales inertielles sont devenues largement utilisés dans la recherche sur l'analyse de la foulée en raison de plusieurs facteurs, tels que la facilité d'utilisation et le faible coût. Considérant le fait que chaque individu a une façon unique de courir, les IMU peuvent être appliqués au problème de la reconnaissance de la foulée qui peut être interprétée comme une signature biométrique. Ainsi, la reconnaissance de la foulée par IMU a un grand potentiel pour jouer un rôle important dans de nombreuses applications liées à la santé. Dans ce travail, nous avons démontré le potentiel de la technologie portable pour l'évaluation des paramètres cinématiques en utilisant l'exemple de la course. Nous avons conclu que la technologie portable ouvre des possibilités d'amélioration de la technique et de réduction des risques de blessures à un large éventail d'athlètes. Étant donné que les IMU sont inclus dans les appareils intelligents qui sont aujourd'hui présents partout, la reconnaissance de la foulée par IMU est devenue un domaine de recherche très attractif et émergent qui fournira de nombreuses découvertes intéressantes.

Traditionnellement, la foulée des coureurs était mesurée en laboratoire; cependant, les résultats de ces études peuvent ne pas être généralisables à la course extérieure. De nombreux coureurs s'entraînent dans un environnement extérieur et ne courent pas régulièrement dans un laboratoire ou sous observation directe. Les études en laboratoire n'analysent généralement que 5 à 15 pas de la course donnée d'un sujet en raison de contraintes de stockage et d'analyse des données, mais les individus peuvent faire environ 2 000 pas par mile. En raison des changements potentiels dans la mécanique de course sur de longues distances, l'analyse d'une multitude d'étapes peut être bénéfique.

L'utilité des capteurs embarqués pour mesurer la biomécanique de la foulée est utile à la fois pour les cliniciens travaillant avec des coureurs pour les surveiller sur plusieurs courses et pour les chercheurs étudiant une variété de coureurs. Les capteurs portables peuvent permettre une mesure continue de la mécanique de la foulée sur des milliers de pas pendant la course d'entraînement typique d'un coureur de fond et, par conséquent, peuvent mieux surveiller divers aspects de la biomécanique de la course en capturant des informations sur chaque foulée

Les chaussures connectées intelligentes seront certainement l'avenir de l'analyse de la course en cours d'exécution. Après des années de recherche et de développement, les chaussures du futur deviennent disponibles dans le commerce. Par rapport à d'autres technologies portables, les chaussures de nouvelle génération sont un produit plus difficile à concevoir, développer et produire. Les chaussures intelligentes seront en mesure de fournir des variables pertinentes et de fournir ces informations à votre smartphone ; elles auront besoin de capteurs de pression, d'un gyroscope et d'un accéléromètre MEMS, ainsi que d'une connectivité applicative tout en conservant confort et légèreté.

À bien des égards, les futures chaussures sont déjà là. La technologie portable peut suivre les données, les collecter et les interpréter de manière significative pour améliorer les performances sportives. Cependant, à mesure que les semelles et les chaussures intelligentes deviennent de plus en plus populaires et que leur utilisation se généralise, leurs capacités vont certainement s'améliorer. Des commentaires de coaching en direct seront disponibles et un assistant intelligent vous guidera tout au long de votre course. Les données de course d'un jour pourraient être utilisées pour optimiser une paire de chaussures de course personnalisées imprimées en 3D. Au fur et à mesure que le matériel et les matériaux s'améliorent, il peut même devenir possible d'adapter automatiquement et dynamiquement la semelle et l'amorti aux conditions de course, par exemple en augmentant l'adhérence lors de la course sur un sol instable ou en adoucissant la semelle intérieure sur de l'asphalte dur.

Une approche plus holistique pourrait voir la technologie de suivi appliquée à toutes les chaussures et les données résultantes liées à d'autres informations sur la santé et la forme physique. Cela vous permettrait de voir des schémas et des liens entre les habitudes quotidiennes et les performances sportives, la condition physique générale et les défis spécifiques, peut-être même la santé physique, mentale et le bien-être général. On s'attend à ce que les chaussures de course intelligentes améliorent l'efficacité de la course.

Ainsi, il est dans l'intérêt de chaque coureur d'ajuster périodiquement sa cadence pour voir si cela rend la course plus facile. Cependant, il a toujours été difficile de savoir par où commencer. Les chaussures connectées vous facilitent la tâche en vous proposant une gamme personnalisée à cibler. Bien que la gamme ne soit pas parfaite pour tout le monde, c'est un bon point de départ. La réception de ces données depuis votre chaussure vous permet d'apporter des corrections pour raccourcir votre foulée et augmenter votre cadence. Les commentaires de coaching de l'application vous permettent de savoir comment les modifications que vous avez apportées améliorent votre forme sur de longues périodes et à intervalles.

Les chaussures intelligentes pourront mesurer l'angle d’attaque du pied en plus des recommandations pour accélérer votre cadence, raccourcir votre foulée ou prendre des jours de récupération supplémentaires avant une séance de fractionné. L'une des capacités les plus intéressantes des chaussures intelligentes sera la capacité potentielle d'aider les coureurs à évaluer leur fatigue.

Les accéléromètres sont partout. Ils se trouvent dans nos smartphones, nos ordinateurs portables, dans les voitures et dans nos appareils portables. Les accéléromètres peuvent détecter des changements d'accélération souvent jusqu'à 1000Hz, nous donnant 1000 points de données chaque seconde. En raison des exigences spécifiques d'une grande partie de notre technologie portable, les accéléromètres sont pris en charge par un certain nombre d'autres appareils au sein du système. Les gyroscopes aident à déterminer l'orientation en utilisant la gravité terrestre et sans eux, les accéléromètres ne seraient pas en mesure de fournir des informations précieuses aux entraîneurs et aux praticiens.

Les objets connectés ont connu une croissance exponentielle au cours de la dernière décennie. La technologie embarquée englobe désormais une grande variété de sous-catégories, dont beaucoup ont connu une croissance et des investissements incroyables. Les appareils de sport portables deviennent plus petits, plus puissants et plus économiques. La nécessité de collecter des données sur le terrain plutôt qu'en laboratoire a poussé les entreprises de technologie sportive à innover et à créer des produits « invisibles » pour l'athlète. Alors que nous nous efforçons d'intégrer la technologie portable dans les activités quotidiennes des athlètes d'élite, une nouvelle ère d'appareils semble se profiler à l'horizon. Non seulement les appareils seront plus petits, mais ils seront intégrés dans les vêtements, les chaussures et les équipements de protection. Cela permettra aux scientifiques du sport de maximiser non seulement la quantité et la qualité des données collectées, mais également l'adhésion des athlètes. Cette méthode quantitative de collecte de données fournit la vraie valeur de la performance d'un athlète, sans avoir besoin de commentaires subjectifs de l'athlète lui-même.

Au fur et à mesure que les praticiens progressent dans leur parcours en utilisant de tels systèmes, ils commencent souvent à voir les failles de ce type de données et se tournent vers l'accéléromètre pour des informations plus approfondies. Cela amène le praticien à se pencher davantage sur les accélérations, les décélérations et les caractéristiques de la foulée. L'analyse des caractéristiques de la foulée devient courante à mesure que les praticiens intègrent une « surveillance invisible » pour quantifier la fatigue des athlètes sur le terrain.

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APPENDIX

A1) Preliminary comparison between angular velocity and force measuring treadmill for running foot strike patterns classification

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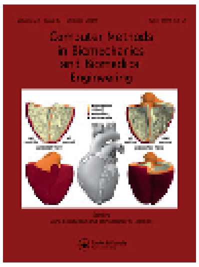

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Preliminary comparison between angular velocity and force measuring treadmill for running foot strike patterns classification

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1. Introduction

As running continues to grow in popularity, so does the interest in research and assessment of running gait. The dramatic increase in the number of recreational and competitive runners in recent years has obvious implications to health professionals.
Current research aims at estimating activity-related parameters. In particular, the assessment of temporal and spatial gait parameters, the detection of gait phases and the discrimination of different gait activities (e.g. walking, jogging, stair climbing) have been investigated. However, gait phase detection and discrimination are investigated separately.
Our goal is to compare treadmill running foot strike pattern using a force measuring platform (included directly in the instrumented treadmill) and gyroscopes for possible similarities/differences in gait analysis and leg joint kinematics. We hypothesized that despite using different tools to analyze running patterns we would be able to reach the same end result and conclusion.

2. Methods

2.1 Participants

Twelve male runners who were free from musculoskeletal injury volunteered to take part in this study. The runners had a mean age of 30.3 (±4.9) years, stature 178.3 (±5.7) cm, and body mass 77.7 (±8.5) kg respectively. Participants were active recreational runners engaging in training at least two times per week whilst completing a minimum of 20km per week and had previous experience of treadmill running. Participants encompassed a range of foot strike characteristics.

An *a priori* power analysis was conducted using the Hopkins method based on a moderate effect size and a power measure of 80%, which suggested that 12 subjects were adequate for the design. The study was approved by the University of Reims Champagne-Ardennes and the Alleray-Labrouste clinic. All participants provided a written informed consent.

2.2 Material

A commercial Force Distribution Measurement Treadmill (FDM-T, Zebris®, Germany) with an appropriate conveyor belt (1500mm long × 500mm wide) was used. The driving system provides a range from 0 to 22 kmh−1 (by minimum increments of 0.1km/h). For a treading area of 150 x 50 cm the sensor unit has more than 5,000 capacitive pressure/force sensors. Using a specific ergometer treadmill design the movement of the treadmill is compensated so that completely stable gait and roll off patterns can be analyzed.

One inertial sensor (HIKOB FOX®) was used which contained a tri-axial accelerometer with a sample rate ranging from 0Hz to 1.3 kHz. Research has indicated that the distal anteromedial portion of the tibia is chosen as a placement site to reduce the effect of angular acceleration and rotational movement. The MEMS gyroscope was placed on the skin using adhesive strapping band.

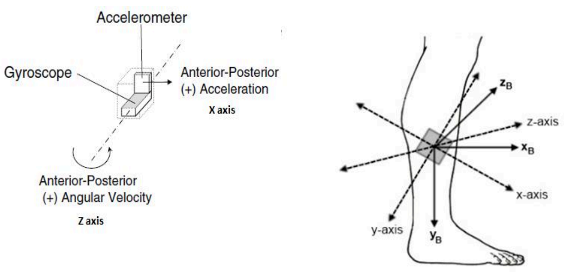

Figure 39: MEMS placement

2.3 Procedures

Participants were given a 5 minutes' adaptation period, in which participants ran at the determined velocity of 3.3 m/s. The treadmill was then stopped for 30 s and participants dismounted the treadmill before mounting the treadmill for data recording. When participants indicated that they were ready to begin, the treadmill was restarted and the velocity of the belt was gradually increased until the speed reached 2.2 m/s for a first recording of 30s then the speed was again increased to reach 3.3m/s for a second recording of 30s. Each participant performed a trial shod and a trial barefoot. During both trials the accelerometer placed on the distal anteromedial portion of the tibia collects the data on the 3 axis acceleration measure.

3. Results

For this comparative study of the qualitative and quantitative data we have used Support Vector Machine model using kernel method. This class of algorithms is particularly used for pattern analysis, it enables us to study and compare general types of relations within the foot strike patterns. We used the foot strike classification as follow: rearfoot strike (RFS), midfoot strike (MFS) and forefoot strike (FFS).

A single gyroscope on the shank segment can provide information on segment inclination range, cadence, number of steps, and an estimation of stride length and walking speed. It has not yet been used to identify foot strike patterns. Joint angles are commonly used in gait analysis and can be derived by the integration of angular acceleration or angular velocity. However, data obtained from integration can be distorted by offsets or any drifts. According to Tong et al. (1999), gyroscopes can theoretically be used to calculate the segment inclination and the relative joint angle. Using gyroscopes, it is possible to discriminate different activities and in addition provide angular information. A system using a gyroscope on the shank provides rich information for gait analysis. A high pass filter to correct any drift and offset, inclination derived from the gyroscope signal is used to calculate the segment inclination range, cadence, number of steps taken and foot strike patterns. The pattern on the shank showed two minima, one occurs when foot flat and the other occurs when toe off. There were also two peaks in this pattern. The large one occurred during mid-swing and the small one occurred at heel off. This pattern provides information that can be used to identify different gait events and may be useful for developing control systems.

		Results	
		RFS	MFS
Réel	RFS	6	4
	MFS	3	23

$$Overall\ Accuracy\ de\ \frac{6+23}{36} = 80.6\%$$

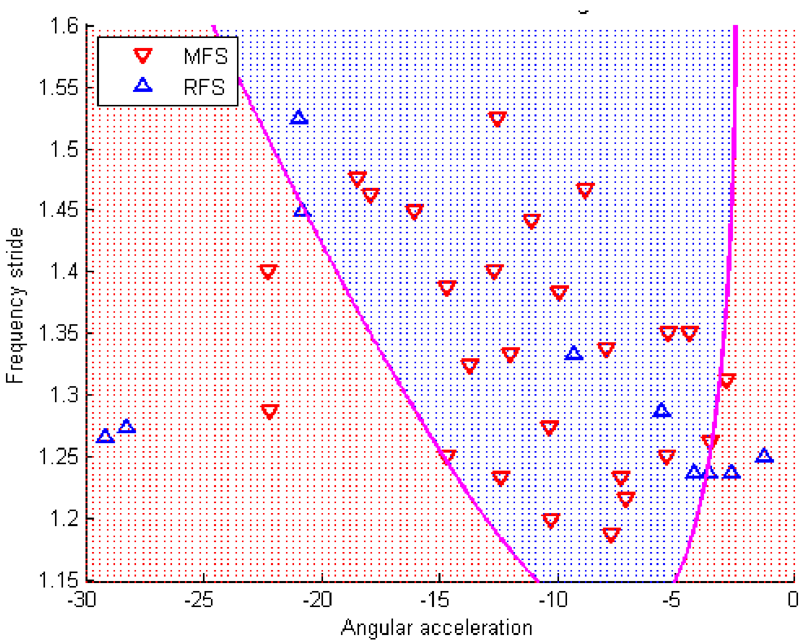

Figure 2: Classification using Kernel method and results correlation

4. Conclusions

The purpose of this research was to determine the agreement between a single inertial sensor and an accepted method to measure running gait specifically foot strike pattern and whether this agreement varies with increasing velocity and whether is shod or barefoot. Agreement for most of the data were 80.6%, coupled with small bias and very large to nearly perfect correlations demonstrate strong agreement between both methods. The 19.4% margin of error could be linked to the placement of the MEMS gyroscope and the way it was attached. Although biomechanical testing has confirmed the validity and reliability of accelerometers in the measurement of accelerations within the frequency and amplitude range of human body motion, evidence indicates that they are sensitive to the site and method of attachment, also faster velocity studies found up to 7% error in step identification.

In conclusion this study found strong correlation between a single inertial sensor positioned on the tibia and an instrumented treadmill for foot strike pattern classification. Using several MEMS gyroscope (foot, shin and sacrum) could probably reduce the margin of error.

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A2) Harmonic decomposition and analysis of running gait

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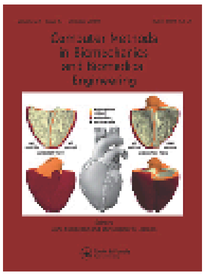

Computer Methods in Biomechanics and Biomedical Engineering

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Harmonic decomposition and analysis of running gait

D. Benjamin, B. Abbes, S. Odof, J. B. Nolot, F. Fourchet, X. Chiementin & R. Taiar

1. Introduction

Running is a mostly periodic activity. Hence, it seems natural to use the harmonic components as gait features. We use the Fourier decomposition of the time series of the gait features as the basis from which to extract the fundamental and higher order harmonics. Intuitively, the magnitude measured at the fundamental frequency is a measure of the overall change undergone by the corresponding feature, and the relative phase between different time series is an indication of the time delay between the different features. The higher harmonics measured with respect to the fundamental harmonic describe the nonsinusoidal but still periodic trajectory that a feature undergoes.

While the fundamental harmonic components capture the majority of the information, they do not capture the subtle variations in the dynamics of different features. Higher harmonics are needed to capture these variations. Intuitively, the magnitude of the fundamental frequency together with the magnitude of the second harmonic and the phase of the second harmonic relative to the fundamental frequency provide a translation independent description of a signal that contains only first and second harmonics. We do not look beyond the second harmonic because the sampling rate and the amount of noise in the gait analysis makes higher harmonic components unstable.

2. Methods

2.1 Participants

Twelve male runners who were free from musculoskeletal injury volunteered to take part in this study. The runners had a mean age of 30.3 (±4.9) years, stature 178.3 (±5.7) cm, and body mass 77.7 (±8.5) kg respectively. Participants were active recreational runners engaging in training at least two times per week whilst completing a minimum of 20km per week and had previous experience of treadmill running. Participants encompassed a range of foot strike characteristics.

An *a priori* power analysis was conducted using the Hopkins method based on a moderate effect size and a power measure of 80%, which suggested that 12 subjects were adequate for the design. The study was approved by the University of Reims Champagne-Ardennes and the Alleray-Labrouste clinic. All participants provided a written informed consent.

2.2 Material

A commercial Force Distribution Measurement Treadmill (FDM-T, Zebris®, Germany) with an appropriate conveyor belt (1500mm long × 500mm wide) was used. The driving system provides a range from 0 to 22 kmh−1 (by minimum increments of 0.1km/h). For a treading area of 150 x 50 cm the sensor unit has more than 5,000 capacitive pressure/force sensors.

One inertial sensor (HIKOB FOX®) was used which contained a tri-axial accelerometer with a sample rate ranging from 0Hz to 1.3 kHz. Research has indicated that the tibia is chosen as a placement site to reduce the effect of angular acceleration and rotational movement. The MEMS gyroscope was placed on the skin using adhesive strapping band.

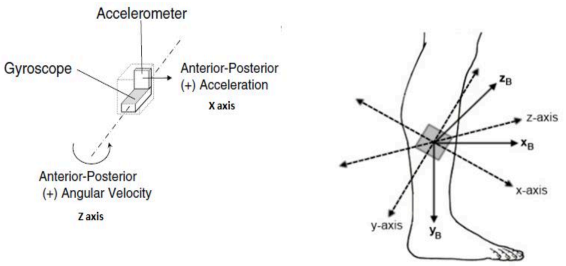


Figure 40: MEMS placement

2.3 Procedures

Participants were given a 5 minutes' adaptation period, in which participants ran at the determined velocity of 3.3 m/s. The treadmill was then stopped for 30 s and participants dismounted the treadmill before mounting the treadmill for data recording in accordance with the recommendation of Alton et al. (1998). When participants indicated that they were ready to begin, the treadmill was restarted and the velocity of the belt was gradually increased until the speed reached 2.2 m/s (BF 8 and C8) for a first recording of 30s then the speed was again increased to reach 3.3m/s (BF 11 and C11) for a second recording of 30s. Each participant performed a trial shod (C) and a trial barefoot (BF). During both trials the accelerometer placed on the distal anteromedial portion of the tibia collects the data on the 3 axis acceleration measure.

3. Results and Discussion

For this comparative study of the qualitative and quantitative data we have used at first a frequency analysis enabling us to determine the fundamental frequency.

We then performed the calculation of Grms. The metric of Grms is typically used to specify and compare the energy in repetitive shock vibration systems. Repetitive shock (RS) vibration systems produce a continuously varying pseudorandom broad spectrum vibration. The root mean square (rms) value of this signal can be calculated by squaring the magnitude of the signal at every point, finding the average (mean) value of the squared magnitude, then taking the square root of the average value. The resulting number is the Grms metric.

	BF 8	BF 11	C8	C11
Fundamental	1.45	1.53	1.47	1.54
Grms	7.10	11.26	11.31	13.06

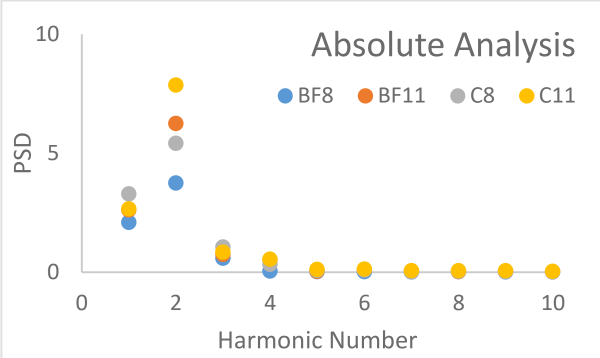
Figure 2: Fundamental and Grms calculation

The Grms signal is described as a time domain measurement, it is typically thought of as a frequency domain measurement taken from the Power Spectrum curve. When Grms is calculated using Power Spectrum information it is often thought of as the area under the curve of the Power Spectrum display. More accurately, it is the square root of the integral of the Power Spectrum.

$$\text{PSD}: \ PSD(\upsilon) = \frac{1}{T}\left|\int_{-\infty}^{+\infty} a(t)\, e^{-2i\pi\upsilon t} dt\right|^2$$

n frequency in Hz and T duration of the signal

In order to compare the content of the signals in frequency quality and to overcome the fundamental differences in frequency, we normalized each PSD on the fundamental frequency to enable us to compare the harmonic's content.



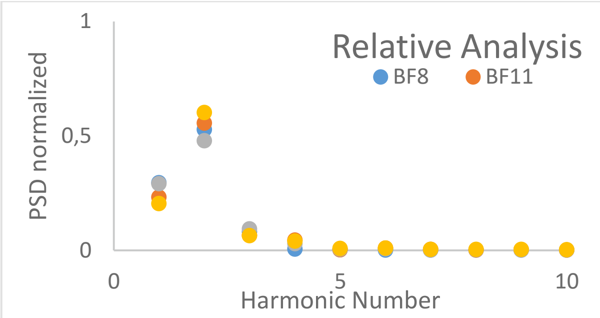


Figure 3: Harmonic analysis

4. Conclusion

The area of the peaks is represented as a function of the harmonic number. The area is either normalized to the complete area (relative analysis) or not (absolute analysis). There is a clear difference between the shod running git analysis (C8 and C11) and the barefoot running gait analysis (B8 and B11). There is a clear rise for the shod analysis on the 2nd harmonic. The energy at 3.3m/s is also much more important. The purpose of this research was to determine the agreement between a single inertial sensor and an accepted method to measure running gait specifically foot strike pattern and whether this agreement varies with increasing velocity and whether is shod or barefoot

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A3) Shock response spectrum analysis in running performance

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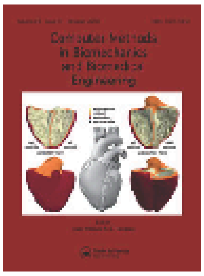

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Shock response spectrum analysis in running performance

D. Benjamin, S. Odof, B. Abbes, J.B. Nolot, D. Erre, F. Fourchet & R. Taiar

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1. Introduction

Running involves repeated impacts between the foot and the surface. Such impacts are characterized by a transient peak in the ground reaction force (impact force), rapid deceleration of the lower extremity (impact shock), and the initiation of a wave of acceleration and deceleration (impact shock wave) that is propagated through the body. The loads produced by repeated impacts have been linked to degenerative joint diseases and athletic overuse injuries including for example stress fractures, shin splints,

osteoarthritis and lower back pain. Although the exact mechanisms of impact related injury are relatively unknown and controversial evidence linking impact and injuries is well documented.
The purpose of this study was to determine the effects of increasing impact shock levels on the spectral characteristics in the body during treadmill running. The main aim is to analyse the ability of the human musculoskeletal system to attenuate the mechanical stresses resulting from the fatigue effect by Shock Responses Spectrum (SRS) of the foot strike–generated shock waves during running. SRS is a frequency-based function that is used to indicate the magnitude of vibration due to a shock or transient event. It's an analysis method of transitional systems (shocks) developed by the military industry to determine how equipment reacts to repetitive stress. From there on, transposing this method to running gait analysis there is only one step.

2. Methods

2.1 Procedures

Twelve male runners who were free from musculoskeletal injury volunteered to take part in this study. The runners had a mean age of 30.3 (±4.9) years, stature 178.3 (±5.7) cm, and body mass 77.7 (±8.5) kg respectively.
One inertial sensor (HIKOB FOX®) was used and placed on the distal anteromedial portion of the tibia to collect data on the 3-axis acceleration measure. The MEMS accelerometer was placed on the skin using adhesive strapping band, such an attachment, as proven in earlier works can accurately measure shock wave amplitude according to Winslow et al. (1989). The accelerometer was aligned along the longitudinal axis of the tibia to provide axial components of tibial acceleration. Participants were given a 5 minutes' adaptation period, in which participants ran at the determined velocity of 3.3 m/s. The treadmill was then stopped for 30 s and participants dismounted the treadmill before mounting the treadmill for data recording. When participants indicated that they were ready to begin, the treadmill was restarted and the velocity of the belt was gradually increased until the speed reached 2.2 m/s for a first recording of 30s then the speed was again increased to reach 3.3m/s for a second recording of 30s. To minimize possible gait modifications, the subjects were not aware of when exactly the data were acquired.

2.2 Spectral analysis of the impact shock wave

Spectral analysis is commonly used to study the structure of composite wave forms such as the impact shock waves. The primary tool of spectral analysis is the Fast Fourier Transformation (FFT) that enables us to determine the runner's natural frequency (Figure 1) which corresponds to the peak of the Power Spectral Density.

$$PSD = \frac{1}{N}\left|\int_{-\infty}^{+\infty} a(t)e^{-j2\pi\upsilon t}\,dt\right|^2$$

Power Spectral Density (PSD) provides a convenient method of separating different frequency components in the impact shock wave such as acceleration moments due to impact shock.
The runner's natural frequency then enables us to determine his time step and his running pattern. We then carry out the cross-correlation CC (τ) between the runner's pattern and the recording's duration a(t).

$$CC(\tau) = \int_{-\infty}^{+\infty} a(t)pattern(t+\tau)dt$$

We observe that at each step, the convolution is maximum. For each maximum value of CC(□) we calculate the SRS of each step and of the entire signal using Lalanne (2009) algorithm for shock analysis. SRS enables us to determine the maximum acceleration a system will undergo when one knows the natural frequency f0 and the quality factor Q for each possible natural frequency. SRS can also be calculated for the entire duration of a recording (Figure 2). We then observe the peaks at the fundamental and harmonic frequencies of the recorded signal. In this context SRS combines both the notion of transfer function and response to transient regimes.
Intra comparison of the SRS offers a lot more finesse to the analysis since the frequency is also taken into account. The aggressiveness of a running step is not only due to the value of the maximum acceleration, but also to the general shape of the movement, only the SRS allows this to be taken into account in the analysis.

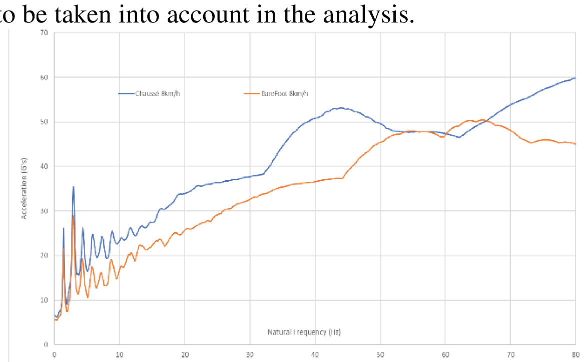

Figure 41: SRS Comparison shod/barefoot at 8km/H

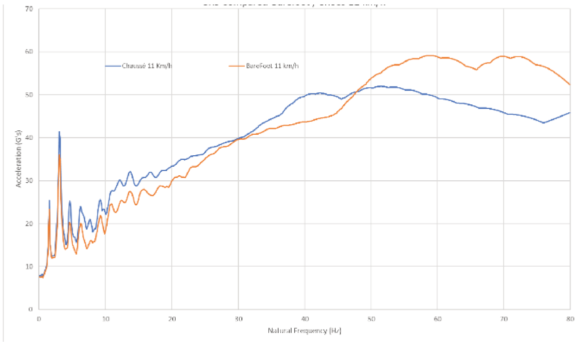

Figure 42 : SRS Comparison shod/barefoot at 8km/H

3. Results and discussion

A goal of the present study was to analyze the effect of fatigue through SRS on the ability of the human musculoskeletal system to attenuate foot strike–generated shock waves. Fatigue hampers the ability of the human musculoskeletal system to protect itself from overloading due to foot strike–generated shock waves. Loss of protection may be manifested in an increased shock wave amplitude as measured on the tibial tuberosity. The results were obtained through this protocol while subjects ran on a motor-driven treadmill. While such a setup simplified data acquisition, the locomotion patterns obtained may differ from patterns for overground running. The runners in the present study were constrained to run at a constant speed, regardless of whether they were fatigued or not. Since one of the main purposes of studying SRS was to introduce the correlation between running fatigue and lower limb injuries it is important to note that this may slightly differ in overground running: When fatigue begins, runners may slow down as a protective means. The result could be moving away from the state of fatigue, in which case the acceleration data might not increase. According to the results in this study, for the acceleration data to increase, fatigue should be present. Thus, the results of this study can be extrapolated to overground running if fatigue indeed prevails. Also, most running injuries in overground running are lower extremity injuries, with a predominance of knee injuries and our results indicating an increase in acceleration data in the tibial tuberosity therefore tend to support this extrapolation. Previous studies have shown that the loading rate of the lower limb is directly and highly correlated with running speed, and the vertical impact force increased with increasing running velocity.

Muscles activation lowers the bending stress on bone and attenuate the peak dynamic loads that can damage musculoskeletal tissues. Previous studies have suggested that the fatigued muscles cannot support “optimal” running and they also suggested that fatigue of the runner may lead to modification of landing phase mechanics. It was also found that the transfer of mechanical energy between the eccentric and concentric phases is drastically reduced during muscle fatigue. Such changes may be involved in the development of injuries.

4. Conclusions

We may conclude that the musculoskeletal system becomes less capable of handling foot strike–induced shock waves when the muscles are significantly fatigued. Understanding the influence of SRS on fatigue and on the magnitude of dynamic loading on the human musculoskeletal system will allow the development of proper training procedures and exercises, and will reduce damage to the musculoskeletal tissues.

A4) Shock Response Spectrum Analysis of Fatigued Runners

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Article

Shock Response Spectrum Analysis of Fatigued Runners

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Abstract: The purpose of this study was to determine the effect of fatigue on impact shock wave attenuation and assess how human biomechanics relate to shock attenuation during running. In this paper, we propose a new methodology for the analysis of shock events occurring during the proposed experimental procedure. Our approach is based on the Shock Response Spectrum (SRS), which is a frequency-based function that is used to indicate the magnitude of vibration due to a shock or a transient event. Five high level CrossFit athletes who ran at least 3 times per week and who were free from musculoskeletal injury volunteered to take part in this study. Two Micromachined microelectromechanical systems (MEMS) accelerometers (RunScribe®) were used for this experiment. The two RunScribe pods were mounted on top of the foot in the shoelaces. All five athletes performed 3 maximum intensity runs: the 1st run was done after brief warmup with no prior exercise, then the 2nd and the 3rd run were performed in a fatigued state. Prior to the 2nd and the 3rd run the athletes were asked to perform at maximum intensity for two minutes on an Assault AirBike to tire them. For all five athletes there was a direct correlation between fatigue and an increase in the aggressiveness of the SRS. We noticed that for all five athletes for the 3rd run the average SRS peaks were significantly higher than for the 1st run and 2nd run ($p < 0.01$) at the same natural frequency of the athlete. This confirms our hypothesis that fatigue causes a decrease in the shock attenuation capacity of the musculoskeletal system thus potentially involving a higher risk of overuse injury.

Keywords: Shock Response Spectrum; Fatigue; Injuries; Gait Analysis; Micromachined microelectromechanical systems (MEMS) accelerometer.

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1. Introduction

Running is the exercise of choice for millions of people all over the world and across the age spectrum. One of the main reasons for its popularity stems from its simplicity. However, running also carries the risk of increased musculoskeletal injuries and there is a need to understand the aetiology of injury in order to efficiently prevent it [1]. One of the important functions of the human musculoskeletal system is to attenuate and dissipate shock waves initiated with foot ground contact [2]. Those shock waves are initiated by most types of motion such as walking and running. The demarcation between walking and running occurs when periods of double support during the stance phase of the gait cycle (both feet are simultaneously in contact with the ground) give way to two periods of double float at the beginning and the end of the swing phase of gait (neither foot is touching the ground) [3]. Generally, as speed increases further, initial contact changes from being on the hindfoot to the forefoot.

Running involves repeated single-leg impacts between the foot and the surface. Such impacts are characterized by a transient peak in the ground reaction force (impact force), rapid deceleration of the lower extremity (impact shock), and the initiation of a wave of acceleration and deceleration (impact shock wave) that is propagated through the body [4].

The impact shock wave experienced by the body due to landings must be attenuated by several structures and mechanisms in the body including bone, synovial fluids, cartilage, soft tissues, joint kinematics and muscular activity. Passively, shock attenuation is achieved by soft tissues and bone. Actively, shock attenuation is achieved through eccentric muscle action [5]. This active mechanism is thought to be far more significant than the passive mechanism in attenuating shock. Since muscles are thought to play a primary role in energy and shock absorption during landing, it has been hypothesized that reduced muscular function, through fatigue, decreases the shock absorbing capacity of the body and subsequently can lead to an increased chance of injury [6]. Fatigue has been defined as any reduction in the force generating capacity of the total neuromuscular system regardless of the force required in any given situation [7].

The loads produced by repeated impacts have been linked to degenerative joint diseases and athletic overuse injuries including for example stress fractures, shin splints, osteoarthritis and lower back pain. Although the exact mechanisms of impact related injury are relatively unknown and controversial evidence linking impact, fatigue and injuries is well documented [8–11–27–32–34].

In this paper, we propose a new methodology for the analysis of shock events occurring during the proposed experimental procedure. Our approach is based on the Shock Response Spectrum (SRS) [12], which is a frequency-based function that is used to indicate the magnitude of vibration due to a shock or a transient event [13]. The main aim is to analyze the ability of the human musculoskeletal system to attenuate the mechanical stresses resulting from the fatigue effect by Shock Responses Spectrum (SRS) of the foot strike–generated shock waves during running. Most of previous studies focused on shocks/impacts, ground force reaction or spectral or vertical impact load rate. Using SRS as a measurement in running gait analysis has never been studied as off today. This innovative approach could pave the way to a whole new way of assessing a runner's gait pattern using smart connected shoes.

The purpose of this study was to determine the effect of fatigue on impact shock wave attenuation and assess how human biomechanics relate to shock attenuation during running. It was hypothesized that fatigue would cause a decrease in the shock attenuation capacity of the musculoskeletal system thus potentially involving a higher risk of overuse injury.

2. Materials and Methods

2.1. Procedures

Five high level CrossFit athletes who ran at least 3 times per week and who were free from musculoskeletal injury volunteered to take part in this study. The runners had a mean age of 26.4 (±3.9) years, stature 182.3 (±5.7) cm, and body mass 81.7 (±8.5) kg respectively. The study was conducted in accordance with the Helsinki Declaration on human experimentation stated in compliance with the 1964 Helsinki Declaration and its later amendments. Every participant provided written consent after information was given on the aim, protocol, and methodology of the study. The original study was approved by the Medical and Ethical Board of the Centre Luxembourg (protocol code LUX_2021_0308_CLAB and date of approval of 3rd of August 2021). Two Micromachined microelectromechanical systems (MEMS) accelerometers (RunScribe®) were used for this experiment. The two RunScribe pods were mounted on top of the foot in the shoelaces (Figure 1).

Figure 1. MEMS accelerometers placement.

The RunScribe pods encompass 9-Axis Motion Tracking which combine a 3-axis gyroscope, 3-axis accelerometer, and 3-axis compass in the same device together with an onboard Digital Motion Processor. This enables us also to measure at a 500 Hz sampling rate: Efficiency (Stride Rate, Contact Time, Flight Ratio), Motion (Footstrike Type, Pronation, Pronation Velocity), Shock (Impact Gs, Braking Gs), Symmetry, and Power.

After a warmup, participants were asked to perform a first 800 m run at maximum intensity. Right after the first run they jumped on an Assault AirBike (Rogue, Columbus, Ohio, USA) (Figure 2) where they were asked to perform at maximum intensity for 2 minutes. They dismounted the Assault AirBike and they were then asked again to perform a second 800 m run at maximum intensity. The same protocol was then repeated with another 2 minutes on the Assault AirBike then a third run at maximum intensity. The RunScribe pods were turned off during the Assault AirBike sessions and were only recording the three 800 m run intervals.

Figure 2. Assault AirBike.

The Assault AirBike, also known as "the Devil's tricycle" was used to induce fatigue because they procure a unique and extremely challenging effort. It is considered among the crossfit community as the most dreaded but most effective tool for HIIT (High Intensity Interval Training) and metabolical conditioning.

2.2. Shock Response Spectrum calculation

Spectral analysis is commonly used to study the structure of composite wave forms such as the impact shock waves. The primary tool of spectral analysis is the Fast Fourier Transformation (FFT) that enables us to determine the runner's natural frequency [14] which corresponds to the peak of the Power Spectral Density (Figure 3):

$$PSD = \frac{1}{N}\left|\int_{-\infty}^{+\infty} a(t)e^{-j2\pi ft}\,dt\right|^2 \quad (1)$$

where N is the number of points of the recording, $a(t)$ is the acceleration modulus, f is the frequency and t is the time.

Power Spectral Density (PSD) provides a convenient method of separating different frequency components in the impact shock wave such as acceleration moments due to impact shock [15].

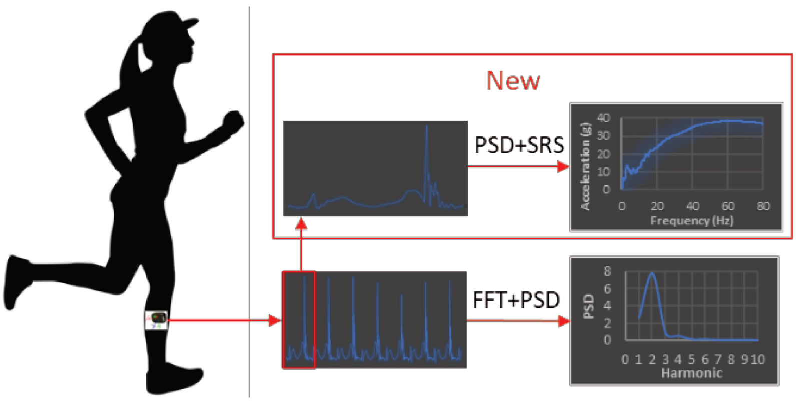

Figure 3. SRS *vs.* PSD running analysis.

In this paper, we propose a new methodology for the analysis of shock events occurring during the proposed experimental procedure. Our approach is based on the Shock Response Spectrum (SRS), which is a frequency-based function that is used to indicate the magnitude of vibration due to a shock or a transient event. The following procedure consisting in several step is adopted in present study:

- **Step 1:**

The acceleration modulus $a(t)$ is extracted from the recording. Figure 4 illustates the acceleration modulus results for one CrossFit athlete.

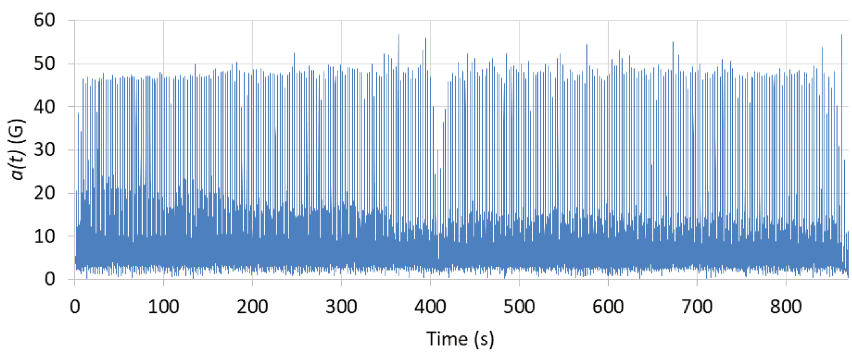

Figure 4. Acceleration modulus $a(t)$ for one CrossFit athlete.

- **Step 2:**

The power spectral density (PSD) given in Equation (1) is then calculated using a Fast Fourier Transform (FFT). This calculation allows us to determine the fundamental frequency of the runner f_0 corresponding to the position of the largest peak of the PSD. The inverse of this frequency gives the time period of the runner's step as: $T = 1/f_0$. The proposed algorithm extracts automatically the "first" step from the entire signal, and thus defines the "pattern" of the runner as shown in Figure 5.

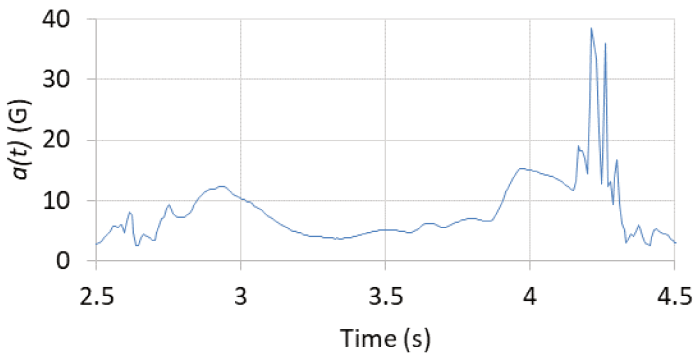

Figure 5. Pattern of one CrossFit athlete.

- **Step 3:**

We then carry out the cross-correlation $CC(\tau)$ between the runner's pattern and the recording's duration $a(t)$:

$$CC(\tau) = \int_{-\infty}^{+\infty} a(t) pattern(t + \tau) dt \tag{2}$$

We observe that at each step, the convolution is maximum. For each maximum value of $CC(\tau)$ we calculate the SRS of each step and of the entire signal as explained in the next step.

- **Step 4:**

The calculation of the SRS is based on the acceleration time history. It applies an acceleration time history as a common base excitation ($\ddot{y}$) to an array of single-degree-of-freedom (SDOF) systems composed of spring (k_i), mass (m_i) and damper (d_i), as depicted in Figure 6.

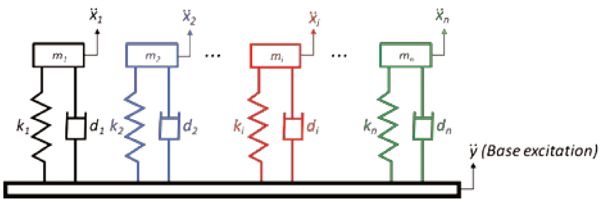

Figure 6. SRS model.

$\ddot{x}_i$ is the absolute response of each system to the input $\ddot{y}$. This can be determined by applying the Newton's law to a free-body diagram of an individual system, as shown in Figure 7.

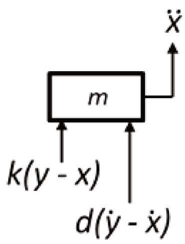

Figure 7. Free-body diagram of an individual system.

The force balance yields the following governing differential equation of motion:

$$m\ddot{x} + d\dot{x} + kx = d\dot{y} + ky \tag{3}$$

By defining the relative displacement $z = x - y$, Equation (3) can be rewritten as:

$$\ddot{z} + 2\xi\omega\dot{z} + \omega^2 z = -\ddot{y} \tag{4}$$

where $\omega_0 = {}^{k}/_{m}$ is the natural frequency in radians per second and $\xi = {}^{d}/_{(2\omega_0 m)}$ is the damping ratio. Moreover, ξ is usually represented by the amplification factor $Q = {}^{1}/_{(2\xi)}$.

Since the base excitation $\ddot{y}$ is an arbitrary function of time, Equation (4) does not have a closed-form solution. To calculate the SRS of each step and of the entire signal, we have used the algorithm for calculation of the SRS proposed in [13]. SRS enables us to determine the maximum acceleration a system will undergo when one knows the natural frequency f_0 and the quality factor Q for each possible natural frequency. In this study a relative damping of 5% is used resulting in $Q = 10$. SRS can also be calculated for the entire duration of a recording. We then observe the peaks at the fundamental and harmonic frequencies of the recorded signal [16]. In this context SRS combines both the notion of transfer function and response to transient regimes.

Intra comparison of the SRS offers a lot more finesse to the analysis since the frequency is also taken into account. The aggressiveness of a running step is not only due to the value of the maximum acceleration, but also to the general shape of the movement, only the SRS allows this to be taken into account in the analysis.

Figure 8 gives the general workflow for SRS determination.

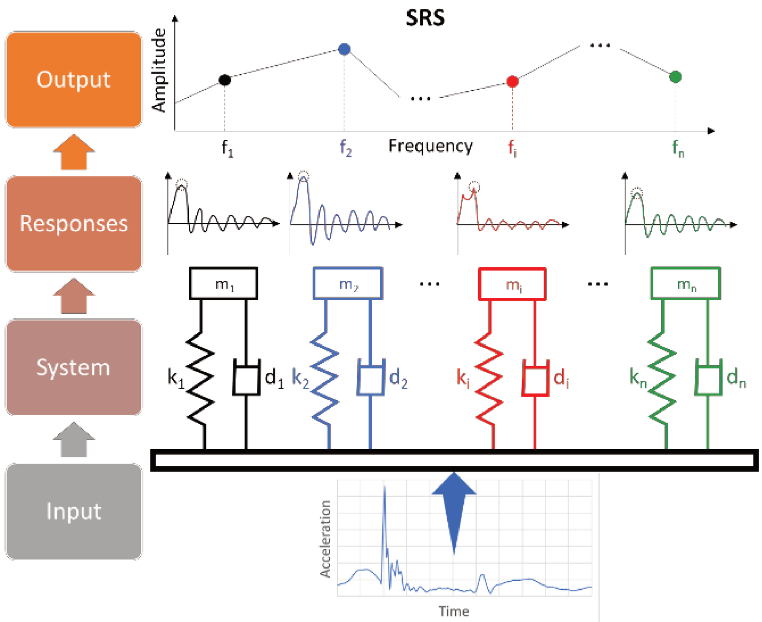

Figure 8. Workflow for SRS determination.

3. Results

A goal of the present study was to analyze the effect of fatigue through SRS on the ability of the human musculoskeletal system to attenuate foot strike–generated shock waves. The results of this study suggest that, for the analysis of impact shock during running, the different components of the acceleration signal can be distinguished in the frequency domain by means of spectral analysis as shown in Figure 9.

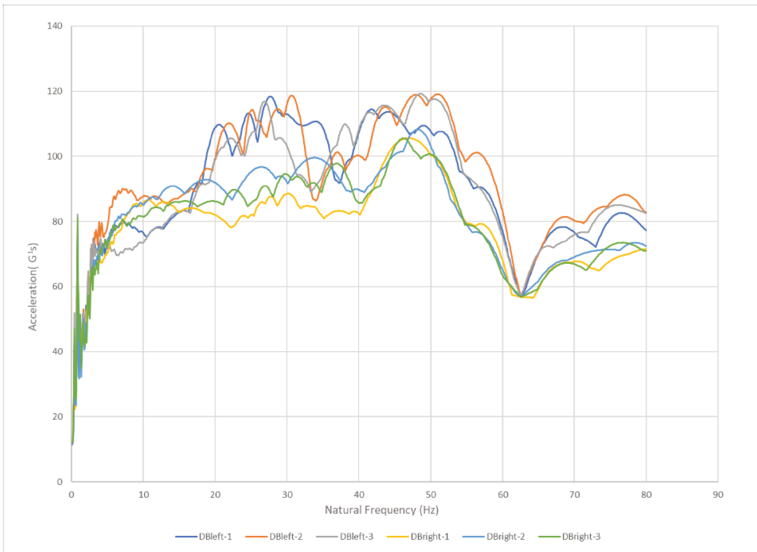

Figure 9. Example of SRS results for one CrossFit athlete extracted for 3 runs on both feet.

The main advantage of spectral analysis over time-domain analysis of the impact shock wave is the ability to separate spectral peaks from the rest of the data. Since the motion, impact, and resonant components of the acceleration signal have different fundamental frequencies: they produce peaks at different points in the power spectrum [12].

The hypothesis is that fatigue hampers the ability of the human musculoskeletal system to protect itself from overloading due to foot strike–generated shock waves, loss of protection may be manifested in an increased shock wave amplitude. For all five athletes there was a direct correlation between fatigue and an increase in the aggressiveness of the SRS as shown in Figure 10. We noticed that for all five athletes for the 3rd run the average SRS peak were significantly higher than for the 1st run and 2nd run ($p < 0.01$) at the same natural frequency of the athlete. This confirms our hypothesis that fatigue causes a decrease in the shock attenuation capacity of the musculoskeletal system thus potentially involving a higher risk of overuse injury.

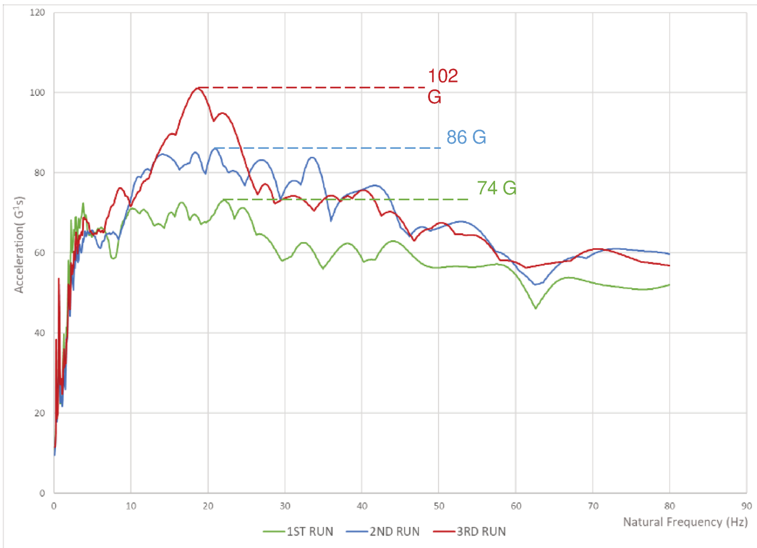

Figure 10. Average SRS peaks for all runners.

When fatigue begins, we could hypothesize that runners will slow down as a protective means. The result could be moving away from the state of fatigue, in which case the acceleration data could have not increase. It was not the case in our study.

Previous studies have shown that the loading rate of the lower limb is directly and highly correlated with running speed, and the vertical impact force increased with increasing running velocity [17].

Muscles activation lowers the bending stress on bone and attenuate the peak dynamic loads that can damage musculoskeletal tissues. Previous studies have suggested that the fatigued muscles cannot support "optimal" running and they also suggested that fatigue of the runner may lead to modification of landing phase mechanics. It was also found that the transfer of mechanical energy between the eccentric and concentric phases is drastically reduced during muscle fatigue. Such changes may be involved in the development of injuries.

4. Discussion

According to the results presented in this study, for the acceleration data to increase, fatigue should be present. We may conclude that the musculoskeletal system becomes less capable of handling foot strike–induced shock waves when the muscles are significantly fatigued. One of the most common running overuse injuries are bone stress fractures (SF) [18–19]. In bones, microcracks are normally present and are thought to be fatigue-related cracks because their numbers increase following repetitive loading. Bone remodeling serves to repair fatigue microcracks. When a bone is loaded repeatedly, resulting in repetitive or cyclic strain, the subsequent accumulation of microdamage is believed to be the threshold of a pathological continuum that is clinically manifested as stress reactions and SF (29). Ultimately, if the activity is not ceased and the bone is not able to self-repair, a complete bone fracture might ensue. Notable, with increasing strains or greater strain rates, the number of loading cycles a bone–29 can withstand before a fatigue failure occurs is reduced [20]. Stress fractures are the clinical manifestation of the accumulation of fatigue damage in bones [21–23]. Although the effect of running and its mechanical strain in bone tissues are well documented, the evidence for SF etiology is less conclusive [21–24–25–34–35]. Nevertheless several researchers reported clear relationships between bone stress related injuries and fatigue. For instance, it is known that the tensile strains on the tensile side of a bending bone is dampened by the contraction of adjacent muscles, aiming at prevent the bone from stress related injury [34–39]. It may then be hypothesized that muscles also play a role of shock absorbers and that consequently muscle fatigue might decrease their absoption properties, resulting in a more aggressive loading rate or loading peak at the bones as fatigue increases [36–37–38].

The obtained results showed that acceleration amplitude steadily increased with fatigue group and that there was a clear association between fatigue and shock waves (as revealed by the SRS). We may then confirm the conclusions of aforementioned studies, that the human musculoskeletal system becomes less capable of single leg strike–induced shock waves absorption when the muscles are significantly fatigued. This condition may promote development of injuries and the present results have a significant implication regarding the etiology of running injuries. Therefore, several recommendations may be effective towards runners community or coaches in order to reduce this stress related injury risk, notably as proposed by the multifactorial model of Brukner and Khan [40].

First it may be advantageous to ensure that the majority of training and exercise are performed in the view to avoid severe fatigue and in line with the load management theory. For instance external parameters must be considered like progressive increment of training loads [30], training surfaces or footwear adaptations [40]. Understanding the influence of SRS on fatigue and on the magnitude of dynamic loading on the human musculoskeletal system will allow the development of proper training procedures and may participate in the reduction of damages to the musculoskeletal tissues.

Secondly and directly in accordance with present research purpose, lower limb muscles resistance to fatigue is a major component of stress related injury prevention in runners. The present outcomes are in line with former findings reporting that fatigue-related imbalance between the plantar flexors and dorsiflexors may compromise the protective action of these muscles on the lower leg bony structures [34]. Here it is plausible that deteriorated properties of the calf muscles due to fatigue may affect the role of these soft tissues to protect the bone from stress injury risk.

Finally, injuries in running are also often provoked by fatigue and improper technique, which are both reflected in the runner's kinematics [27–32–33]. A gait retraining approach has been proposed by several researchers through a modest increase in step rate or a transition from rearfoot to forefoot strike and was considered as effective notably at reducing impact forces and vertical load rate and then at preventing running-related bone stress injuries [31–32].

An individualized approach is nevertheless of high interest and most likely available nowadays. Indeed state of the art research on kinematics in sports is using optical motion capture systems that are inaccessible to most athletes. With the recent development of microelectromechanical systems (MEMS), inertial sensors have become widely used in the research of wearable running gait analysis [26] due to several factors, such as being easy-to-use and low-cost. Considering the fact that each individual has a unique way of running, inertial sensors can be applied to the problem of gait recognition where assessed gait can be interpreted as a biometric signature. Thus, inertial sensor-based gait recognition has a great potential to play an important role in many health-related applications. In this work, we demonstrated the potential of wearable technology for assessment of kinematic parameters using the example of running. We concluded that wearable technology opens possibilities for technique improvement and injury risk reduction to a wide spectrum of athletes. Since inertial sensors are included in smart devices that are nowadays present at every step, inertial sensor-based gait recognition has become very attractive and emerging field of research that will provide many interesting discoveries.

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Institutional Review Board Statement: The study was conducted according to the guidelines of the Declaration of Helsinki and approved by the Ethical Review Board of the University of Reims Champagne-Ardenne and the medical board of the Centre Luxembourg.

Informed Consent Statement: Informed consent was obtained from all subjects involved in the study.

Data Availability Statement: The data presented in this study are available on request from the corresponding author. The data are not publicly available due to ethical reasons.

Conflicts of Interest: The authors declare no conflict of interest.

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A5) Shock Response Spectrum algorithm

```
function TForm1.spectre_de_choc(npts_signal: integer; w0, surtension,
  dt: extended; xpp: array of extended): TSRCData;
var
i: integer;
a,a1,a2,b,b1,b2,c,c1,c2,d,d2,e,s,u,v,wdt,w02,w02dt :extended;
p1d,p2d,p1a,p2a,pd,pa,wtd,wta,sd,cd,ud,vd,ed,sa,ca,ua,va,ea : extended;
  psi: extended;
  w: extended;
 srca_prim_min,srca_prim_max: extended;
 srcd_prim_min,srcd_prim_max: extended;
 srca_res_min,srca_res_max: extended;
 srcd_res_min,srcd_res_max: extended;
 reponsed_prim,reponsea_prim: extended;
 deplacement_z: extended;
 vitesse_zp: extended;
 b1a,b2a: extended;
 deplacementd_z: extended;
 vitessed_zp: extended;
 deplacementa_z: extended;
 vitessea_zp: extended;
 srcd_res,srca_res: extended;
 srcd_pos,srcd_neg,srcd_maximax,srca_pos,srca_neg,srca_maximax: extended;
begin
//initialisation et preparation des calculs
psi:=1/2/surtension; //amortissement relatif
w:=w0*sqrt(1-psi*psi); //pulsation propre amortie
d:=2*psi*w0;
d2:=d/2;
wdt:=w*dt;
e:=exp(-d2*dt);
s:=e*sin(wdt);
c:=e*cos(wdt);
u:=w*c-d2*s;
v:=-w*s-d2*c;
w02:=w0*w0;
w02dt:=w02*dt;
  //Calcul du SRC primaire
  //initialisation des parametres
  srca_prim_min:=1e100;
  srca_prim_max:=-1E100;
  srcd_prim_min:=1e100;
  srcd_prim_max:=-1E100;
  deplacement_z:=0;
```

```
vitesse_zp:=0;
//calcul des reponses sup et inf pendant le choc à la frequence f0
for i:=1 to npts_signal-1 do
begin


a:=(xpp[i-1]-xpp[i])/w02dt;
b:=(-xpp[i-1]-d*a)/w02;
c2:=deplacement_z-b;
c1:=(d2*c2+vitesse_zp-a)/w;
deplacement_z:=s*c1+c*c2+a*dt+b;
vitesse_zp:=u*c1+v*c2+a;
reponsed_prim:=-deplacement_z*w02;
reponsea_prim:=-d*vitesse_zp-deplacement_z*w02;
srca_prim_max:=abs(max(srca_prim_max,reponsea_prim));
srca_prim_min:=min(srca_prim_min,reponsea_prim);
srcd_prim_max:=abs(max(srcd_prim_max,reponsed_prim));
srcd_prim_min:=min(srcd_prim_min,reponsed_prim);

end;


//calcul du SRC Residuel
//initialisation des parametres
srca_res_max:=reponsea_prim;
srca_res_min:=reponsea_prim;
srcd_res_max:=reponsed_prim;
srcd_res_min:=reponsed_prim;
c1:=(d2*deplacement_z+vitesse_zp)/w;
c2:=deplacement_z;
a1:=-w*c2-d2*c1;
a2:=w*c1-d2*c2;
p1d:=-a1;
p2d:=a2;
if (p1d=0) then pd:=(PI/2)*(P2d/abs(p2d)) else pd:=ArcTan(p2d/p1D);   //a verifier artan et sgn
if (pd>=0) then wtd:=pd else wtd:=Pi+pd;

//Angle de phase prmier pic acc
b1a:=-w*a2-d2*a1;
b2a:=w*a1-d2*a2;
p1a:=-d*b1a-a1*w02;
p2a:=d*b2a+a2*w02;
if (p1a=0)  then pa:=(Pi/2)*(p2a/Abs(p2a)) else pa:=arctan(p2A/p1a);
if (pa>=0)  then wta:=pa else wta:=Pi+pa;
for i:=1 to 2 do
  begin
  //dep rel residuel
   sd:=sin(wtd);
   cd:=cos(wtd);
   ud:=w*cd-d2*sd;
   vd:=-w*sd-d2*cd;
```

```
     ed:=exp(-d2*wtd/w);
     deplacementd_z:=ed*(sd*c1+cd*c2);
     vitessed_zp:=ed*(ud*c1+vd*c2);
     //Acce absolue residuelle
     sa:=sin(wta);
     ca:=cos(wta);
     ua:=w*ca-d2*sa;
     va:=-w*sa-d2*ca;
     ea:=exp(-d2*wta/w);
     deplacementa_z:=ea*(sa*c1+ca*c2);
     vitessea_zp:=ea*(ua*c1+va*c2);
     //src residuel
     srcd_res:=-deplacementd_z*w02;
     srca_res:=-d*vitessea_zp-deplacementa_z*w02;
     srcd_res_max:=max(srcd_res_max,srcd_res);
     srcd_res_min:=min(srcd_res_min,srcd_res);
     srca_res_max:=max(srca_res_max,srca_res);
     srca_res_min:=min(srca_res_min,srca_res);
     wtd:=wtd+pi;
     wta:=wta+pi;
    end;

 srcd_pos:=max(srcd_prim_max,srcd_res_max); //src positif des deplacement relatifs
 srcd_neg:=min(srcd_prim_min,srcd_res_min); //src negatifs des deplacements relatifs
 srcd_maximax:=max(srcd_pos,abs(srcd_neg)); //Src maximax des deplacements relatifs

 srca_pos:=max(srca_prim_max,srca_res_max); //src positif des acc relatifs
 srca_neg:=min(srca_prim_min,srca_res_min); //src negatifs des acc relatifs
 srca_maximax:=max(srca_pos,abs(srca_neg)); //Src maximax des acc relatifs

  //remplis les resultats

   result.srca_prim_max:=abs(max(srca_prim_max,reponsea_prim));
   result.srca_prim_min:=min(srca_prim_min,reponsea_prim);
   result.srca_res_max:=max(srca_res_max,srca_res);
   result.srca_res_min:=min(srca_res_min,srca_res);
   result.srca_pos:=max(srca_prim_max,srca_res_max);
   result.scra_neg:=min(srca_prim_min,srca_res_min);
   result.srca_maximax:=max(srca_pos,abs(srca_neg));
     //deplacement

   result.srcd_prim_max:=abs(max(srcd_prim_max,reponsed_prim));
   result.srcd_prim_min:=min(srcd_prim_min,reponsed_prim);
   result.srcd_res_max:=max(srcd_res_max,srcd_res);
   result.srcd_res_min:=min(srcd_res_min,srcd_res);
   result.srcd_pos:=max(srcd_prim_max,srcd_res_max);
   result.scrd_neg:=min(srcd_prim_min,srcd_res_min);
   result.srcd_maximax:=max(srcd_pos,abs(srcd_neg));
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Analyse quantifiée de la foulée en rapport avec les pathologies du sport

L'objectif de cette thèse était de déterminer l'effet de la fatigue sur l'atténuation des ondes de choc d'impact et d'évaluer la relation entre la biomécanique humaine et l'atténuation des chocs pendant la course. Dans cette thèse, nous proposons une nouvelle méthodologie pour l'analyse des événements de choc survenant au cours de la procédure expérimentale proposée. Notre approche est basée sur le spectre de réponse aux chocs (SRS), qui est une fonction basée sur la fréquence utilisée pour indiquer l'amplitude des vibrations dues à un choc ou à un événement transitoire Nous avons utilisé des technologies embarquées tel que les centrales inertielles (IMU) (RunScribe®, San Francisco, CA, USA) pour notre expérimentation.

Les blessures de surmenage en course à pied sont souvent provoquées par la fatigue ou une mauvaise technique, qui se reflètent toutes deux dans la cinématique du coureur. La recherche de pointe sur la cinétique et la cinématique dans le sport utilise des systèmes d'analyse de mouvement qui sont inaccessibles à la plupart des athlètes. Le potentiel des capteurs embarqués pour l'analyse cinétique et cinématique des coureurs est extrêmement pertinent et rentable. Tout au long de nos recherches, nous avons démontré le potentiel des capteurs portables pour l'analyse cinétique et cinématique des coureurs. Nous présentons plusieurs études utilisant des centrales inertielles (IMU) pour l'évaluation du niveau de performance et surveillance de la fatigue. Nous avons extrait de nombreux paramètres de foulée pour les évaluations de performance et de santé. Les capteurs embarqués constituent un outil précieux pour les coureurs, des débutants aux experts, pour l'évaluation de la technique de course.

Notre hypothèse est que la fatigue entraîne une diminution de la capacité d'atténuation des chocs du système musculosquelettique, impliquant ainsi potentiellement un risque plus élevé de blessure due au surmenage.

Mots-clés : Analyse quantifiée de la foulée, centrale inertielle (IMU), fatigue, spectre de réponse au choc, blessure

Gait analysis related to running injuries

The objective of this thesis was to determine the effect of fatigue on impact shock wave attenuation and assess how human biomechanics relate to shock attenuation during running. In this paper, we propose a new methodology for the analysis of shock events occurring during the proposed experimental procedure. Our approach is based on the Shock Response Spectrum (SRS), which is a frequency-based function that is used to indicate the magnitude of vibration due to a shock or a transient event. Five high level CrossFit athletes who ran at least three times per week and who were free from musculoskeletal injury volunteered to take part in this study. Two Micromachined Microelectromechanical Systems (MEMS) accelerometers (RunScribe®, San Francisco, CA, USA) were used for this experiment.

Injuries in running are often provoked by fatigue or improper technique, which are both reflected in the runner's kinematics. State of the art research on kinetics and kinematics in sports is using motion analysis systems that are inaccessible to most athletes. The potential of wearable sensors for runners' kinetic and kinematics analysis is extremely relevant and cost effective. Throughout our research we demonstrate the potential of wearable sensors for runners' kinetic and kinematics analysis. We present several studies using inertial measurement units (IMU) for performance level assessment, training assistance, and fatigue monitoring. We extracted many gait parameters for performance and health assessments. Wearable sensors provide a valuable tool for runners, from beginners to experts, for running technique assessment.

Our hypothesis is that fatigue leads to a decrease in the shock attenuation capacity of the musculoskeletal system, thus potentially implying a higher risk of overuse injury.

Keywords: Gait analysis, micro-electro-mechanical systems (MEMS), fatigue, shock response spectrum, injury

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