TECHNISCHE UNIVERSITÄT MÜNCHEN
TUM School of Engineering and Design

Perception, Adaptation, Customization: An Investigation of Subjective and Objective Methods for the Evaluation of Sport Footwear

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Vollständiger Abdruck der von der
TUM School of Engineering and Design
der Technischen Universität München zur Erlangung des akademischen Grades
eines Doktors der Ingenieurwissenschaften (Dr.-Ing.)
genehmigten Dissertation.

Vorsitz: Prof. Dr. Klaus Bengler

Prüfende der Dissertation:

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Die Dissertation wurde am 31.05.2024 bei der Technischen Universität München eingereicht und durch die TUM School of Engineering and Design am 10.02.2025 angenommen.

They talk of soft sands on the beach, or so I'm told,
But I yearn to feel it with my feet, in steps bold.

(Rabindranath Tagore)

Acknowledgments

I would first of all like to extend my heartfelt appreciation to Professor Veit Senner. His open-hearted support not only gave me the chance to discover my potential but also helped me every step of the way. His encouragement kept me going in moments of difficulty and doubt. I also extend my heartfelt gratitude to Mr. Jürgen Mitternacht, whose generous sharing of knowledge illuminated my academic journey. I extend my thanks to Professor Rodger Kram, whose joy in sharing knowledge not only illuminated my academic path but also acquainted me with the ethos of *paying it forward*.

Special recognition is owed to Prof. Cudmani for allowing me to use their equipment and for providing valuable advice and consulting. Additionally, I am grateful to Dr. Stefan Haug for his assistance with statistics. My sincere appreciation to our company partners at Scott (Robert Pilz), Aspetar (Athol Thomson), and Euro Sport Turf (Mark Trübenbacher). Furthermore, I express my gratitude to Dr. Kreuzpointer and Fabian Stöcke, the former and current head of the Prevention and Performance Lab, for facilitating the use of necessary equipment. My thanks to all the participants who generously dedicated their time to engage in my experiments, making substantial contributions to the advancement of this research. Thanks in addition to Mr. Karl Hughes for his dedicated support and assistance in coaching writing.

My genuine gratitude is also due to my colleagues, whose camaraderie was invaluable in nurturing my development and integration. I won't forget the moments and experiences we shared. Finally, I am deeply thankful to my dedicated students, including Steven Alevras, Long Le Hoang, Oriol Battle, Marc Sloe, Daniel Oberhuber, Michael Briemle, Ilona Herbert, Malte Kohlsaat, and many others, whose invaluable contributions significantly contributed to the successful completion of this research.

My thanks also to my uncle Dr. Abdullah Mesbah and my friend, Dr. Katrin Manzenreiter. Their incisive contributions and readiness to help were invaluable and greatly appreciated.

Most of all, I am profoundly grateful to my father, mother, and two sisters. Despite the physical distance, I always had their unstinting support, infusing energy into every step of my academic journey. Thank you 'Baba' and 'Maman', thank you Bahareh and Banafsheh.

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Abbreviations

ABBREVIATION	MEANING	UNIT
ACL	Anterior cruciate ligament	
ATS	Artificial turf shoe	
ASTM	American Society for Testing and Materials	
bar		Unit for pressure
bpm	Beat per minute	
°C	Degree Celcius	degree of temperature
DIN	Deutsches Institut für Normung	
Deg	Degree of arc (geometrics)	degrees of arc
EMG	Electromyography	
F	Force	Newton [N]
GRF	Ground Reaction Force	Newton [N]
Hz | kHz	Herz, kilo Herz	Unit of frequency
LFWD	Light falling weight deflectometer	
K,M Pa	Kilo, Mega Pascal	Unit of pressure
m | mm | cm | km	Metre | millimetre | centimetre | Vilometre	Unit of distance
N	Newton	Unit of force
Nm	Newton metre	Unit of moment or torque
NGS	Natural grass shoe	
RT	Rotational Traction	Nm
SD	Standard deviation	
v	Speed | absolute of velocity	Metres per Vilome [m/s] or Vilometres per hour [km/h]
V	Volt	Unit for electric potential
VAS	Visual analogue scale	
°	Degree	degrees of arc (geometrics)
%	Percent	Fraction of 100
ANN	Artificial neural network	

ABSTRACT

Sport footwear undergoes a shift from mechanical to the biomechanical framework in the interaction with an athlete. Any alterations to shoes, such as changes in mass, cushioning, and traction, can profoundly affect performance and exacerbate risk of injury. The subtle alterations in shoe attributes from a biomechanical standpoint have been explored to varying degrees, though not exhaustively. A comprehensive understanding of somatosensory perception of these alterations is crucial for elucidating the interplay between footwear characteristics and athletic performance—a domain that remains underinvestigated. This study seeks to bridge the gap between biomechanical quantification and somatosensory perception by studying athletes' discernment of variations in shoe mass, cushioning, and traction during sporting activities. Additionally, the research delves into the mechanisms underpinning athletes' adaptations to such alterations.

Running shoes were altered by mass and sole cushioning in the studies reported on in the first and second chapters of this research series. It was observed that alterations in shoe mass allowed runners to perceive an additional shoe mass of up to 150g (total mass up to ~ 430 g) during short-term running sessions (2 minutes). In Chapter two, modifications in shoe insole cushioning enabled runners to detect minimal changes of 0.7 g-Max during the same short-term running period. The neuromuscular system, as discussed in both chapters, demonstrates a tendency towards a kinetic pattern. When confronted with external factors that disrupt this pattern, such as changes in shoe mass or cushioning, the neuromuscular system responds by adjusting muscle activation, vertical ground reaction force, and plantar peak and mean pressure to uphold the integrity of the original movement pattern. However, kinematic parameters, such as joint motion, step frequency, and contact time, remain constant with no changes and conform to the runner's personalized pattern.

Chapter three focuses on changes in torque in football boots. These alterations were explored by adjusting the outsole configuration. The adjustments made to the boots underwent evaluation through a combination of field tests and laboratory experiments, which simulated human movement using mechanical prosthetic foot models. Our findings reveal that players are capable of detecting minor changes in torque between different outsole configurations, up to 1.2 Nm (rotational traction), particularly during risky movements like cutting. The findings also show that employing both artificial neural networks (ANN) and laboratory tests constitutes a robust approach with the potential to improve the tailoring of outsole configurations. Future studies should consider this combined approach for mitigating the risk of injuries by aligning with the torque perception thresholds of players.

ZUSAMMENFASSUNG

Sportschuhe durchlaufen wegen ihrer Interaktionen mit dem Sportler einen Paradigmenwechsel von einem rein mechanischen zu einem biomechanischen Konstrukt. Folglich sollten alle Veränderungen an Schuhen, wie z. B. Änderungen der Masse, der Dämpfung, der Traktion, Biegesteifigkeit und Torsionsfähigkeit, usw. gründlich auf ihre Auswirkungen auf die Leistung und das Verletzungsrisiko des Läufers untersucht werden. Die Notwendigkeit, subtile Veränderungen der Schuheigenschaften aus biomechanischer Sicht quantitativ zu untersuchen, wurde erkannt und in unterschiedlichem Maße, wenn auch noch nicht erschöpfend, untersucht, nicht zuletzt weil dies nicht einfach zu untersuchen ist. Trotzdem ist ein umfassendes Verständnis der somatosensorischen Wahrnehmung dieser Veränderungen von entscheidender Bedeutung, um das komplizierte Zusammenspiel zwischen Schuheigenschaften und sportlicher Leistung aufzuklären – ein Bereich, der noch unzureichend erforscht ist.

Diese Studie versucht, die bestehende Lücke zwischen biomechanischer Quantifizierung und somatosensorischer Wahrnehmung zu schließen, indem sie die Fähigkeit von Sportlern zur Unterscheidung von Variationen in Schuhmasse, Dämpfung und Traktion während sportlicher Aktivitäten untersucht. Darüber hinaus befasst sich die Forschung in dieser Studie mit den komplizierten Mechanismen, die der Anpassung von Sportlern an solche Veränderungen zugrunde liegen.

Schuhe für den Laufsport wurden im ersten und zweiten Kapitel dieser Forschungsreihe durch Veränderung des Gewichts der Schuhe und der Sohlendämpfung modifiziert. Es wurde beobachtet, dass es bei Veränderungen des Schuhgewichts den Läufern möglich war, während kurzer Laufeinheiten (2 Minuten) eine zusätzliche Schuhmasse bis zu einem Minimum von 150 g (Gesamtmasse des Schuhs bis zu ~ 430 g) wahrzunehmen. Darüber hinaus konnten bei Modifikationen der Dämpfung der Schuheinlegesohlen im zweiten Kapitel, von den Läufern minimale Veränderungen der Maxima der Auftrittsbeschleunigung von 0,7 g-Max während der gleichen kurzen Laufperiode erkannt werden. Das neuromuskuläre System zeigt, wie in beiden Kapiteln diskutiert, eine Tendenz zu einem individuellen stabilen kinetischen Muster. Wenn das neuromuskuläre System mit äußeren Faktoren konfrontiert wird, die dieses Muster stören, wie z. B. Veränderungen der Schuhmasse oder der Dämpfung, reagiert es, indem es die Muskelaktivierung, die vertikale Bodenreaktionskraft und den plantaren Spitzendruck anpasst, um das ursprüngliche Bewegungsmuster möglichst weiterhin aufrechtzuerhalten. Kinematische Parameter wie Gelenk-Winkelbewegung, Schrittfrequenz und Bodenkontaktzeit bleiben weitgehend unverändert und folgen dem individuellen Muster des Läufers.

Das dritte Kapitel konzentriert sich auf die Untersuchung von Veränderungen des Drehmoments zwischen Schuh und Boden beim Fußballspielen, übertragen durch die Gelenkkette des Spielerbeins. Diese Veränderungen wurden durch die Variation der Außensohlenkonfiguration von Schuhen bei Fußballspielern untersucht. Die an den Schuhen vorgenommenen Anpassungen wurden durch eine Kombination aus Feldtests und Laborexperimenten bewertet, bei denen menschliche Bewegungen mit Hilfe von Roboter- oder Prothesenfußmodellen simuliert wurden. Unsere Ergebnisse zeigten, dass Spieler in der Lage sind, auch geringfügige Unterschiede des Drehmoments zwischen verschiedenen Außensohlenkonfigurationen von bis zu minimal 1.2 Nm (Test der Rotationstraktion) zu erkennen, insbesondere bei riskanten Bewegungen wie dem Schneiden. Weiterhin hat sich gezeigt, dass der Einsatz von künstlichen neuronalen Netzen (KNN) und Labortests einen robusten Ansatz darstellt, mit dem Potenzial, die Anpassung von Außensohlenkonfigurationen zu optimieren. Zukünftige Studien sollten diese Anpassungen daraufhin untersuchen, ob Verletzungsrisiken minimiert werden, wenn sie sich an den Drehmoment-Wahrnehmungsschwellen der Spieler orientieren.

1. PREFACE

1.1 THE HISTORY OF HUMAN LOCOMOTION

Bipedalism has independently emerged in terrestrial creatures, including dinosaurs, birds, primates, hopping marsupials, and a few insects and lizards (Clemente, 2014). Morphological alterations observed in mammals indicate that having substantial tail muscles increases their susceptibility and likelihood of adopting bipedalism. (Snyder RC, 1962; Snyder RC, 1952; Persons et al., 2011). However, several survival strategies have influenced the evolution of bipedalism until recently, for instance: carrying objects and tools, avoiding or/and escaping predators. Throughout the course of human evolution, our ancestors experienced a multitude of biological and anatomical transformations (see Figure 1.1). These changes included the development of a robust achilles tendon, serving as a reservoir for storing and perhaps releasing elastic energy (Vereecke et al., 2013). Our ancestors evolved feet that act as levers equipped with spring-like mechanisms (Ker et al., 1987), along with the development of a robust, adducted hallux for enhanced push-off capabilities (Bramble et al., 2004). Furthermore, during bipedal locomotion, humans utilize their feet as rigid levers, contrasting with the flexible mid-foot region observed in the feet of great apes during push-off. This flexibility is attributed to the absence of arches and mobile mid-foot joints, adaptations intended for enhancing grasping abilities in tree-dwelling locomotion (De Silva, 2010; Elftman and Manter, 1935).

In the transition from Australopithecus to Homo, a reduction in toe length and relative curvature was observed, which served to mitigate bending forces across the phalanges during bipedal push-off (please see Figure 1.2). These observations provide the foundation for the prevailing model of human foot evolution, which suggests that the selection for bipedal walking came at the cost of arboreal locomotor capabilities, resulting in a distinct dichotomy in the anatomy and function of human and great ape feet. Nonetheless, it's crucial to highlight a substantial concern regarding this division – the presumption that all unique characteristics of the human foot exclusively serve the purpose of bipedal walking. This assumption fails to consider the potential adaptation of the human foot for activities beyond walking, such as running (Bramble & Lieberman, 2004).

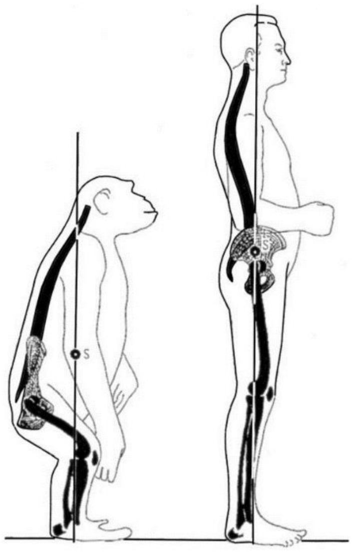

Figure 1.1 Hip and spine dual extension: This involves a vertically aligned femur and the presence of lumbar lordosis. The image was reproduced from Kummer, (2004).

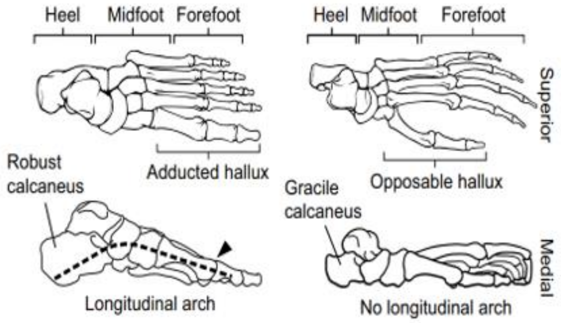

Figure 1.2 Human and chimpanzee feet: Examining the foot skeletons of humans and chimpanzees from both a superior perspective (top) and a medial perspective (below). The arrowhead highlights the dorsal doming of the metatarsal head, a feature found in humans but absent in chimpanzees. The image was reproduced from Holowka et al. (2018).

In addition to the studies mentioned earlier, Lovejoy (1988) made a unique discovery regarding the three-million-year-old femoral neck of Australopithecus afarensis. He found that it shares a similar structure with that of modern humans. This finding implies that even our ancient ancestor 'Lucy' had already evolved to be specialized for walking on two legs. It's truly remarkable how the human walking apparatus has anatomically and physiologically evolved over time. When the femoral neck of a chimpanzee (please see Figure 1.3-left) was compared with that of a human (Figure 1.3-middle), a distinct difference was noticed. The chimpanzee's femoral neck is robust and reinforced at the top, designed to withstand the high bending stress associated with climbing and leaping. In contrast, the human femoral neck has a thinner bone structure at the top, adapted specifically for the stresses of upright walking and running. Interestingly, a fossil femoral neck from a contemporary of 'Lucy' (about 3.2 million years old) (Figure 3-right) exhibits the same structure as that of modern humans, indicating its exclusive adaptation for bipedal walking. More interestingly, in 2019, the discovery of an 11.6-million-year-old skeleton of a pre-human ancestor named 'Udo' (belonging to the species Danuvius guggenmosi) was made in *Hammerschmiede, Bavaria (*Böhme et al., 2019). 'Udo' is significant because this ancient hominin is believed to have been capable of walking upright, adding to our understanding of the evolution of bipedalism in early human ancestors. The discovery of 'Udo' contributes to the growing body of knowledge about the evolution of hominins and their locomotion. Further research into the evolution of human bipedalism is essential, as it has the potential to both challenge and support existing theories and hypotheses, leading to new insights and perspectives.

Figure 1.3 The fermal neck of chimpanzee (left), the human fermoral neck (middle) and the fossile femoral neck from a contemporary of 'Lucy' (right) (The image was reproduced from Lovejoy; 1988.

In general, as early hominids began to walk upright on two legs, they encountered new challenges and demands on their feet. Bipedalism exposed the soles of the feet to different surfaces and environmental conditions, such as rough terrain, sharp objects, and extreme temperatures, which necessitated some form of protection. Additionally, walking upright altered the distribution of weight and pressure on the feet, potentially leading to discomfort or injury without proper footwear to provide support and cushioning. Therefore, the emergence of bipedalism likely spurred the development of rudimentary forms of footwear among early hominids to address these challenges and facilitate more efficient and comfortable movement.

1.2 SPORT SHOE EVOLUTION

Our understanding of ancient footwear remains incomplete and is primarily derived from chance discoveries of well-preserved artifacts. To date, the oldest known shoe, discovered approximately 5,500 BCE in Armenia within Eurasia (see Figure 1.4.a) (Pinhasi et al., 2010). This shoe, referred to as Areni-1, was crafted from leather and was predominantly worn around the heel and hallux, where the most significant pressure is exerted during typical walking. One of its primary functions was presumably to provide protection against cold and shield the feet from potential injuries when encountering sharp surfaces. Since then, the evolution of footwear has been marked by rapid advancements driven by various needs and purposes. Footwear has been tailored for diverse occupational and military contexts and has transcended mere functionality to encompass fashion, status display, and, naturally, sports and athletics. Specialized athletic footwear has been in existence since at least the 1800s when track athletes added spikes to their leather shoes. In the late 1800s, innovative designs incorporating rubber soles, such as those introduced by the Keds Company, entered the market. In the mid-1900s, studded football boots, designed and produced by Adidas AG (Germany), made their debut when they were worn by the German national team during the 1954 World Cup (See Figure 1.4.b). The athletic shoe market underwent a transformative shift in the early 1970s with the emergence of Nike, Inc., an American athletic shoe producer, which gained prominence in the running shoe sector due to its innovations involving foam-based materials. With the rapid evolution of sports training, the growth of sport science, and significant advancements in physiotherapeutic support and sports injury treatments, the needs and expectations of athletes have become increasingly refined. This evolution is reflected in the intricacy of sports shoe design and manufacturing processes. In recent years, additive manufacturing (AM) methods have assumed a progressively vital role in shaping sports shoe design, contributing to improved performance, injury reduction, and overall comfort enhancement. In contrast to conventional tooling and molding prototyping methods, AM techniques make efficient use of raw materials to design and prototype specific footwear components, such as outsoles or midsoles (see Figure 1.4c), while minimizing waste and ensuring precise dimensions (Kim & Oh, 2008; Liu et al., 2013).

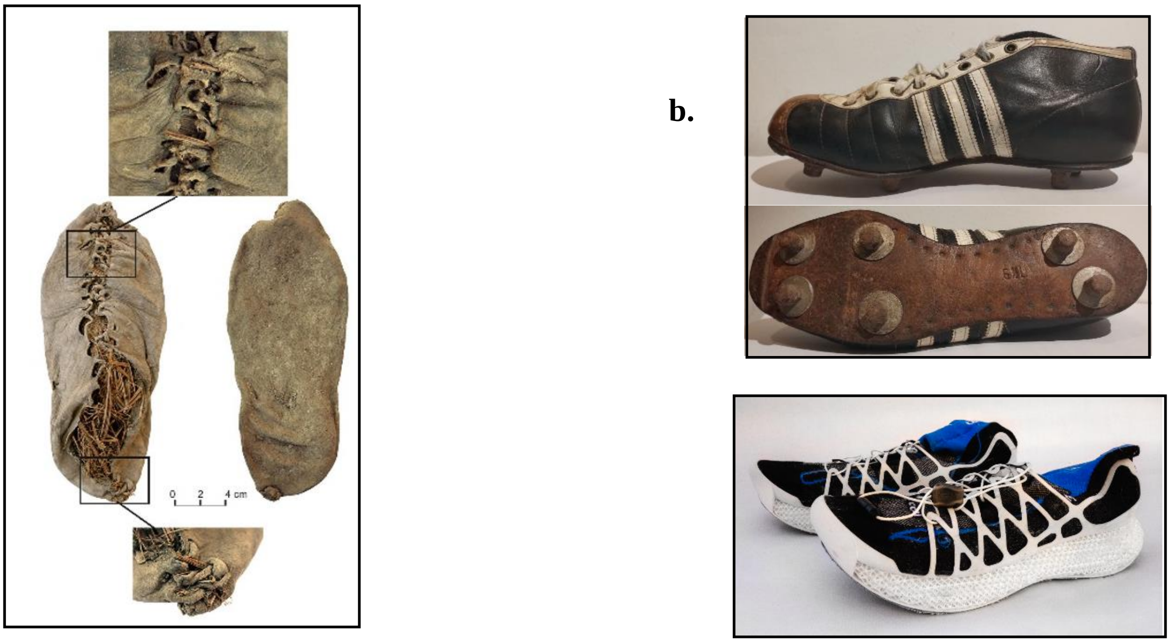

Figure 1.4.a 5.500-year-old leather shoe was preserved in Armenia (The image was reproduced from Pinhasi et al., 2010).**b** Adi Dassler's first football boot designed back to 50s (1954). **c** 3D running shoe was prototyped at Technical University of Munich, 2018.

1.3 THE LOST PUZZLE PIECE

Sport shoes as a purely mechanical system, become part of a biomechanical system once they interact with the athlete. Sports shoes present the foot a new perceptual mode since they enclose the foot and, if they fit properly, allow the foot to rely on alternate kinesthetic feedback to identify changes.Biomechanics and physiology researchers have quantified the advantages of shoe features such as lightweight mass, cushioning of the midsole/insole and the traction of the outsole to understand how the body reacts to such interventions (Wang et al., 2020; Kulmala et al., 2018; Lv et al., 2019). Simultaneously, perception of shoe changes is important to understand the role of footwear features (mass, cushioning, and traction) in athletic performance (Saunders et al., 2004; Perl et al., 2012), safety (Caselli, 2006) and sports product marketing (Clinghan et al., 2008). How athletes adapt to various perturbations or interventions (changes in mass, cushioning, and traction) requires both biomechanical measurements and perception measurements. However, few past studies have investigated footwear performance from both perspectives. Accordingly, fundamental questions remain to be investigated: Can athletes perceive changes in shoe mass, cushioning, and traction while performing the movements of their specific sport? And, how athletes respond to these interventions. These questions are explored in the following chapters.

The present series of studies incorporates a number of studies from different fields, such as mechanical engineering, neuroscience, psychophysics, kinetic, and kinematics to gain a better understanding of the above-mentioned questions.

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CHAPTER 1

(SHOE MASS)

2.1 AIM of the STUDY

Mass (weight) perception has been one of the earliest senses studied by experimental psychologists. In an early exploration of this field, Weber (1834) categorized mass discrimination in two ways: I. Solely through touch and II. Via involving both touch and the muscular sense. He explained that "we receive feedback from muscle tension, pressure, and joint position senses, and the culmination of these inputs provides us with a sense of heaviness". Ross and Brodie (1987) later confirmed that an object's weight or mass can be perceived through the reactive force it exerts on a limb's pressure receptors or through the effort required to hold it raised against gravity. Studies related to mass perception have primarily focused on the just noticeable difference (JND) when using hands (JND is the minimum change in stimulus intensity needed to produce a noticeable difference in sensory perception). The perception of mass changes in the hand, taking into account tactile and muscular components, has been extensively studied (Ross, 2010; Van Beek et al., 2021). In contrast, our understanding of how the foot perceives mass remains limited. Foot mechanoreceptors may have a higher threshold of activation compared to the hand, possibly due to thicker skin (Kennedy and Inglis., 2002). Few studies have investigated mass perception through the feet during running (Slade et al., 2014; Saxton et al., 2020). Shoes introduce a different perceptual challenge for the foot because they enclose it and, if properly sized, do not shift during activity. This necessitates the reliance on distinct kinesthetic feedback to assess mass. As shoes become an extension of the human body, they transition from pure mechanics to biomechanics. Consequently, it is crucial to examine any alterations in shoe properties, such as mass. There is still much to learn about the effects of shoe mass on the running economy (Carranza et al., 2020; Franz et al., 2012; Fuller et al., 2015; Hoogkamer et al., 2016). Fuller et al. (2015) found no detrimental effect on metabolic cost if the combined mass of both shoes is below 440 g. However, if the combined shoe mass exceeds 440 g, there is a positive correlation with metabolic cost. Nevertheless, few studies have explored the effects of shoe mass on biomechanical and spatiotemporal measurements during running. Wang et al. (2020) demonstrated that an increase in shoe mass can lead to elevated ground reaction forces and muscle activity. In a groundbreaking paradigm proposed by Nigg et al. (2017), they explained that when such a perturbation is introduced to the system (increasing shoe mass in running), runners attempt to adapt to this change. However, in this paradigm, it remains unclear how runners perceive mass and what the JND is. Furthermore, past studies have inadequately examined the integration of mass perception with running biomechanics. Specifically, it is unclear what changes in running biomechanics occur the just a noticeable difference is detected. Hence, this chapter has two main objectives: s*ections 1 through 4* investigate the effect of shoe mass (by adding mass to the shoe) on ground reaction forces (GRF), muscle activities, and joint motions. Sections 5 and 6 explore subjective measurements and aim to detect the JND by introducing additional shoe mass. To have a better understanding of all sections (please see the following Figure 2.1):

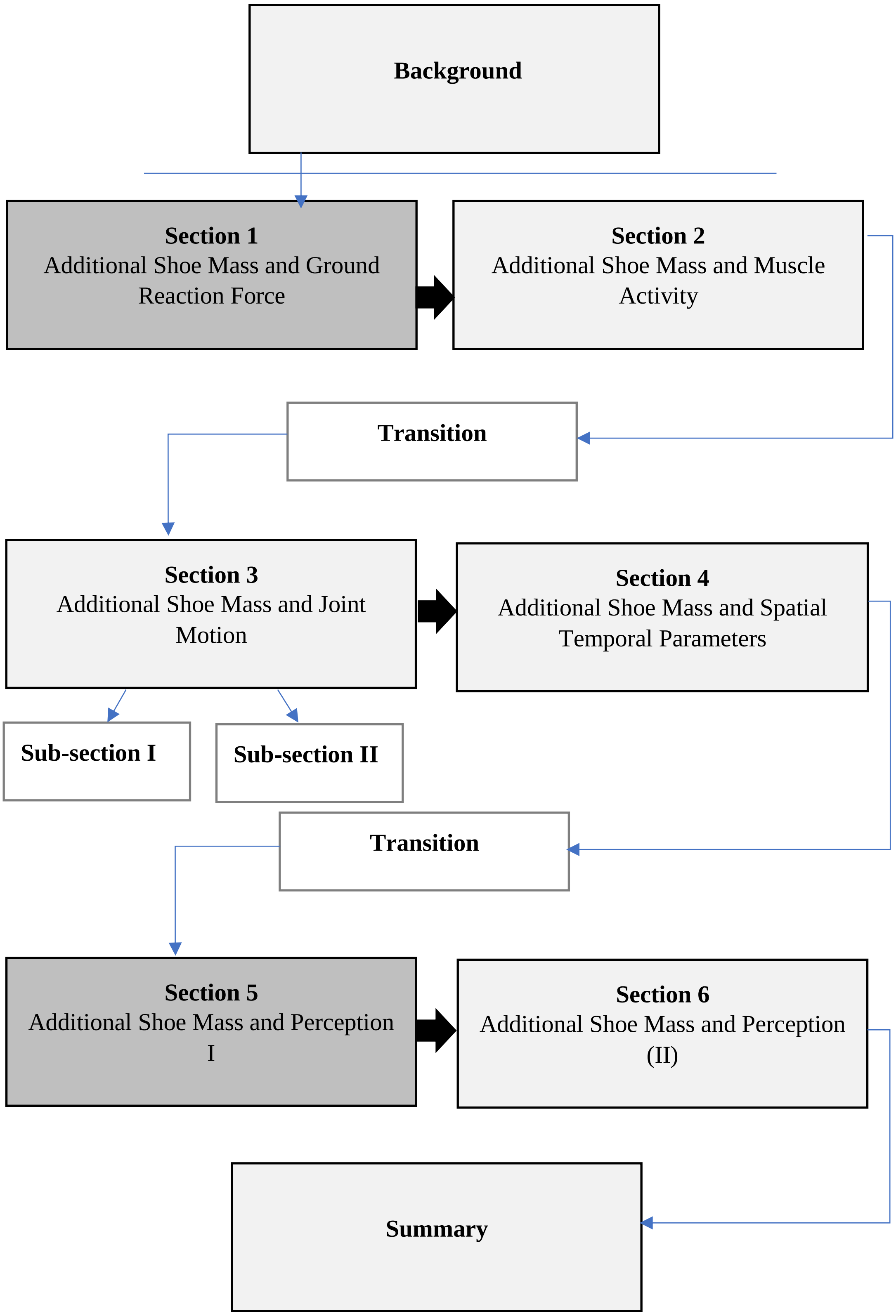


Figure 2.1 Flowchart of all sections in this chapter. The boxes colored dark gray denotes the sections in which the published papers are placed.

2.1.1 SECTION 1

Additional Shoe Mass and Ground Reaction Force

This section is based on the published paper: Keshvari, B., Alevras, S., Mitternacht, J., & Senner, V. (2022). Evaluating the Effect of Shoes with Varying Mass on Vertical Ground Reaction Force Parameters in Short-Term Running. *International journal of exercise science*, 15(1), 191–205.

2.1.1.1 INTRODUCTION

The risk of injuries associated with running, particularly to the lower extremities, is substantial (Hespanhol et al., 2011; Fields et al., 2010; Messier, 2008; Van et al., 2007; Lun et al., 2004; Wen et al., 1978). Such risks are related to a variety of intrinsic factors (Carson et al., 2012) including gender (Riley et al., 2008), foot type (Cowan et al., 1993), and extrinsic factors such as running shoes (Goss et al., 2012; Theisen et al., 2014). Various types of running shoes have been developed to meet the needs of different types of runners, running styles, and running conditions (Pollard et al., 2008). To further understand the effects of shoes on running biomechanics, researchers have compared different running shoes in terms of running economy (Cheung et al., 2016; Ridge et al., 2015; Sobhani et al., 2014) and kinematics (Fredericks et al., 2015; Grimston et al., 1991, Wen et al., 1978). However, in the field of kinetics, few studies have investigated the effect of the mass of running shoes on active peaks and impulses.

For running with an initial rearfoot ground contact, the curve of the vertical Ground Reaction Force (GRF) consists of two peaks: an initial passive peak which results from the foot's initial contact with the ground, while the active peak results from the forces applied by the foot to the ground as it begins to push off (Gruber et al., 2011; Jiang et al., 2020). Active peak is one of the parameters of vertical GRF largely controlled by muscular activity (Nigg et al., 1995). The magnitude and timing of active peaks may lead to musculoskeletal overuse injuries and osteoarthritis (Wiik et al., 2017; Shafizadegan et al., 2016). How the active peak of vertical GRF is affected by shoe type has been investigated in a few studies (Hannigan et al., 2020, Kulmala et al., 2018; Logan et al., 2010; Sobhani et al., 2014; Squadrone et al., 2009; Willy et al., 2014). Kulmala et al. (2018) determined that a maximalist shoe (320g) in comparison to a traditional running shoe (304g) reveals a significantly higher peak vertical GRF despite the greater foam midsole cushioning. In another study, Squadrone et al. (2009) determined that Vibram Fivefingers shoes (148g) elicited a greater active peak vertical GRF when compared with traditional running shoes (341g). The findings of the aforementioned studies show that running shoes have an effect on peak vertical GRF but they are unclear as to whether these biomechanical changes were the result of the structural design or the mass of the shoes. Recently, a study by Wang et al. (2020) investigated the effect of shoe mass alone on peak vertical GRF and estimated the *minimum additional mass* which can significantly influence peak vertical ground reaction. However, their experimental design has certain limitations. Firstly, subjects were not blinded to the shoe mass conditions.

Several studies have determined that lighter shoes can improve performance (Franz et al., 2012; Rodrigo-Carranza et al., 2020). Therefore, it is not surprising that lighter running shoes are more in demand for different levels of runners and that shoe mass is ranked as one of the most important characteristics of running shoes (Honert et al., 2020). Wang et al. (2020) note that participants' expectations about the effect of shoe mass in running biomechanics (especially in running economy) may have generated biases in their performance (Slade et al., 2014) due to their open study as opposed to a blind test, exaggerating the effect size (Hróbjartsson et al., 2014). Wang et al. (2020) also did not include a measure of exertion, such as Borg's rating of perceived exertion (RPE) scale (Borg, 1982), which influences peak vertical GRF (Morin et al., 2011). However, Wang et al. (2020) used preferred velocity in their experiment, stating that 'allowing participants to self-select the running pace facilitated a more natural response to running with an additional mass on the shoes'. This approach is recommended by other researchers (Clarke et al., 1983; Jørgensen et al., 1990; Queen et al., 2009).

The effects of shoe mass on vertical GRF can also be evaluated with another variable, impulse, which has rarely been investigated. The impulse may provide a better measure of the total load on the body because the impulse combines the force and its duration (Carson et al., 2012; Mueller et al 1999; Inklaar et al., 1994). Quantifying the time over which a force is applied allows evaluation of the pathomechanics associated with overuse injuries in the foot plantar (Queen et al., 2007). The purely mechanical system of running shoes becomes a biomechanical system once it interacts with the athlete (Stefanyshyn et al., 2015). Therefore, a firm understanding of shoe characteristics such as shoe mass will help researchers predict and/or determine running-related injuries that result from accumulated microtrauma (caused by an increase of vertical ground reaction forces (Kiernan et al., 2018).

Somatosensory perception of shoe mass has been investigated in a few studies to determine the 'subjective threshold' by using the Just Noticeable Difference (JND) of various shoe masses worn on the feet. This determination comes from scaling of afferent signals from muscle, joint, and pressure receptors, or expectations of the relation between efferent and afferent signals (Carello et al., 2004; Ellis et al., 1999). JND is the minimum amount of shoe mass that must be changed to produce a noticeable variation in sensory experience. Slade et al. (2014) observed the JND of various shoe masses during optional movements, (jumping, walking, etc.). In their study, participants were asked how accurately they perceived the mass among weighted shoes through their feet (namely in a subjective measurement). The perceived mass or subjective threshold was measured subjectively using the Visual Analogue Scale (VAS) or binary questions. The perceived mass can also be ascertained by dividing the JND by the object's mass (intensity of a stimulus) (Weber, 1978), i.e., the 'Weber fraction' ratio. The question of whether changes in the shoe mass of the sensory threshold in such a subjective measurement can be observed by comparing vertical GRF parameters such as active peaks and impulses is not well understood.

Therefore, the first aim of this study was to evaluate the effects of shoe mass on vertical GRF concerning the two variables: active peak vertical GRF force and impulse. The second aim of this study was to investigate the minimum additional mass of running shoes which can influence peak vertical GRF and impulse, thus establishing an *objective threshold*, which could be compared with the subjective threshold of the somatosensory perception of additional shoe masses reported in the literature.

2.1.1.2 METHODS

This study included a pilot study in the form of an online survey, a one-day pretest, and the main test on a separate day one week later. The online survey (the pilot study) was completed with 105 runners, most of whom were members of different running clubs in and around Munich (Germany), and some were also members of the Technical University of Munich Sports Center. Besides collecting biometric information, participants provided the following information:

Q. 1. Average velocity range (AVR) of all running sessions in the last 6 months

Q. 2. Average distance of all running sessions in the last 6 months

Q. 3. Average duration of all running sessions in the last 6 months

Q. 4. Average number of running sessions per week in the last 6 months

The goal of the pilot study is *to gain information from local runners to optimize the design of the experiment* and determine interest in participating in the main test.

*Participants.*Twenty-two male subjects (24.4 years old (± 3.2) and 77. 9 kg (± 8.6) participated in the main study. Most of the participants had filled out the online survey. Runners were required to be injury-free six months before the time of testing. All participants gave written, informed consent before the experiment. The consent form assured the confidentiality of participant data presented the risks of the study, and the protection of personal data through appropriate procedures for anonymization according to Regulation EU (2016). In addition, the consent form assured participants they were free to withdraw from the research at any time without giving a reason and without penalty for not taking part. This research was carried out by the ethical standards of World Medical Association Declaration of Helsinki (2013) and the IJES Ethics Statement (Navalta et al., 2020).

Protocol. Indoor running shoes (Victory Performance, Deichmann GmbH, Germany) were chosen in a range of sizes from EU 42 to 45 and with an average mass of 283 g per shoe. The weight deviation within shoe size was ±3g and between the shoe sizes ± 21g. Each shoe size was categorized as one Baseline shoe (without additional mass) and shoes with four additional masses such as +50 g, +150 g, +250 g, and +315 g (Table 2.1.1 and Figure 2.1.1).

Table 2.1.1 Shoe mass number and total mass of a single shoe *(size 43)*

Shoe ID	Total mass	Ratio
BS*	270g (±3g)**	1
shoe 2	320g	1.18 ×BS
shoe 3	420g	1.55 ×BS
shoe 4	520g	1.92 ×BS
shoe 5	585g	2.16 ×BS

*Baseline shoe=BS. ** Mass tolerances within the shoe size

1. Indoor running shoe (Baseline)

1. b Lead strip added to fore and rear parts

1. c Concealed with the black tape

Figure 2.1.1 Preparation of experimental shoes

Increasing different levels of shoe mass partly follows Slade et al. (2014), who tested for the somatosensory perception of different shoe masses. However, to increase the shoe mass, one main criterion, i.e., the durability of the material used to increase the shoe mass while running at various velocities, was tested in parallel to this study. In comparison to other materials (e.g. plasticine), we determined that lead strips were more durable up to an additional mass of 330g (in upper velocity of 4.4 m/s (16 km/h)), after which it was too heavy to remain fixed. To ensure the stability of the additional mass, we set the upper limit at 315g.

The Gaitway-3D 150/50 treadmill from H/P Cosmos® (Germany) with a conveyor belt 1500 mm (long) × 500 mm (wide) was used. Its speed range can be varied between 0 to 6.1 m/s (22 km/h. This treadmill was equipped with four strain gauge transducers from Arsalis® (2020). The sensors are defined with linearity and hysteresis defined $F_z < 0.2\%$ and F_y and $F_x < 0.8\%$ and a sample rate of 1000 Hz. The heel-strike impacts during running generated vibrations on the treadmill, which could not be completely prevented. The disturbances in the measurement data due to these vibrations were largely eliminated by a Butterworth low-pass filter of 10^{th} order and 18 Hz.

The tool for measuring participants' effort and exertion was the Borg's RPE scale (6-20). Participants were asked to rate their exertion during the activity, taking into consideration feelings of physical stress and fatigue, while disregarding factors such as leg pain or breathlessness but focusing on the whole feeling of exertion.

The purpose of day one (pre-test) was to familiarize participants with the concept of perceived exertion (Borg's RPE: 6-20), acquaint them with the treadmill run, and establish their preferred velocity for the day two (main) test. In total, the participants had ten minutes to warm up according to the following protocol: In the first five minutes, they ran five minutes on the treadmill at their desired velocity (with the BS). They then performed six movements: quad piriform walk, hip opener, arm circles, Frankenstein walk, leg crossover, and inchworm (following the experimenter's instructions).

To start the experiment, it was necessary to find the *preferred velocity* for each participant for day one which would also be used on day two. Participants defined their preferred velocity for the experiment under blind test conditions aided by the experimenter. To determine the preferred velocity for each participant, we took an average range velocity (ARV) from across the sample population of participants (n = 105) in the online survey (Q.1). This was 2.7m/s - 3.3 m/s (10 – 12 km/h). The experimenter then set the treadmill to a velocity within the ARV. The participants requested a reduction or increase in velocity until their preferred velocity was reached. After the warm-up, we gave them extra time to find their velocity. In the first session, participants ran with the BS for 2 minutes before putting on another pair of weighted shoes and running for 2 minutes. They repeated this step for all weighted shoes and the BS. Participants were asked to complete the RPE scale (6-20), thirty seconds before accomplishing the running with BS and weighted shoes. Shoe order was randomized among participants to avoid the learning effect. After completing all possible pair tests with the BS, subjects were given up to one hour until they felt ready to start the second session, following the same procedure as in the first session. After the second session, participants could choose whether to continue with the main test or not.

On day two (main test), one participant dropped out for reasons undisclosed. In total, 21 male runners (24.5 years old (± 3.1) and 77.1 kg (± 7.9)) attended the main experiment voluntarily. All wore identical socks – Falke KGaA – (45% Polypropylene, 35 % cotton, 20 % Polyamide) to minimize sensation variation. After warming up for 10 minutes (as on day one, pretest) participants ran on the treadmill with the BS for 2 minutes. Immediately after this, the participants changed to one of the weighted pairs of shoes and then ran for 8 minutes. Afterward, they put on the BS and ran for 2 minutes before putting on another pair of weighted shoes and running for 8 minutes. They repeated this step for all weighted shoes and the BS (see Table 2.1.2). Participants took 100 seconds of rest after tests 2, 4, 6, and 8 (Table 2.1.2). During the 8 minutes run with all weighted shoes and the BS (tests 2, 4,6, 8, and 10), the data on ground reaction forces were collected for 30 seconds at each of the following intervals: first phase from 1´30´´, middle phase from 4´30´´, and final phase from 7´30´´. The Borg's RPE scale was completed five times in Tests 2, 4, 6, 8, and 10 (see Table 2.1.2) and at the following intervals (including the warm-up duration): 20´30´´, 33´10´´, 45´50´´, 58´30´´ and 71´10´´ (in the final phase). The whole experiment including warm-up and rest duration lasted 73´20´´.

Table 2.1.2 Running with different shoe masses at day-two test (an example of a random order)

#Test	ID	Running Duration	Shoe Change Time	#Test *	ID	Running duration	Rest
1	BS	2 min	60 seconds	*2*	shoe 2	8 min	100 seconds
3	BS	2 min	60 seconds	4	shoe 4	8 min	100 seconds
5	BS	2 min	60 seconds	*6*	shoe 3	8 min	100 seconds
7	BS	2 min	60 seconds	8	shoe 1	8 min	100 seconds
9	BS	2 min	60 seconds	*10*	BS	8 min	-

*Borg Rating of Perceived Exertion (RPE) was used in test order 2,4,6,8 and 10. Note:

On day two (main test) and during each 8-minute run, the peak of vertical GRFs was measured for three phases (phase one from 1´30´´, phase two from 4´30´´, and phase three from 7´30´´) for 30 seconds. The vibrations of the treadmill while impact running could not be completely prevented. Hence, custom software written in Visual Basic 6.0 converted the output from each force sensor to net vertical GRF. Data collected at 1000 Hz were smoothed with a fourth-order low-pass Butterworth filter at a cutoff frequency of 18 Hz. The averages of all peaks for BS and weighted shoes in each phase were then calculated.

Impulse was measured for each shoe mass for each three phases. Vertical forces of stance phases were measured for 30 seconds with respect to the weighted shoes and BS. The integral of force-time was calculated for each stance phase. Finally, the average integral of the stance phases for a certain shoe mass was calculated.

Regarding pre-test and test conditions, the room temperature was constant (23 - 25°C) for both days one and two. In addition, participants were asked to comply with the following instructions before both the pre-test and main test:

1. Avoid heavy training the day before the tests.
2. Abstain from alcoholic beverages the day before both tests.
3. Ensure a minimum of 6 hours of sleep the day before both tests.
4. Do not consume a heavy meal 3 hours before either test.

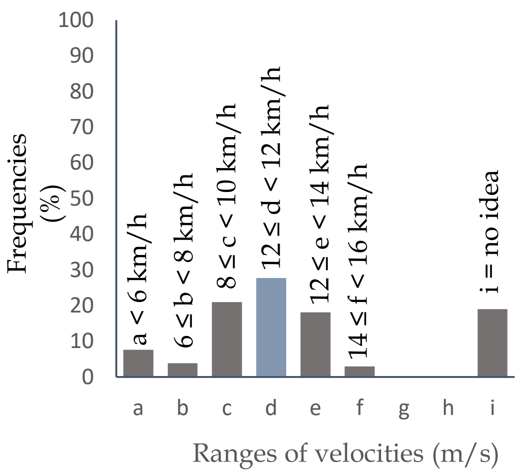
There was a week interval between the tests, which both began at the same time of day. Participants were asked not to touch or handle the shoes at any stage during the test of mass perception through the feet. The experimenters put the shoes on the subjects' feet, laced them, and removed immediately them after the test. Participants were not allowed to walk or jump in their shoes. They began running on the treadmill immediately after lacing. These procedures prevented the participants from gaining any perceptual information about shoe mass that could confound their perception.

In both-day experiments, participants also filled out a binary questionnaire in terms of mass perception after all pair tests. More information will be explained in section 5 and section 6.

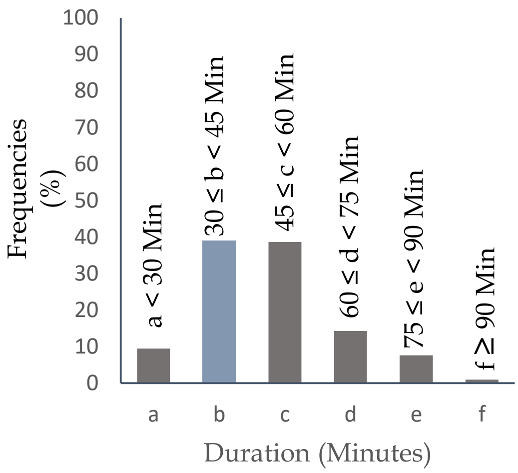
Statistical Analysis. Data on peaks vertical GRF and impulses (force-time integral) were collected and measured from all participants with different shoe masses for 30 seconds at the three time phases. Then, we determined the 'objective threshold', the minimum additional mass of running shoes that can significantly influence peak vertical GRF and impulse. Repeated measures ANOVA with Bonferroni correction (significance level of 0.05) was used to compare different shoe masses with the BS in the statistical software SPSS. Data on perceived exertion was collected five times for each participant. Data on perceived exertion was collected for all participants five times in Tests 2, 4, 6, 8, and 10 (in the final phase). A Spearman's rank order was used to determine whether there is a correlation between time (five-time phases) and RPE.

2.1.1.3 RESULTS

The results of the online survey (Figure 2.1.2) showed that the 'Average velocity of all running sessions in the last 6 months' chosen with the highest percentage of respondents (27.6 %) ranged between 2.7 m/s - 3.3 m/s (10-12 km/h) (Figure 2.1.2.a). In the second question, 'Average duration of all running sessions in the last 6 months', 40 % of participants reported an average duration of 30-45 minutes, and 39.5% reported 45-60 minutes (Figure 2.1.2.b)



2.1.2a Average range velocity of all running sessions in the last 6 months (Q.1)



2.1.2b Average duration of all running sessions in the last 6 months (Q.2)

Figure 2.1.2 Results of two questions of the survey (n =105)

In the third question (Q.3), 31% of participants answered 'Average distance of all running sessions in the last 6 months' as between 5-7 km, and 26.5 % answered with a rate of 7-9 km. And finally, nearly a majority of participants (49.5 %) reported just one running session per week (Q. 4).

Active Peak and Impulse: The active peaks of vertical GRFs for weighted shoes and the BS were measured in three phases. To measure the effects of time and shoe mass and their interaction, an ANOVA with repeated measures was used. Mauchly's Test of Sphericity was performed if the variances of the differences between all combinations of related groups were equal. Our results were consistent with the Mauchly Test requirements. In particular, there was a significant effect of shoe mass on *peak vertical GRF* (Wilks' Lambda = 0.001, F (4.17) = 8.421). Shoe mass also had a significant effect on the vertical GRF *impulse* (Wilks' Lambda = 0.000018, F (4.17) = 15.23). There were no significant effects of time on the GRF variables. There was also no interaction effect of these two independent variables on both dependent variables. In addition, the shoe mass independent variable showed a large effect size on both dependent variables, VGRF = 0.66 and Impulse = 0.78. Time, another independent variable showed a small effect size (Kim, 2015) on both dependent variables VGRF and Impulse respectively 0.002 and 0.02 (see Table 2.1.3 and Table 2.1.4).

Table 2.1.3 Multivariate Test of the peak of GRFs-main and interaction effects

	AM[1]	Value	F	Sig.	PES [3]	HPd[4]	Error df
Time	WL[2]	0.99	0.01	0.98	0.002	2	19
Mass	WL	0.33	8.42	0.001	0.66	4	17
Time × Mass	WL	0.73	0.59	0.76	0.26	8	13

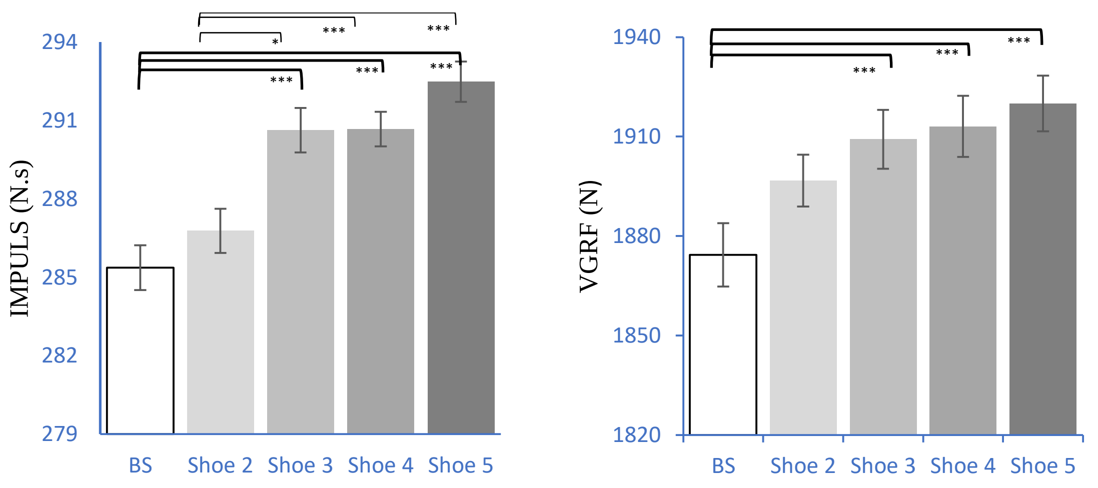
[1] Adjusting Method, [2] WL= Wilks-Lambda. [3] PES= Partial Eta-Squar. [4] Hypothesis df

Table 2.1.4 Multivariate Test of Impulse- main and interaction effects

	AM[1]	Value	F	Sig.	PES [3]	HPd[4]	Error df
Time	WL[2]	0.97	0.26	0.77	0.02	2	19
Mass	WL	0.21	15.23	0.000018	0.78	4	17
Time × Mass	WL	0.61	1.009	0.47	0.38	8	13

[1] Adjusting Method, [2]WL= Wilks-Lambda. [3] PES= Partial Eta-Squar. [4] Hypothesis df

To compare group means, post hoc comparisons using Bonferroni correction were performed. As shown in Figure 2.1.3, the results indicate that peak vertical GRF and the impulses of running shoes with additional mass, 150 g, 250 g, and 315g, differed significantly from those of the BS. In addition, the impulses of the running shoe with 50g additional mass differed significantly from those with additional masses of + 150g, + 250g, and + 315g.



a. Impulse **b.** Peak vertical GRF

Figure 2.1.3.a Mean comparison (and standard errors) of impulses**. b** Peaks vertical GRF ($P \leq 0.001$ and $P \leq 0.05$ are respectively indicated with *** and *.)

Additional points added to the paper: This part has been added to the paper (after publishing) for a better understanding of further sections (section 6).

- In phase 1, the posthoc test with Bonferroni correction determines that VGRF is significantly different in comparison between:
  BS vs. Shoe 3, p = 0.04, mean difference = - 42.9 (an increase of 2 %)
  (95% confidence level = -70.4, -15.4)
  BS vs. Shoe 4, p = 0.010, mean difference = -37.3 (an increase of 1.9 %)
  (95% confidence level = -64.7, -9.8)
  BS vs. Shoe 5, p < 0.001, mean difference = - 60.6 (an increase of 3 %)
  (95% confidence level = -83.9, -37.4)
- In phase 2, the posthoc test with Bonferroni correction determines that VGRF is significantly different in comparison between:
  BS vs. Shoe 3, p = 0.01; mean difference = -39.8 (an increase of 1.3 %)
  (95% confidence level = -70.1, -9.5)
  BS vs. Shoe 4, p = 0.001; mean difference = - 48.8 (an increase of 2.6 %)
  (95% confidence level = -72.2, -25.4).
  BS vs. Shoe 5, p < 0.001); mean difference = - 44.1 (an increase of 2.3 %)
  (95% confidence level = -74.2, -14.3)
- In phase 3, there is no significant difference (p = 0.06).

Borg's Rating of Perceived Exertion: The average RPE was 12.9 (SD =1.3) and the frequency of RPE was highest between 13 and 14 (see Figure 2.1.4). A Spearman's rank-order correlation was used to determine whether the ratings of perceived exertion were influenced by the duration of the experiment. The null hypothesis was defined as a time not affect RPE during the experiment. The result shows there was no statistically significant correlation (p = .19) between RPE and time (five-time phases).

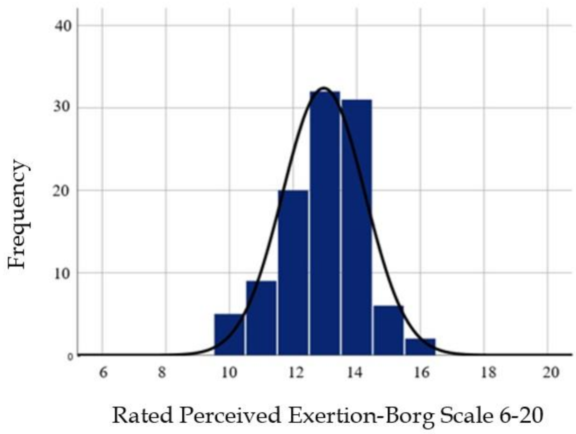


Figure 2.1.4 Histogram of Borg rating of perceived exertion RPE (n = 21 participants ×5 shoes =105)

2.1.1.4 DISCUSSION

The first aim of our study was to evaluate the effect of shoe mass on active peak vertical GRF and vertical GRF impulse, among identical shoe structures. Our findings determined that when additional masses reach +150 g (total shoe mass = 333 g), +250 g (total shoe mass = 433 g), and +315 g (total shoe mass = 598 g), there were significant increases in both variables, active peak vertical GRF and vertical GRF impulse. In particular, our findings show that the active peak of vertical GRF significantly increased with a minimum additional mass of +150 g (between the BS (283 g) and shoe 3 (433 g)).

These findings are in line with those of Wang et al. (2020), who showed an increase in shoe mass from 255 g to 415 g, with a minimum additional mass of 160 g, resulting in a significant increase in the peak of vertical GRF. Nevertheless, our observations revealed discrepancies with the results reported by Wang et al. (2020) concerning the magnitude of the peak vertical ground reaction force (GRF) increment associated with minimal additional mass. In our study, there was a 1.86 % increase in active peak vertical GRF, while Wang et al. (2020), reported a 3.5 % increase (based on an increase of shoe mass up to + 190 g). This difference could be due to the methodological difference. In the Wang et al. study (2020), a ground-mounted force plate was used, however, the force plate method may limit the length of the run and makes it difficult to simulate natural running at a constant velocity in a laboratory situation (Riley et al., 2008). In addition, an instrumented treadmill has been shown to be a highly valid tool for the assessment of vertical GRF, especially for active peaks (Kluitenberg et al., 2012). Instrumented treadmills also allow measurement of peak vertical GRF at the 'preferred velocity' over a longer period (Queen et al., 2009). Furthermore, in our study, a number of factors were controlled for biases in the experiment. We used a blind test condition and included the BS running period between other running periods with the weighted shoes (see Table 2.1.2).

In another study, Kulmala et al. (2018) determined that the Maximalist shoe (321g) showed an increase of 2.8 % in the active peak vertical GRF in comparison to the Baseline shoe (304 g). Although their study was able to show a difference between the two shoes, as these shoes were not identical, the different shoe designs (e.g. midsole thickness) may thus have played a role in registering difference in vertical GRF.

In our study, the preferred velocity methodology followed previous studies (Wang et al.2020; Queen et al., 2007; Clarke et al.1983; Jørgensen et al., 1990), where it was concluded that 'asking the subjects to run at the same velocity may have altered their preferred normal running style'. However, to reach such a preferred velocity, we *did not* simply start from a lower limit e.g. 1.6 m/s (6 km/h), and increase the velocity until the preferred velocity was reached. Nor did we decrease from an upper limit e.g. 4.4 m/s (16 km/h). To avoid bias, we started with an online survey tool (Q.1) and pre-test to reach the preferred velocity (explained above in the Methodology section). It is important to emphasize that we used their preferred velocity on the treadmill, which can be different from their velocity on *terra firma* during their running (Chénier et al., 2018). In addition, another two questions (Q.2 and Q.3) in the survey helped us to establish the average range, duration, and distance run and therefore inform the experimental design. Finally, based on the information from the fourth question (Q.4) in the survey, we were able to complete the experiments (pre-test and main test) in two days on consecutive weeks.

In our research, the effect of heavier shoes (shoe 3, shoe 4, and shoe 5) on peak vertical GRF and impulses can be interpreted by employing Nigg's paradigm theory (Nigg et al., 2017). He determined that 'changing into different shoes usually fails to produce major changes (despite a few minimal changes) in joint kinematics because the neuromuscular system prefers a specific, individual kinematic pattern, called the preferred movement path'. Based on this paradigm, we speculate that when shoe mass interferes with this "path", the neuromuscular system counteracts with muscle activation, such that the original movement pattern is preserved. However, fatigue might have a distorting influence on Nigg's paradigm. In our study, the perceived exertion, which correlates to the measurement of fatigue (Fontes et al., 2010), was measured using Borg's RPE scale. According to the Spearman rank order, there was no statistically significant correlation between RPE and the order of the five tests. This leads us to accept the null hypothesis where there was no discernable time effect on RPE during our experiment. In addition, the average rate of perceived exertion of the Borg's RPE Scale (6-20) was 12.86 (SD = 1.29), with the maximum frequencies occurring at 13 and 14 in the middle of the Borg's RPE scale. Thus, we can conclude that the effect of fatigue was not significant.

In addition to the incremental shoe mass of +150 g, it is noteworthy that shoes 4 and 5 exhibited a notable augmentation in the active ground reaction force (GRF) peaks, with an increase of 2.08 % and 2.45 %, respectively, relative to the baseline (BS) condition. The influence of shoe mass on the vertical GRF was also quantified through impulse (Ns). The introduction of elevated shoe mass in shoes 3, 4, and 5 yielded a proportional increase in impulse, with increments of 1.84 %, 1.85 %, and 2.49 %, respectively, as compared to the baseline condition. Measuring the cumulative force over the stance phase was a sensitive variable, which also illustrates the difference between shoe 2 (additional 50 g mass) and the other weighted shoes. Since the magnitude of active GRF peaks is positively correlated with tibial peak force (Matijevich et al., 2019), the increase of active peaks and impulses in our study may indicate an increased risk of overuse injuries e.g., tibial stress fracture injuries (Grimston et al., 1991). On the other hand, the biological tissue adapts to the level of stress placed upon it, and this adaptation may lead to positive remodeling if the applied stresses are below the tensile limit (Hreljac, 2004). Future studies should investigate whether the minimum shoe mass difference observed in our study (+150 g) can remodel biological tissue positively in longitudinal studies over short and long periods of running time. However, it can be expected that it is very hard to find a 1.86% change in GRF manifesting as bone changes.

The second aim of our study was to 'investigate the minimum additional mass of running shoes which can significantly influence peak vertical GRF and impulse, thus establishing an *objective threshold*, and to compare this with the subjective threshold of the somatosensory perception of additional shoe masses reported in the literature'. Our findings determined that the minimum additional mass producing a significant difference in the active peak and impulse was 150 g. The difference in the objective measurement was determined by a Weber fraction ratio of 0.53 (150 g: 283 g). On the other hand, the Weber fraction in the study of Slade et al. (2014) was calculated by dividing just noticeable differences in mass (142 g) by the original mass of the shoe (220 g), which is equal to 0.64 (142 g: 220 g). Comparing the results of our study with the subjective measurements such as those found by Slade et al. (2014) shows that these ratios are close. However, the difference may be due to the different original shoe masses in the two studies, which are around 63 g. This difference may have affected the subjective threshold, where the Weber fraction decreases when the mass increases up to 200 g (Ross et al., 1987). In our study, an increase of shoe mass up to 150 g (BS = 283g) did not show a significant effect on active peak vertical GRF and vertical GRF impulse. However, in a subjective study by Slade et al. (2014), the perception of mass up to an additional mass of ~ 140 g (BS = 220 g) could not be observed by runners. Thus, it can be cautiously recommended that shoe developers e.g. trail running shoe designers, could therefore design a shoe to enhance other shoe characteristics such as stability and traction that involves adding up to +140 g mass and a Weber fraction of 0.64 without affecting subjective perception during running. However, future studies should investigate the effects of long-term running, where exposure to fatigue, energy consumption, and kinetic changes play a greater role.

2.1.1.5 CONCLUSION

The findings of this study show that active vertical peak GRF and vertical GRF impulse differed significantly between the baseline shoe and weighted shoes with different shoe masses of 150g, 250g, and 315g but not 50g. The additional shoe mass of 150g (1.5 × BS) was shown to be the mass threshold between other masses when compared to the BS. Determining the minimum shoe mass that can influence the kinetic parameters, may be a step towards reducing running-related injuries (resulting from accumulated microtrauma) and also understanding the perceived force and perceived heaviness during running. Future studies should investigate the effects of long-term running, where exposure to fatigue, energy consumption, and kinetic changes may play a greater role.

2.1.2 SECTION 2

Additional Shoe Mass and Muscle Activity

2.1.2.1 BACKGROUND

Past research has demonstrated that various types of running shoes can affect the risk of injury by influencing running biomechanics (Bonacci et al., 2013; Sinclair, 2014; Squadrone and Gallozzi, 2009; Willy and Davis, 2014). These studies examined different running shoe categories, including minimalist shoes, marathon shoes, five-finger shoes, and neutral running shoes, with masses ranging from 150 g to 360 g. However, it remains unclear whether the risk of injuries is primarily attributed to design, mass, or the interplay of both factors. While shoes are inherently mechanical systems, their interaction with runners transforms them into biomechanical systems. Consequently, any alterations to shoes, such as changes in mass, should be thoroughly investigated concerning their effects on performance and injury risk. There is much to learn about the effect of shoe mass on the running economy (Rodrigo-Carranza et al., 2020; Franz et al., 2012; Fuller et al., 2015; Hoogkamer et al., 2016). Fuller et al. (2015) demonstrated that there was no detrimental effect on the metabolic cost when the combined mass of both shoes was below 400g. However, if the combined shoe mass exceeds 400 g, a positive correlation with metabolic cost is observed. Nonetheless, few researchers have evaluated the effect of increased mass on running biomechanics and, specifically, muscle activities during different running phases. Wang et al. (2020) found that running with an additional mass of 240 g (while the original shoe mass was 175 g), equivalent to a Weber fraction of 0.57, resulted in increased muscle activity in the Gastrocnemius lateralis. The investigation of such a perturbation can be approached in two distinct manners: I. Conducting a limited number of repetitions during a single stance phase with constant velocity on a force plate, as demonstrated in the study by Wang et al. (2020), or II. Engaging in short-term running on a treadmill, particularly during multiple stances with a constant velocity. However, both methods may encounter objections related to muscle fatigue, which can be effectively monitored using Borg's rating of perceived exertion (RPE) (Williams et al., 2017). The primary objective of this section is to examine the effect of shoe mass on muscle activities and to determine the just noticeable difference during short-term running.

2.1.2.2 METHODS

Ten male runners participated in our experiment. They had the following average (standard deviation) characteristics: age 24.3 (±2.7) years, height 174.1 (±7.1) centimeters, and weight 71.2 (±11.9) kilograms. To be eligible for the study, runners needed to have been injury-free for at least six months before the testing. All participants provided written, informed consent before the experiment. The consent form guaranteed the confidentiality of participant data, outlined the study's risks and ensured the protection of personal data through anonymization procedures in compliance with the EU General Data Protection Regulation (Regulation (EU) 2016). Additionally, participants were informed that they were free to withdraw from the research at any time without needing to provide a reason, and there would be no penalty for not participating further. Each participant wore Victory Performance indoor running shoes, which were available in sizes ranging from 42 to 44. These indoor running shoes (Victory Performance, Deichmann GmbH, Germany) had an average mass of 283 g. The mass variation within shoe sizes was ±3g, and the difference between shoe sizes was ±21 g. For each shoe size, there were three identical shoes, each prepared with different shoe masses as follows: the Baseline shoe (BS), Shoe 2 with an additional mass of +150 g, and Shoe 3 with an additional mass of +315 g. The shoe mass was increased by adding a lead stripe to the forefoot and rearfoot areas while keeping the center of mass approximately constant. All shoes were covered with black tape. Participants were asked to try various shoe sizes to find the one that offered the most comfortable fit. If they were unable to find a suitable fit, their data were excluded from the analysis. In our study, the experimenter assisted participants in putting on the shoes, because Slade et al. (2014) found that participants were capable of discerning slight differences in shoe mass when handling different shoes by hand.

A surface EMG system (Noraxon EMG TeleMyo 2400T) with a sample rate of 1500 HZ was used with electrodes placed on the dominant leg for four muscles: (I) Rectus Femoris, (II) Tibialis Anterior, (III) Gastrocnemius Lateralis and (IV) Gastrocnemius Medialis. The placement of the electrodes is shown in Figure 2.2.1.

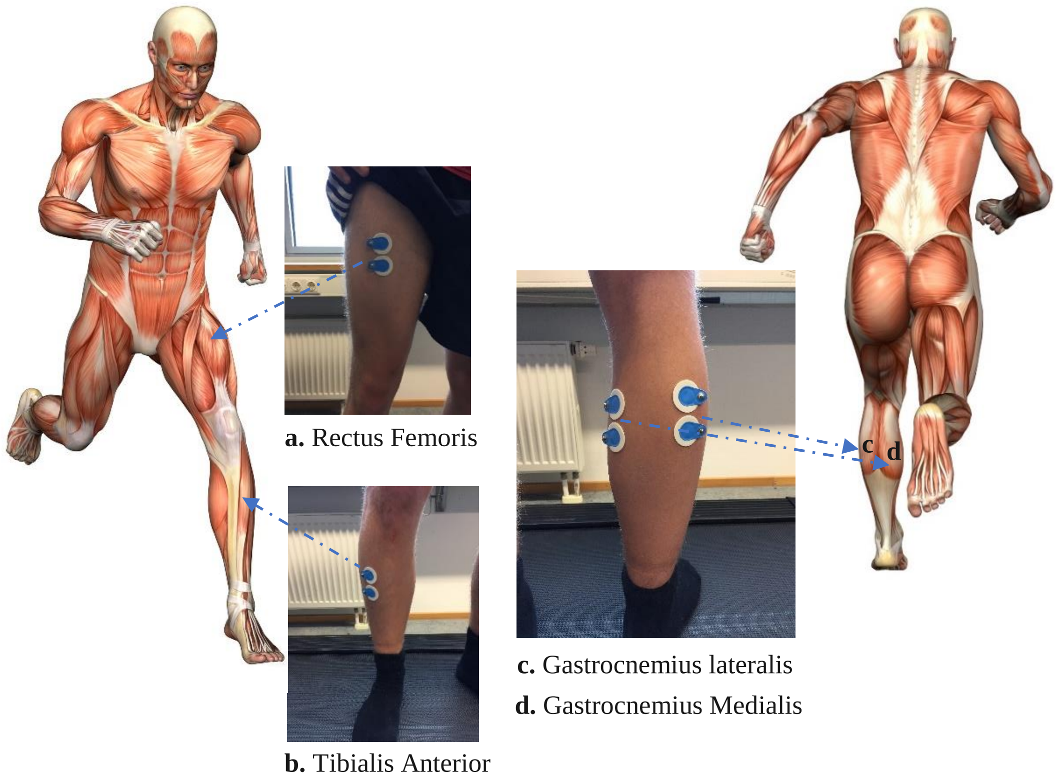


Figure 2.2.1 Muscles used in Surface EMG in a pilot study (copyright © dreamstime.com)

Participants equipped with EMG electrodes were asked to warm up for 10 minutes on a treadmill (Horizon Paragon 6) at a velocity of 2.7 m/s. During warm-up, they were asked whether they would like to modify the velocity. Participants were informed that the chosen velocity would be used in the rest of the experiment. After warm-up, participants ran with a randomly chosen shoe for 3 minutes. In the last minutes of each run, the subjects had to rate their exertion according to Borg's rating of perceived exertion (RPE) scale (6-20). Participants were afforded one minute to rest and change shoes. This procedure occurred for each of the test shoes. EMG data were acquired for all three minutes. However, to analyze the data, 120 seconds of the total 180 seconds were chosen with a minimum of 10 seconds margin from the beginning and ending points (see Figure 2.2.2). This period was divided into three phases, the *early phase*: 0-10 seconds, the *middle phase*: 55-65 seconds, and the *final phase*: 110-120 seconds. The data was filtered using a Bandpass (50 Hz -500 Hz).

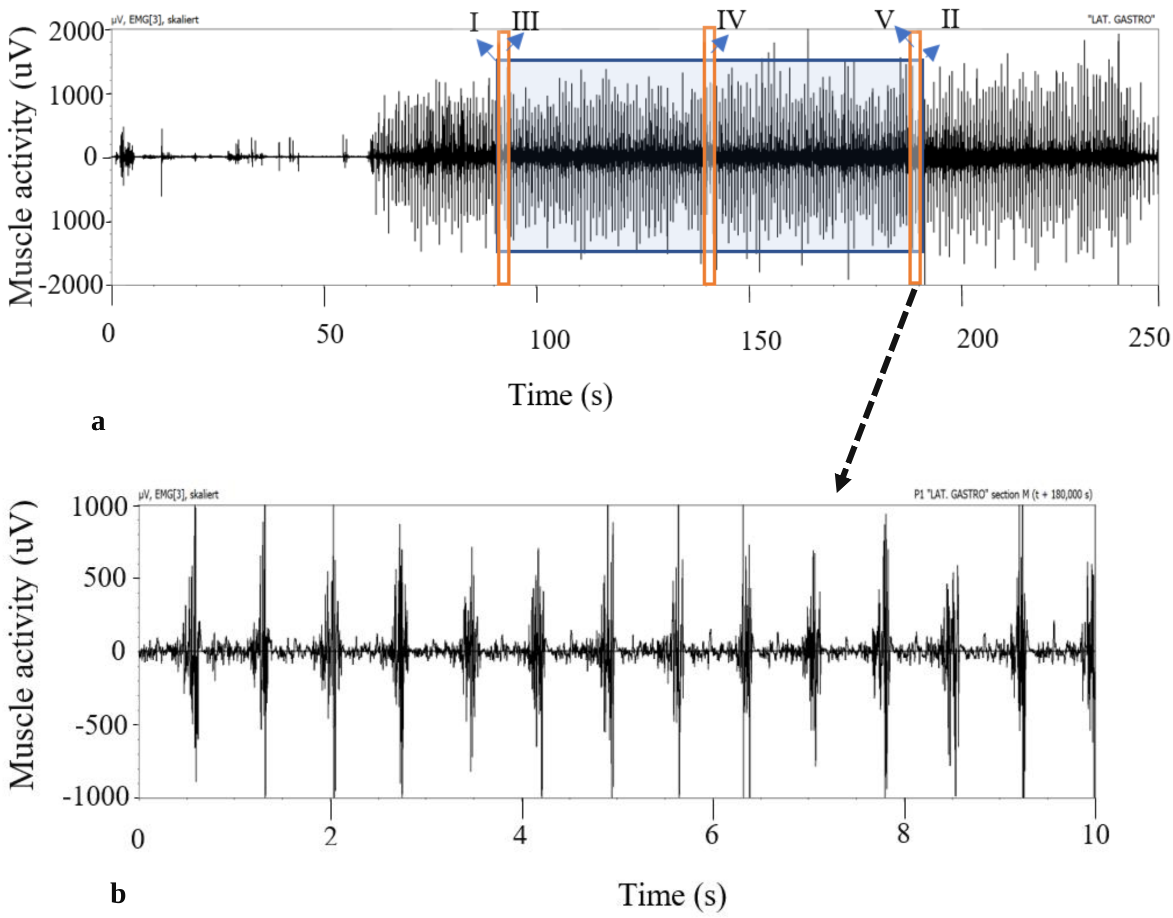


Figure 2.2.2.a An example of raw data of Muscle gastrocnemius lateralis. *i-ii:* Period of 120 seconds used in data analysis. *iii, iv and v:* Three phases each with 10 seconds duration. **b.** gastrocnemius lateralis for 10 seconds.

Mean muscle activity and integral of muscle activity in both swing and stance phases are calculated in our study (an example of a final diagram of running gait, please see Figure 2.2.3). In the following EMG analysis, 40 % of stride time is assumed to be the ground contact duration with 60 % swing phase duration. This was based on past studies of Novacheck (1998) who found the ground contact time (stance phase) is 39 % when running at 3.2 m/s (11 km/h). Data was analyzed using pair sample T-tests with alpha value = 0.5. All data was normalized based on BS.

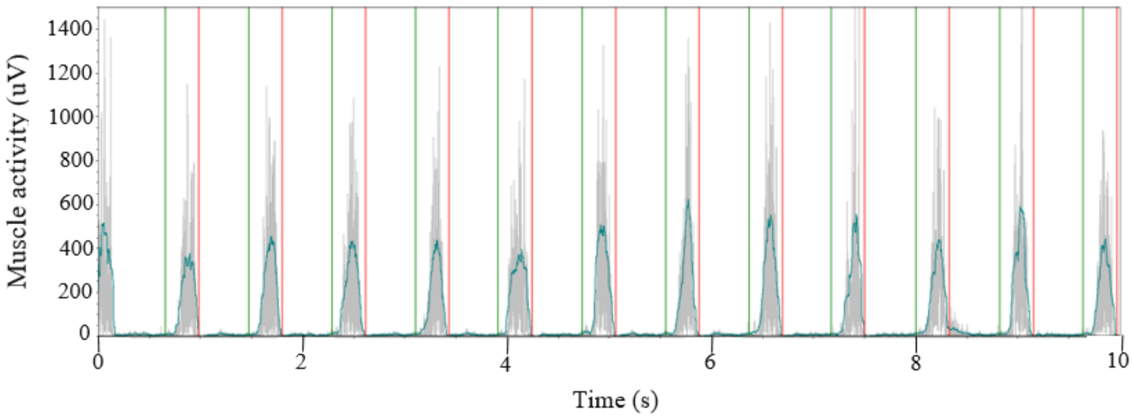


Figure 2.2.3 Selected 10 seconds of data (1500 samples per second) for a representative participant for the Gastrocnemius lateralis after filtering band pass 50 Hz-500 Hz.

2.1.2.3 RESULTS

Integrated (uV.s) and mean (uV) EMG of four muscles are shown in Tables 2.2.1 to 2.2.4. Overall, muscle activity (uV) was influenced by increasing shoe mass. According to our findings, three lower extremity muscles: Tibialis anterior, Gastrocnemius lateralis and Gastrocnemius medialis showed significant changes in swing and stance phases. These changes reached statistical significance when the Weber fraction was equal to 0.53 and corresponding to the addition of 150 g (please see Figure 2.2.4 to 2.2.7)

Table 2.2.1 Mean (uV) and Integrated (uV.s) of Rectus Femoris

	Stance		swing	
	Integrated	mean	Integrated	mean
BS	12.8 (1.5)	42.8 (1.6)	3.46 (0.5)	7.8 (1.2)
S2	13.9 (1.7)	45.6 (1.9)	3.9 (0.68)	8.9 (1.5)
S3	15.2 (1.9)	49.9 (1.9)	4.3 (0.93)	9.7 (2.1)

Table 2.2.2 Mean (uV) and Integrated (uV.s) of Tibialis Anterior

	Stance		Swing	
	Integrated	mean	Integrated	mean
BS	20.9 (4.2)	70.1 (14.9)	36.5 (5.7)	82.4 (13.3)
S2	22.5 (4.4)	73.9 (14.9)	36.2 (5.4)	80.9 (12.5)
S3	24.4 (4.6)	80.3 (15.4)	41.6 (5.2)	92.5 (14.1)

Table 2.2.3 Mean (uV) and Integrated (uV.s) of Gastrocnemius lateralis

	Stance		swing	
	Integrated	mean	Integrated	mean
BS	31.4 (3.7)	104.7 (11.3)	3.8 (0.8)	10.4 (2.1)
S2	31.4 (4.2)	102.8 (12.2)	3.71 (0.8)	9.1 (2.2)
S3	32.9 (4.2)	107.9 (12.6)	4.65 (1.2)	11.6 (3.0)

Table 2.2.4 Mean (uV) and Integrated (uV.s) of Gastrocnemius medialis

	Stance		Swing	
	Integrated	mean	Integrated	mean
BS	35.2 (3.1)	117.7 (9.4)	4.6 (0.7)	10.5 (1.6)
S2	34.7 (3.45)	114.2 (10.2)	4.50 (0.6)	10.1 (1.3)
S3	38.7 (3.5)	127.5 (10.7)	4.7 (0.9)	10.6 (1.9)

The ratio of the EMG in the modified shoes to the EMG for the baseline shoe condition yields a muscle activity coefficient. T-tests between (*shoe 2/BS)* and (*shoe 3/BS)* for all muscles have been analyzed. In the following Figures, the muscle activity coefficients for mean and integrated EMG for the stance and swing phases.

I. Mean of Muscle activities in the stance phase

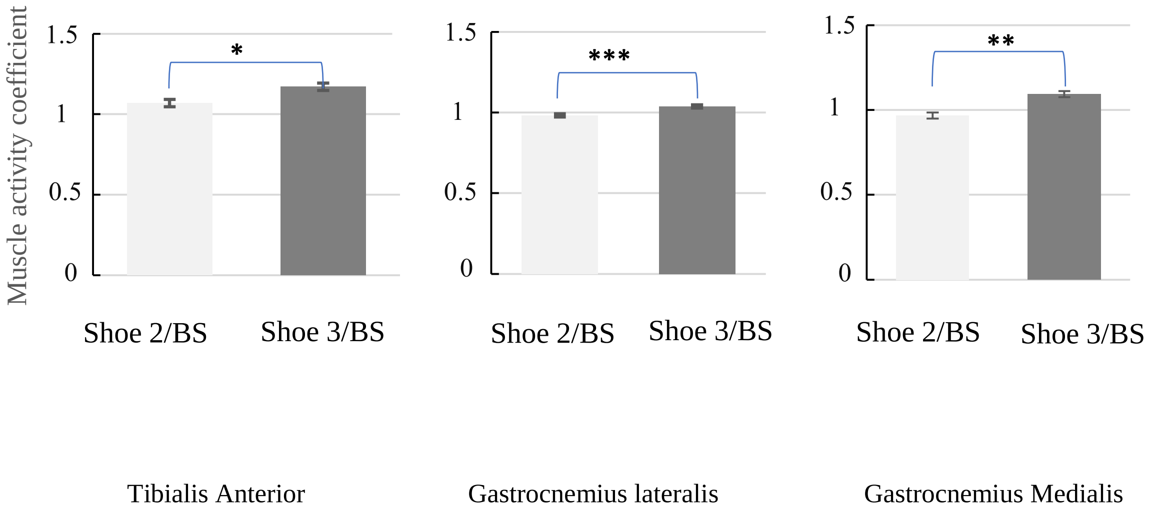
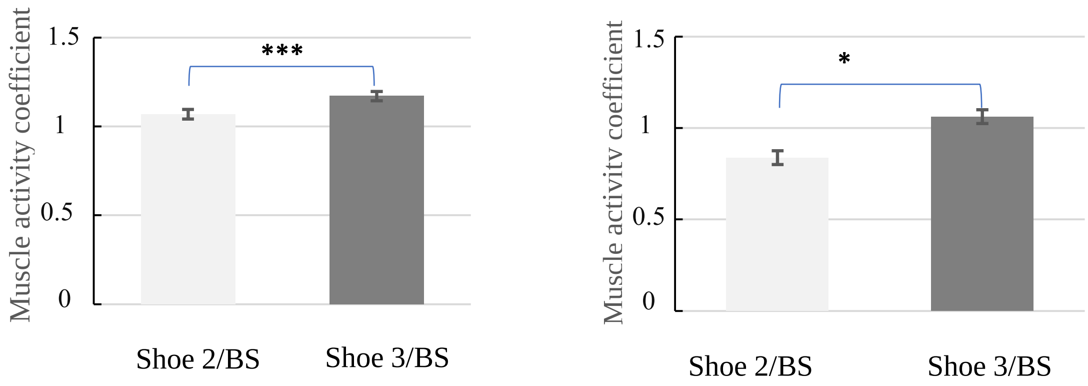


Figure 2.2.4 Mean Muscle activity in stance phase (Paired sample t-test for two groups: p-value is less than 0.05, it is flagged with one star (*). p-value is less than 0.01, it is flagged with 2 stars (**). the p-value is less than 0.001, it is flagged with three stars (***).

II. Mean of Muscle activities in the swing phase

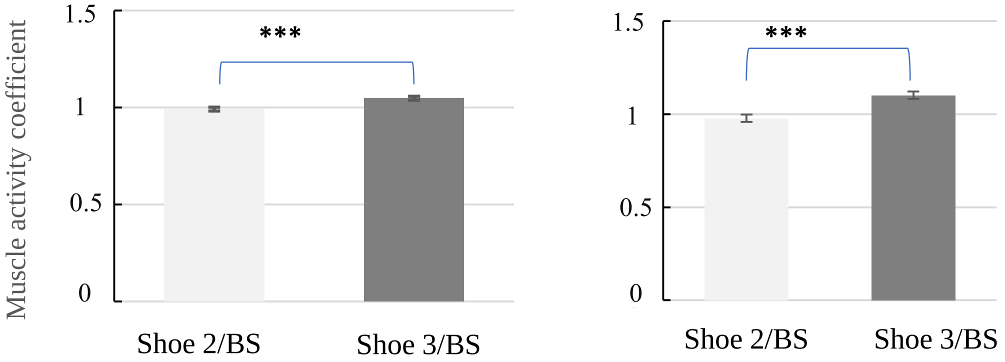


Tibialis Anterior

Gastrocnemius Lateralis

Figure 2.2.5 Mean Muscle activity in the swing phase (paired sample t-test for two groups): p-value is less than 0.05, it is flagged with one star (*). p-value is less than 0.01, it is flagged with 2 stars (**). the p-value is less than 0.001, it is flagged with three stars (***).

III. Integrated EMG in the stance phase

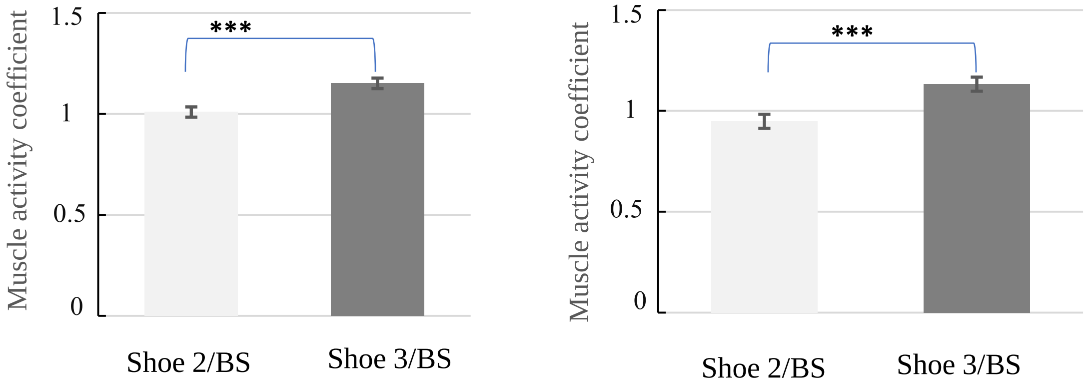


Gastrocnemius lateralis

Gastrocnemius Medialis

Figure 2.2.6 Integrated EMG in swing phase paired sample t-test for two groups: p-value is less than 0.05, it is flagged with one star (*). p-value is less than 0.01, it is flagged with 2 stars (**). the p-value is less than 0.001, it is flagged with three stars (***).

IV. Integrated EMG in the swing phase



Tibialis Anterior Gastrocnemius lateralis

Figure 2.2.7 Integrated EMG in swing phase (paired sample t-test for two groups: p-value is less than 0.05, it is flagged with one star (*). p-value is less than 0.01, it is flagged with 2 stars (**). the p-value is less than 0.001, it is flagged with three stars (***).

Increasing shoe mass significantly changed the EMG of three muscles: Tibialis Anterior, Gastrocnemius Medialis, and Gastrocnemius Lateralis differed in both stance and swing phases for both mean and integrated metrics. As shown in Table 2.2.1 - 2.2.4, the muscle, Tibialis Anterior showed smaller changes in mean muscle activity in comparison to other muscles. This was due to an increase in muscle activity in the swing phase. Average Borg rating of perceived exertion (RPE) values in the first phase were 11.2 (SD = 2.4), in the second phase 11.5 (SD = 2.5), and in the third phase 11.8 (standard deviation = 2.3), and averaged overall 11.5 (SD = 2.3). Those values are considered to be a light level of exertion.

2.1.2.4 DISCUSSION

Our findings revealed significant increases in the mean and integrated muscle activities (EMG) during both the swing and stance phases. The intensity of three lower extremity muscles, namely Tibialis Anterior, Gastrocnemius Lateralis, and Gastrocnemius Medialis, increased when the Weber fraction was 0.53 (150 g : 283 g). These results align with those of Wang et al. (2020), who demonstrated that an increase in total shoe mass (greater than 400 g) was associated with increased activity in the Gastrocnemius Lateralis. There are several differences between the present and the Wang et al. studies that make direct comparisons challenging. For example, in the Wang et al. study, no information was provided about the exertion levels of the participants. In our study, we measured participants' exertion using the Borg Rating of Perceived Exertion (RPE) scale (6-20). Our results indicate that the physical activities performed during the experiment corresponded to a light level of intensity, approximately an

'11' on the RPE scale. This information can help address concerns about the potential effect of fatigue during the experiment. Secondly, the use of non-blinded testing may introduce performance bias, and thirdly, Wang et al.'s (2020) experiment was based on a limited number of trials conducted on a force plate, whereas our study involved a continuous treadmill running protocol with an average of 35 steps. In our findings, when shoe mass was increased from 433g to 598g the Tibialis Anterior muscle was active throughout both the stance and swing phases of running, which is consistent with the findings of So et al. (2022). Tibialis Anterior exhibited significantly greater activity compared to the other three muscles in our study. We presume that Tibialis Anterior activity was predominantly concentric/isometric, facilitating foot clearance from the support surface during the swing phase of the running gait. It can be speculated that these lower limb muscles (Tibialis Anterior, Gastrocnemius Medialis, and Gastrocnemius Lateralis) exhibited vibration relative to the bones during each stance phase of running, particularly in the pre-activation phase. Continuous vibration in this manner can potentially damage soft tissues and lead to soreness (Van Gent et al., 2007). To prevent such injuries, Nigg et al. (2017) introduced the Muscle Tuning Paradigm, in which runners adjust their muscular responses to minimize vibrations. An analogy can be drawn here with a wine glass: when tapped with a metal spoon, it vibrates and produces a sound, which corresponds to its natural frequency. Similarly, in our case, an alteration in shoe mass can influence the interaction between the foot and the ground. If the vibration frequency at impact closely matches the natural frequency of the soft tissues, it can lead to a phenomenon known as resonance, where vibrations are maximized. It is plausible that such vibrations result in increased EMG activity, particularly in the early stance phase (pre-activation phase).

Future studies should examine whether an increase in shoe mass might be detected by the Golgi Tendon Organ (GTO), particularly during the isometric/concentric phase of the swing phase. Subsequently, the central nervous system can use this proprioceptive information to monitor the amount of force generated by a muscle. It can be speculated that during the swing phase, particularly in the isometric/concentric phase of the swing phase, when the muscle generates more force, there is a greater stretch of collagen and fibers, especially concerning the Tibialis Anterior (TA). This increased force may also lead to a faster rate of action potential transmission along the axon (see Figure 2.2.8).

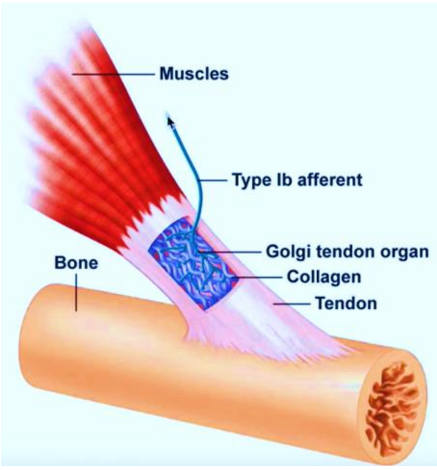


Figure 2.2.8 A schematic of Golgi tendon organ implemented in ligament between muscle and bone (The image was reproduced from Tokoph et al. (2019))

Therefore, one possible mechanism for perceiving the additional shoe mass (which should be examined in a future study) could be presumed to occur during the swing phase.

2.1.2.5 CONCLUSION

The effect of an additional mass of 150g and 315g was investigated on four muscles during 2 minutes of running. Our findings determined that when the Weber fraction is 0.53 (150 g : 283 g), there were differences in the mean and Integrated of three muscles: Tibialis Anterior, Gastrocnemius Medialis, and Gastrocnemius Lateralis, in comparison to the baseline shoe, during both the stance and swing phases. Future studies should investigate spatial-temporal variables and joint motion when such a perturbation is imposed on the system.

2.1.3 SECTION 3

Additional Shoe Mass and Joint Motion

2.1.3.1 BACKGROUND

Researchers have recently found that different types of running shoes can affect running biomechanics and may influence the risk of injury (Bonacci et al., 2013; Sinclair, 2014; Squadrone and Gallozzi, 2009; Willy and Davis, 2014). Past studies have mostly researched running economy and few have concentrated on joint motion. Stacoff et al. (2000) show with bone pin studies that the skeletal movement in running changes little when the shoe conditions are changed. Similarly, Nigg et al. (2017) quantified the effect of different footwear conditions on ankle and knee joint kinematics during running. Their results indicate that the mean kinematics of the ankle and knee joints were almost identical between a conventional running shoe and a racing flat and a minimalist shoe. Along with changes in design, there are also differences in the mass of running shoes (minimalist shoes are about 175 g, and conventional running shoes are about 300 g). To understand the effect of shoe mass on joint motion, Wang et al. (2020) performed running biomechanical measurements for running shoes of various weights. They show that even when the total shoe mass increased up to 415 g (original shoe mass was 175 g), there were no significant changes in the kinematics of the ankle and knee joints. The effects of shoe mass on joint motion merit further investigation. An understanding of how an additional mass influences joint motion may aid in injury assessment and possible prevention. In addition, such investigations may lead to a better understanding of how runners perceive and adapt to changes in shoe mass. Therefore, this section investigates how joint motion in running changes with additional shoe mass. It is divided into two parts: In the first part, the effect of additional shoe mass on knee joint kinematics is evaluated during short-term running. In the second part, the kinematics of the knee joint of four participants were evaluated over longer-term running (30 minutes).

2.1.3.2 METHODS

PART ONE.

Eight male runners with a mean age of 24.4 years old (SD = 2.2), body mass of 72.6 kg (SD = 6.2), and height of 178.3 cm (SD = 6.06) participated in the one-day experiment. They were injury-free for the 3 months before the experiment. This research was carried out in accordance with the ethical standards of the 1964 Helsinki Declaration (World Medical Association Declaration of Helsinki, 2013).

Protocol. Indoor running shoes (Victory Performance, Deichmann GmbH, Germany) were chosen in a range of sizes from EU 42 to 45 and with an average mass of 283 g. Four pairs (one baseline without additional weight and three pairs with additional weights) of the shoes per shoe size, were prepared in different weights. To increase the shoe weight, lead stripe in different ranges +50 g, +150 g, and +315 g, was adhered to the shoes in the forefoot and rearfoot areas. This kept the center of mass of the shoe fairly constant. Four shoes: Baseline Shoe, BS=283g, shoe 1 = 1.18 × BS, shoe 2 = 1.55 × BS and shoe 3 = 2.16 × BS used in this experiment. The weight deviation within shoe size was ±3 g and between the shoe sizes ± 21g. The experiment started with a short warm-up of five minutes of treadmill running with the Baseline shoe. Then, the experiment divided into part 1 and part 2. In part 1, participants wore at first the BS and ran on the treadmill for two minutes. Immediately after, they took off the BS and the other weighted shoes. Participants took one-minute rest after each two minute test. In part 2, participants first wore one of the weighted shoe pairs and ran two minutes on the treadmill. Immediately after that, they changed to BS and ran for two minutes. This procedure continues for the other shoe conditions. The total experiment (with the rests) comprised about 60 minutes. The running velocity was fixed at 2.7 m/s (10 km/h). Participants could choose the desired velocity chosen at the beginning of the test while warming up.

Kinematic measurement. Eight Vicon Nexus cameras were placed surrounding the treadmill (see Figures 2.3.1 and 2.3.2). Next, 21 reflective markers were fixed over the anatomical landmarks and placed on the subject's lower extremities according to Plug-in-gait (Stief et al 2013).

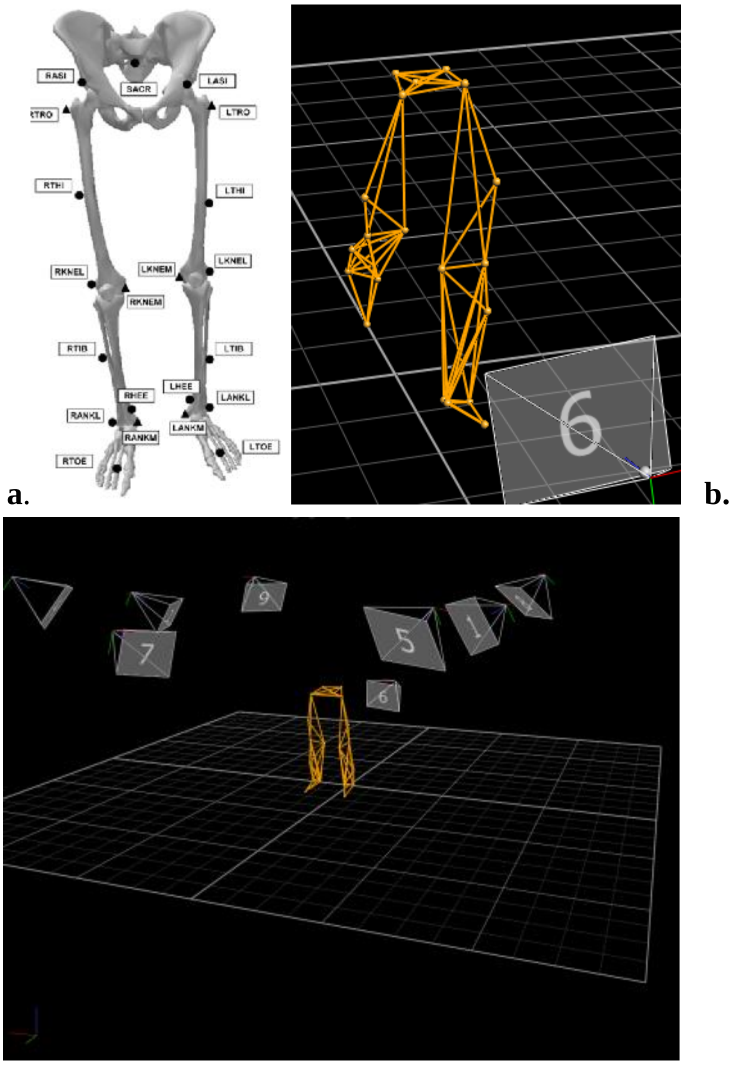


Figure 2.3.1.a Using plug-in gait while running an experiment using treadmill **2.3.1.b** Plug-in gait model followed by Stief et al., (2013) **2.3.1.c** Position of cameras while experiment

Figure 2.3.2 Positions of the number of Vicon cameras on the day of the experiment

The calibration method consists of recording 3000 samples of the trajectory of one reflector on a calibration wand for all the cameras. A ten-second recording of a volunteer, with the 21 reflectors attached to his body was found sufficient to define the model. During both part 1 and part 2, eight seconds (100 frames) of kinematic data for each shoe condition was obtained at 120 Hz in the middle of each test, after one minute of running. The mean of maximal knee flexion angle in the stance phase and swing phase was measured for BS and the weighted shoes in section 1 and 2. A Garmin Heart Rate chest band was worn as well as other 21 reflectors according to Stief et al. (2013) proposed in their custom-made protocol. Heart rate was recorded with a Garmin Forerunner 920XT smartwatch and collected in the last 10 seconds of each two-minute run. This dependent variable has given valuable information about vigorous-intensity physical activity.

PART TWO.

Four trained participants were recruited for this experiment. They reported running a minimum of 60 km per week. Running velocity in this study was based on each subject's desired velocity. In a pilot study, participants' desire velocity was defined (see following Table 2.3.1).

Table 2.3.1 Participants characteristics in the experiment (sample of all four participants)

Subject	**Velocity m/s (km/h)**	**Weight (kg)**	**Height (cm)**	**Starting Condition**
A	2.9 (10.5)	64	185	Weighted
B	3.6 (13)	79	185	Weighted
C	2.9 (10.5)	85	186	Baseline
D	3.4 (12.5)	64	169	Baseline

Kinematic data was captured by using the Vicon motion analysis System (VICON Motion Systems Ltd). Ten 3D-motion cameras were used in this 3D-motion setup**.** Eight MX-T10 cameras and two Bonita B10 cameras, set with a sampling rate of 250 Hz captured the running motions. The raw data was processed with Vicon Nexus 2.5. Markers were placed on the subjects as per a lower body Plug-in gait model based on the Newington-Helen Hayes gait model Baudet et al. 2014 (see Figure 2.3.3.a). This marker set comprises 16 markers, 6 per limb and 4 for the pelvis (see Figure 2.3.3.b). In our case, this offered the possibility to track trajectories and obtain angular kinematics of each joint in the lower limb.

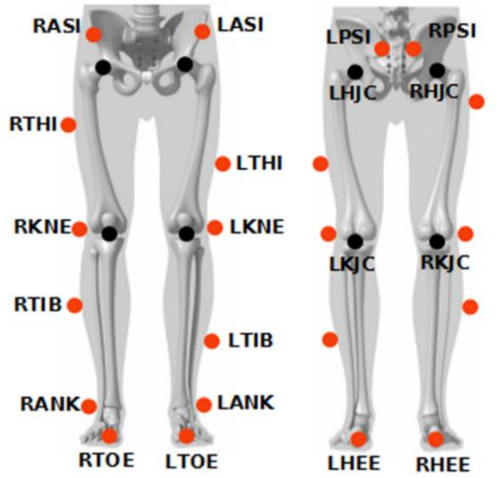


Figure 2.3.3.a Newington-Helen Hayes Lower-Body Plug-In-Gait suggested by Baudet et al. (2014).

Figure 2.3.3.b Marker Setup on the subject; View from anterior and posterior

2.1.3.3 RESULTS

PART ONE.

Peak knee flexion in the stance phase. ANOVA one-way repeated measure analysis and post hoc test (with Bonferroni correction) were used to determine whether the weighted shoe conditions were kinematically different from the baseline shoe (Alpha level is 0.05). Two assumptions of repeated measure were fulfilled: I. Normality II. Mauchly's test indicated that the assumption of sphericity had not been violated, χ^2 (5) = 4.8, p = 0.439. A repeated-measures ANOVA showed that peak knee flexion did not differ significantly across the four different shoe conditions, F (3, 21) = 0.633, p = 0 .58 (please see Figure 2.3.4).

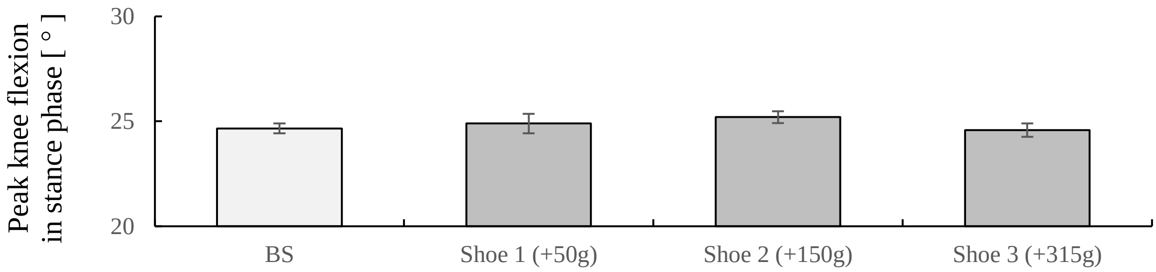


Figure 2.3.4 Mean of peak knee flexion for different running shoes (baseline and weighted shoes) in the stance phase. Using error bar SEM (+/-2)

Peak knee flexion in the swing phase. Data was analyzed with a one-way, repeated measure. ANOVA. The two assumptions of this test were defined: I. the data were normally distributed and II. Mauchly's test indicated that the assumption of sphericity had been violated, χ^2 (5) = 11.65 and p = 0.04. Therefore, modifications needed to be made to the degrees of freedom so that a valid F-ratio could be obtained. The Greenhouse–Geisser correction was used in our calculation. After using such a correction, a repeated-measures ANOVA determined that peak knee flexion/extension differed significantly among the four different shoe conditions F (1.55, 10.85) = 13.85, p = 0.002. A post-hoc test with Bonferroni correction determined that there was a significant difference between baseline shoes vs. shoe 3 (p = 0.015), and baseline vs shoe 4 (p = 0.00007) (please see Figure 2.3.5).

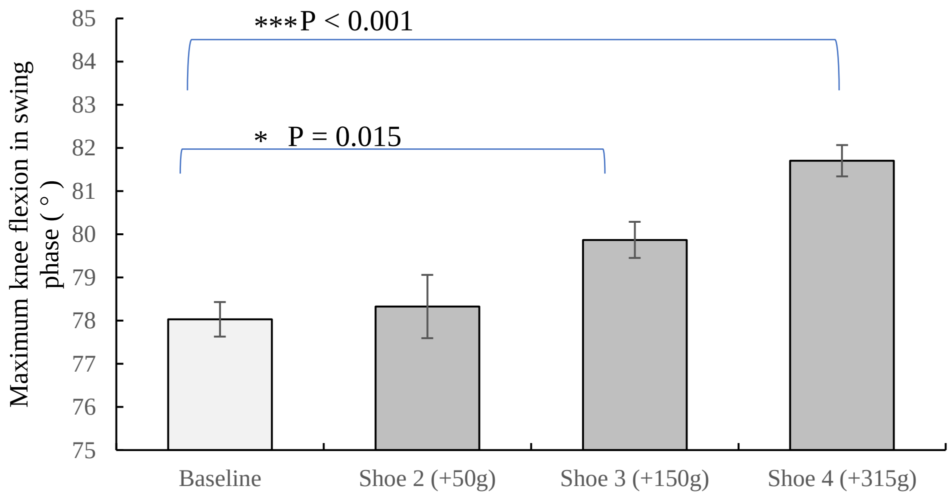


Figure 2.3.5 Mean of peak knee flexion for different running shoes (baseline and weighted shoes) in the swing phase (Using error bar SEM (+/-2)).

The target heart rate of vigorous-intensity physical activity is ascribed to heart rates between 77 % and 93% of the maximum heart rate (Riebe et al. 2018). The maximum age-related heart rate is subtracted from the average age (24.4 years old is the average age in the study, section 3-part one) from 220 bpm (Shookster et al. 2020). Thus, theorically, the group average maximum heart rate should be 195.58 bpm, and vigorous-intensity physical activity levels should be between 155 and 182 bpm. The data on heart rate is shown in Table 2.3.2.

Table 2.3.2 Mean heart rate data (beats/min) of all participants in different sections with baseline and weighted shoes

Section	S1	S1	S1	S1	S2	S2	S2	S2
Shoe	*BS*	Shoe 1	Shoe 2	Shoe 3	*BS*	Shoe 1	Shoe 2	Shoe 3
Mean	154.18	152.75	156.33	156.42	157.98	157.58	159	160.33
Std. Dev	16.16	14.52	15.00	18.52	13.22	14.99	12.81	13.63
Minimum	121	131	130	110	141	138	137	141
Maximum	181	180	182	184	180	183	177	185
95% Confidence interval of Mean	145.04 163.32	144.53 160.97	147.85 164.82	145.94 166.90	150.51 165.46	149.10 166.06	151.71 166.29	152.62 168.04

PART TWO.

Knee extension/flexion data for swing and stance phases with weighted shoes in comparison to baseline shoes is shown in Figure 2.3.6. Mean knee flexion (°) in five sections of the stride was calculated and is shown in Table 2.3.3. An increase in knee flexion was tangible for subjects C and D with an increase of 8° and 9° respectively.

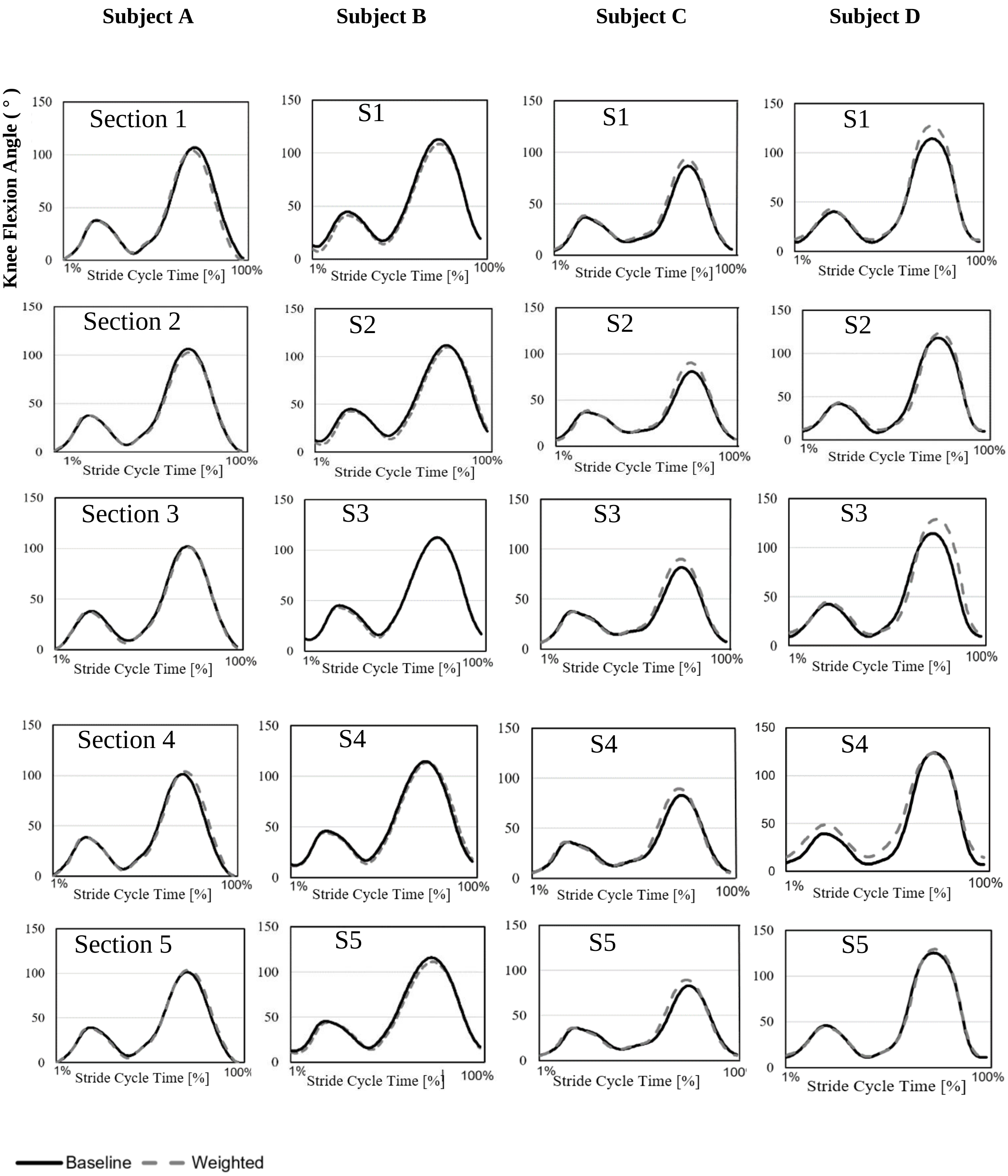


Figure 2.3.6 Mean knee flexion of right and left legs for four participants on five sections (S1-S5)

Table 2.3.3 Mean peak knee flexion of five sections in stance and swing phases per participant (mean of left and right legs)

	Stance Phase		Swing Phase	
	BS	**WS**	**BS**	**WS**
Subject A	42.0	39.3	107.2	104.2
	(4.0)*	(1.3)	(4.8)	(1.7)
Subject B	45.4	43.1	113.6	111.0
	(1.1)	(1.8)	(1.9)	(2.7)
Subject C	37.1	38.1	83.7	91.7
	(0.6)	(1.2)	(2.7)	(1.7)
Subject D	42.9	44.0	118.0	127.5
	(2.1)	(0.8)	(4.5)	(2.5)

*standard deviation

2.1.3.4 DISCUSSION

In Part I of the study, it was found that there were no significant changes in knee flexion during stance phase between the baseline shoe and the other weighted shoe conditions. In the swing phase, there were small (up to 4°) but statistically significant changes between baseline shoe and other weighted shoe conditions. The addition of a mass of 150 g to each shoe resulted in an increase of peak flexion up to 2° (in comparison to BS) while adding 315 g caused an increase of peak flexion up to 4°. These small changes in the magnitude of joint kinematics using skin and shoe-mounted markers have been observed at discrete events during running in the studies of Paquette et al. (2013), Hardin et al. (2004), Squadrone et al. (2014). A difference of 2° falls below this threshold of reliability of skin marker-based 3D motion (Ferber et al., 2002). Similraly, in the stance phase, the greatest change of knee flexion was 0.5° (in comparison to the baseline shoe) this may be due to skin movement (not the actual joint angle). Our findings are in line with those of Nigg (2017), who showed that when a perturbation is imposed on a runner's shoe, they maintain their running "signature" by maintaining their joint motion.

In Part 2 of the study, when the running velocity was faster than 3.3 m/s (12 km/hour), the peak knee flexion exceeded 110° for subjects B and D. An increase of velocity in the study has an effect on an increase of knee flexion in swing phase, is in line with those of Liu et al. (2021). In addition, our findings also show that there is a larger kinematic change between weighted shoe vs. baseline shoe in the swing phase (up to 7 ° and 9 °), in comparison to the stance phase. Thus, by combining the data of studies 1 and 2, It can be concluded that additional shoe mass does not affect the stance phase in comparison to the modest effect that added shoe mass has on swing phase joint kinematics.

2.1.3.5 CONCLUSION

Our findings indicate that knee joint motion during the stance phase is conserved with the addition of up to 598 g of single shoe mass. However, small but significant increases in knee joint flexion/extension occurred with an additional mass of 150 g during the swing phase in constant-velocity running. In the second study, an increase in shoe mass up to 400 g resulted in a notable disparity in knee joint motion in the swing phase.

2.1.4 SECTION 4

Additional Shoe Mass and Spatial-Temporal Parameters

2.1.4.1 BACKGROUND

Perturbation experiments on running with loads carried on the back and/or legs have taken both biomechanical and metabolic approaches. Such studies go back to the 1970s when Cureton et al. (1978) who looked at back-carried loads, remarked that 'running form and efficiency were not hampered by carrying the added weight'. On the other hand, an increase of 1000 g additional mass on the leg increased metabolic power by up to 16 % (Coifman et al., 2024). In terms of biomechanical approaches, Taylor et al. (1980) measured cadence in various animals include humans carrying back loads in running. They found a constant stride frequency with added backloads of up to 30 % of body mass in humans. Past studies determined no effect of large backloads on contact time and cadence, however, they reported that metabolic cost is six times as expensive to carry a given mass on the feet or ankles as it is on the back (Catlin et al., 1979; Frederick, 1985; Myers et al., 1985). In detail, Myers et al. (1985) reported changes in stride length and flight time when extra mass was added to the ankles compared to the lower limbs. A more recent study, by Mc Callion et al. (2014), determined that when running barefoot, stride frequency was significantly higher than in conventional and minimalist footwear. It was not clarified whether the change was due to the cushioning effect, mass, or interaction of these two variables. However, in the other study, Rodrigo-Carranza et al. (2020) determined that across three different shoe mass conditions (Baseline, + 50 g, and + 100 g) neither cadence nor contact time changed. It is hard to find a consensus since the baseline shoe and additional mass differed across these studies. Therefore, this section aimed to investigate the effects of additional shoe mass on spatiotemporal parameters such as cadence and contact time, in short-term running.

2.1.4.2 METHODS

Twelve male subjects weighing 74.9 (SD = 8.2) kg and height 181.7 cm (SD = 4.8) participated in this two-day experiment. The subjects were recreational runners with differing levels of running experience, reaching from 5 km races to marathon running. They reported that they had no lower limb injuries over the last six months. The study focussed on male participants to avoid sex disparities due to different biomechanical locomotion patterns in men and women (Ferber et al., 2003). Running shoes, *Roadhawk FF- Asics* (see Figure 2.4.1.a) were provided in sizes ranging from US 7.5 to US 12.5. This running shoe model has a 20 mm midsole thickness at the heel and a 12 mm midsole thickness at the forefoot. The original mass of the shoe was 233 grams (US size 9). Although the additional mass used during the experiments was the same, the total mass of the shoes for each subject differed slightly due to the baseline mass of the shoes changed according to their size. The mass deviation within shoe size was ±2 g and between the shoe sizes ± 28 g. Two types of shoes were used in the experiment: a baseline shoe (BS) and weighted shoes. To increase the shoe mass, 400g lead strips (see Figure 2.4.1.b) were placed into the tongue of the shoe (in total for right and left, 800g). The position is similar to the one chosen by Franz et al. (2012). To avoid any bias due to weight perception, two lightweight plastic strips (Figure 2.4.1.b) were also added to the tongues of the baseline shoes as a decoy. In addition, participants were told that both shoes were manipulated but without providing any further information about shoe mass. Placebo sticks and lead sticks (see Figure 2.4.1.b) were used for manipulating shoe mass.

2.4.1a **2.4.1b**

Figure 2.4.1.a Overall view of the running shoes used in the experiment **.b** Two lead strips in the middle of the photograph are each 400g and were inserted into the tongues of the shoes (O1 and O2) vs. two plastic decoy strips with a mass of 10g each (P1 and P2).

To capture the spatiotemporal parameter, cadence, and contact time, an OptoGait system (Optogait; Microgate, Bolzano, Italy) was used in our experiment. This system is an optical measuring instrument composed of a sending and a receiving measuring bar, each with 96 LED lights. Two parallel bars of the device system are integrated into the side edges of a motorized treadmill (HP cosmos Pulsar 4P; HP cosmos Sports & Medical, Gmbh, Nußdorf, Germany) at the 3 mm over the contact surface so that no extra fixation and calibration is needed. The device is connected to a computer which is controlled by the experimenter. Data is sampled at 1000 Hz so that parameters such as contact and flight time are measured with a precision of 1/1000 seconds (with a space resolution of 1.04 cm). Stride cycles were captured from foot contact to the subsequent foot contact. Contact time is defined as the time from when the foot contacts the ground to when the toes lift off the ground. Stride frequency is expressed as the number of ground contact events per minute (García-Pinillos et al., 2018). More than one project has proved the validity and reliability of measurements of spatiotemporal parameters with this tool. (Healy et al., 2019; Lee et al., 2014; Lienhard et al., 2013).

Procedure. Subjects participated on two different days within two consecutive weeks. They were asked to not do any intensive activity the day before the tests. In addition, they were asked to maintain the same eating habits on the two testing days, and the two testing phases were started at the same time of the day. The conditions were randomized, meaning the starting condition differed between the participants to avoid systematic measurement errors due to a learning or fatigue effect. The subjects were given identical socks to minimize comfort bias. Participants were asked to run on the treadmill with either the weighted shoe or baseline shoe in random order between the two days. To define the velocity for the experiment, a session called the *pre-testing phase* was conducted a few days before the actual experiment. Participants were satisfied by the velocity of 2.9 m/s (= 10.5 km/h) for both days of experiments.

On day 1, subjects did a warm-up of at least five minutes in a barefoot condition to familiarize themselves with the treadmill (H/P cosmos Pulsar 4P; H/P cosmos Sports & Medical, Gmbh, Nußdorf, Germany). This warm-up was done to avoid any bias in the reproducibility of gait on the treadmill. Testing on both days comprised 30 minutes where subjects ran at a velocity of

2.9 m/s (= 10.5 km/h) once with random shoes (either baseline or weighted). The measurements were done at 5 different measuring times within 30 minutes (see Figure 2.4.2). The measuring durations were varied from 5 seconds to over 30 seconds. Objective exertion was defined by measuring the heart rate. The heart rate was measured with a Polar M400 watch that included a *Polar H7 Heart Rate Sensor*. The heart rate was observed over the last seconds of each measurement, and the mean was noted at the end of each measuring section. Similar procedures occurred on Day 2 approximately one week after with the opposite shoe condition.

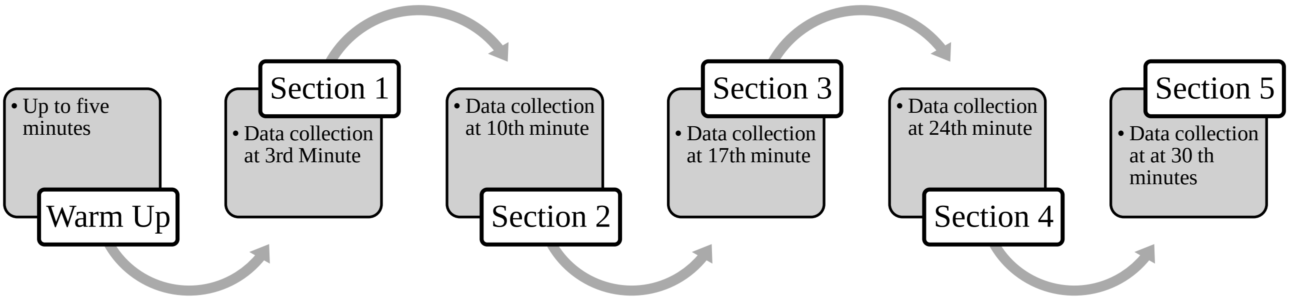


Figure 2.4.2 Experiment procedure with five sections of acquiring Optogait data

Data analysis method. Between 8 and 15 consecutive strides were analyzed for the step kinematics. This follows previous studies that investigated similar parameters (Chapman et al., 2012; Folland et al., 2017) and using the same measuring device (Roche-Seruendo et al., 2018). A maximum of 30 consecutive steps of data remained after the elimination process. Invalid steps were eliminated based on OptoGait User Manual 1.12.1(Rizzetto, 2018), so that no invalid times or frequencies could bias the results. The minimum contact time was defined at 10 milliseconds. If a step did not fulfill this criterion, it was considered invalid and automatically deleted by the system.

2.1.4.3 RESULTS

A paired sample T-test between the baseline and weighted shoe determined that there is no significant difference in stride frequency between the baseline shoe and weighted shoe (paired t-test: t = 0.9 df = 11, p = 0.37). The mean stride frequency for baseline and weighted shoes were 159.4 steps/min (SD = 6.9) and 158.9 (SD = 6.7) respectively. A non-parametric test, Wilcoxon signed-rank test was used to compare contact time between conditions. It showed that an increase in shoe mass up to 400 g did not elicit a statistically significant change in contact time (Z = -3.13, p = 0.75). Indeed, the median contact time was 0.31 ms for both weighted and baseline conditions. Finally, the mean heart rate in all five sections per participant was measured for the baseline and weighted shoe conditions. Paired sample test showed that there was a significant difference between the two conditions (Paired T-test: t = -2.32, df = 11, p = 0.04). The mean heart rate for baseline and weighted shoes are as follows 152.3 (SD = 11.1) and 155.1

(SD = 10.7) (please see Table 2.4.1 and Figure 2.4.3). The data on contact time was not normally distributed, while the data on heart rate was normally distributed.

Table 2.4.1 Descriptive analysis with a 95% Confidence Interval of the Difference in heart rate between baseline and weighted shoe conditions

	Mean difference	Std. Deviation	Std. Error Mean	Lower limit	Upper limit
BS - WS	-2.80	4.18	1.21	-5.46	-0.14

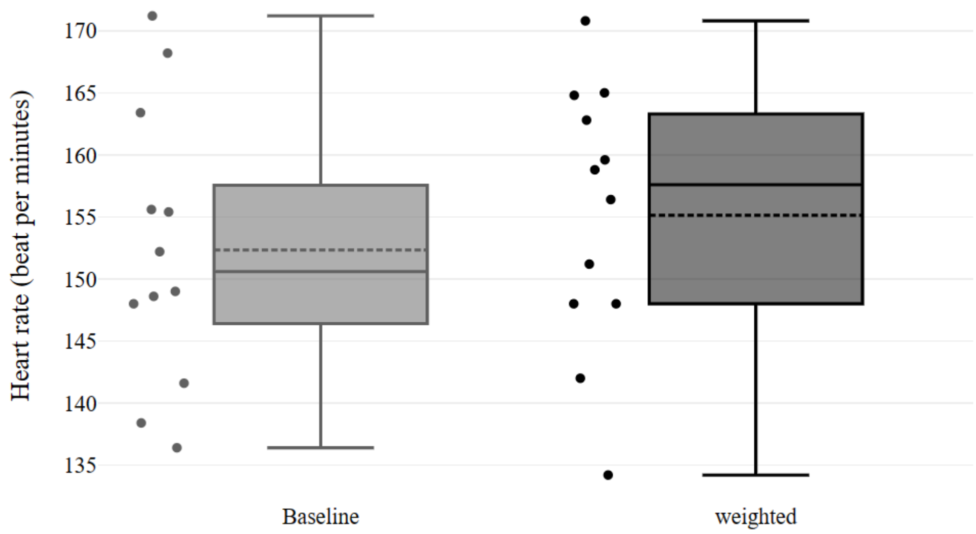


Figure 2.4.3 Mean heart rated frequency for the 12 participants during the 30-minute run with two shoe conditions

2.1.4.4 DISCUSSION

In this study, an additional mass of up to 400 g was inserted into the shoe, which is approximately half the mass of a typical human foot (de Leva, 1996). However, our findings show that adding mass did not alter step frequency or contact time during running. These results are by Rodrigo-Carranza et al. (2020) and Taylor et al. (1980), who found that additional mass on the foot by +100 g and on the back by + 20 kg did not affect spatiotemporal parameters during running. Specifically, in our current study, a Weber fraction set up to 1.71 did not significantly affect contact time and step frequency in comparison to previous sections (Section 1, Section 2, and Section 3) where the Weber fraction was 0.53. Additionally, we found that the mean heart rate significantly increased from the baseline. One of the main differences between our current study and the other two studies is the performance on two different days. Adaptations in stride frequency and contact time may affect the rate of force production, the length, and the velocities of muscle contraction, thereby impacting energetic costs (Barclay, 1994; Kram and Taylor, 1990). Such a change in energetic cost could be seen as an increase in heart rate frequency when the additional mass is 400 g. In general, changes in stride frequency and contact time may potentially lead to alterations in leg stiffness (Cavagna et al., 1997), and in combination with altered neural activation, this can affect the storage of elastic energy during landing and its reutilization during the subsequent push-off (Bonacci et al., 2009). According to the findings of De Ruiter et al. (2014), an individual's optimal stride frequency (and contact time) could simply be determined by measuring heart rate. Therefore, a possible explanation that may account for the lack of change in stride frequency and contact time is that participants ran with weighted shoes (633 g) close to their optimal stride frequency and contact time, at which energetic cost is minimal (Hunter and Smith, 2007) (very small changes in heart rate between the two conditions). Our participants may have selected a near-optimal stride frequency (Morgan et al., 1994) and contact time through a process of self-optimization.

2.1.4.5 CONCLUSION

This study aimed to evaluate and compare spatial-temporal parameters such as stride frequency and contact time in two conditions: running with a baseline shoe (total mass 250 g per shoe) and running with the same shoe but with additional mass (total mass of 600 g per shoe). Participants did not show any differences in step frequency or contact time in the two conditions. However, an increase in heart rate has been observed while running with the weighted shoe in comparison to baseline shoe. It is probable that participants ran with the weighted shoe close to their optimal stride frequency and contact time to minimize energetic cost.

2.1.5 SECTION 5

Additional Shoe Mass and Perception (I)

This Paper was published in the Journal of American Podiatric Medical Association: Keshvari, B., Alevras, S., & Senner, V. (2023). Sensory Perception of Varied Shoe Masses in Running. Journal of the American Podiatric Medical Association, 113(1), 21-229. https://doi.org/10.7547/21-229

2.1.5.1 INTRODUCTION

When selecting a running shoe, different features including comfort, (Clinghan et al., 2008; Tay et al., 2017), cushioning (Enke et al. 2009), brand (Branthwaite et al. 2009), and color (Trinkaus et al. 1991) influence the decisions of shoe consumers. Some studies have determined that shoe mass also ranks as one of the main criteria in choosing running shoes among different levels of runners (Honert et al., 2020; Kong et al., 2010). This might be because of the common perception that a lighter shoe will increase running efficiency, although recent research has produced mixed results in this respect (Bonacci et al., 2016; Cheung et al 2016; Divert et al., 2008; Franz et al., 2012; Hoogkamer et al., 2012: Fuller et al., 2016; Fuller et al., 2014). The perception of mass was first determined by Weber who in 1834 differentiated between static mass perception through touch, and active mass perception, which incorporates tactile and muscular components (Weber, 1834). In addition, he determined that the smallest difference in mass that can be detected is proportional to the mass of the object (Weber fraction = smallest difference: object mass). How the hand perceives changes in mass (based on touch and muscular components) has since been extensively studied. Modern experiments show that the Weber fraction ratio for hand touch alone is higher than when the object is lifted with the hand. (Feyzabadi et al., 2013; Ross et al., 2010).

In contrast, relatively little is known about how the foot perceives mass. Some early studies of lower limb mass perception were performed with masses cantilevered away from the limb (Hajnal et al., 2007; Donn et al., 1989). Recent findings show that when participants wore running shoes and were asked to evaluate their relative masses with movements while walking and jumping, the subjects were poor at perceiving mass across a range of common running shoe masses (~220 – 360 g) (Slade et al., 2014; Hausler et al., 2016). Saxton et al. (2020) investigated the perceived mass of weighted running shoes while running. Their findings show that wear time and baseline shoes (BSs) may influence mass perception by the lower limb. In another study by Greenya et al. (2014), the learning effect was measured in freely chosen actions such as walking, jumping, and standing over 60 seconds. Participants were asked to repeat the task for a second time, but their perception of shoe mass did not appear to improve with practice. Because of the diversity of independent variables and their effects on somatosensory shoe mass perception, more studies are needed to improve our understanding of mass perception through the feet. Therefore, the aims of this study are, first, to assess how accurately subjects perceive additional shoe mass during running, and second, to investigate the potential learning effect of shoe mass discrimination during running.

2.1.5.2 METHODS

Participants. Twenty-two male participants age 24.4 (± 3.2) years and body weight 77. 9 (± 8.6) kg were recruited from the Munich University Sports Center in Germany. Six months before the time of testing, participants were required to be injury-free. Before the experiment, they had to give written informed consent. The written informed consent form refers to the confidentiality of the objectives and study risks, and includes data privacy. Moreover, this form ensures that subjects are free at any time to stop participating in this research without providing any reasons and without incurring penalties. This research was conducted according to the ethical standards of World Medical Association Declaration of Helsinki (2013).

Tools of Study: Four sizes of running shoes ranging from 42 to 45 were selected (Victory Performance; Deichmann GmbH, Essen, Germany). Each shoe size was categorized as one BS (without additional mass) and shoes with four additional masses: +50 g (shoe I = 1.18 × BS), +150 g (shoe II = 1.55 × BS), +250 g (shoe III = 1.92 × BS), and +315 g (shoe IV = 2.16 × BS). The average weight of shoes in different sizes is 283 ± 21 g and the weight deviation for each shoe size is ±3 g.
To increase the shoe mass, lead tape was attached to the fore and rear parts of the shoes. However, the center of mass for each shoe was kept constant during this procedure. The experiment was conducted as a blind test by partially concealing the 20 shoes with black tape (see Figure 2.5.1).

a. **b.** **c.**

Figure 2.5.1 Procedure of weighting shoe sample in the experiment **a.** An Indoor running shoe (baseline shoe) **b.** Lead tape added to fore and rear parts **c.** Concealed with the black tape.

Study Procedure: Initially, participants warmed up with the BS on the treadmill for the first 10 min of the experiment. Subjects also defined their desired velocity for the experiment under blind test conditions aided by the experimenter. Next, participants ran on the treadmill (H/p/cosmos Arsalis, 2020) for 2 min with BS, and then they continued for another 2 min with a random pair of weighted shoes. After that, they were allowed to take a rest for 1 min. For comparison, all possible pair tests with the BS were repeated according to this process. After completing all possible pair tests with BSs, subjects were given up to 1 hour until they felt ready to start the second session following the same procedure as in the first session. In session 2, participants ran with the preferred velocity selected in the first session. After completing the first session, subjects were given up to 1 hour until they felt ready to start the second session following the same procedure as in the first session.

Methods of Data Collection and Analysis: After each pair of tests with BS, participants responded to a comparative question "Is the second shoe perceived to be heavier?" The answer is binary, where yes means " the second shoe was perceived to be heavier" and no means "the second shoe was not perceived to be heavier." For further statistical analysis, correct responses were converted to 1 and incorrect responses were converted to 0. A mixed-effect logistic regression with and without interaction effect was used in SAS 9.4 (SAS Institute Inc, Cary, North Carolina). Several steps were taken to prevent participants from perceiving indirect information about the shoes. Participants were asked not to touch or handle the shoes at any stage during the test of mass perception through the feet. The experimenters put the shoes on the subjects' feet, laced them, and removed them after the test. Participants were not allowed to walk or jump in their shoes. They began running on the treadmill immediately after lacing. These procedures prevented the participants from gaining any perceptual information about shoe mass that could confound the perceptions from simply wearing the shoes. In addition, to minimize possible tactile cue differences, participants were provided with similar socks (45 % polypropylene, 35 % cotton, and 20 % polyamide).

2.1.5.3 RESULTS

The effect of two independent variables, mass, and session, on perceived mass (dependent variable) among 22 subjects was investigated with mixed-effect logistic regression with and without interaction. In the mixed effect logistic regression without interaction, mass had a significant effect on perceived mass ($F_{4,193} = 10.66$, $p < .0001$), whereas session did not ($F_{1,193} = 1.06$, $p = .30$). Moreover, the model with the interaction effect showed that session and interaction effect (mass × session) were not significant ($p > .005$).
In addition, a post hoc test (ie, the Dunnett-Hsu test) was used to compare all weighted shoes with the BS. According to Table 2.5.1, the test comparisons between the BS and other weighted shoes show a significant difference (at an alpha level of 0.05), except for shoe I in comparison with the BS. The results show that the odds ratios when comparing the BS and weighted shoes are higher than 1 (see Figure 2.5.2) except for the BS versus shoe I. Accordingly, there are significant differences between the BS and weighted shoes but not shoe I (see Table 2.5.1).

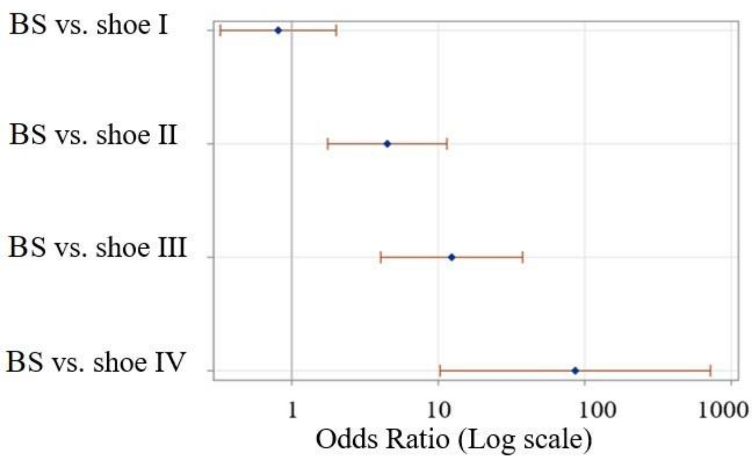


Figure 2.5.2 Odds ratio and all pair tests with baseline shoe

Table 2.5.1 Multiple comparisons of all Weighted shoes with the Baseline shoe

Label	Es	SD	AP	AL	AU
G1	- 0.21	0.46	.97	- 1.36	0.93
G2	1.50	0.47	.007	0.31	2.68
G3	2.51	0.56	<.0001	1.10	3.92
G4	4.45	1.07	.0002	1.77	7.13

G1 =BS vs. Shoe I, G2 =BS vs. Shoe II, G3 =BS vs. Shoe III, G4 =BS vs. Shoe IV
Es = Estimated; SD = standard deviation ; AP = Adjusted P-value; AL = adjusted low; AU = adjusted upper

The accuracy in session 1 for shoes I, II, III, and IV in comparison with the BS are 31 %,72 %, 86 %, and 100 %, respectively, whereas in the second session, the values are 31%, 68 %, 86 %, and 95%. Repeating the task did not affect the perception of shoe mass significantly. The total drop in accuracy from the first session to the second session across all five pair tests shows that there is no learning effect. The accuracy of shoe mass perception dropped 2% from session 1 (72 %) to session 2 (70 %). In addition, the insignificant $P = 0.30$ determined that there is no effect of the session on perceived shoe mass (see Figure 2.5.3).

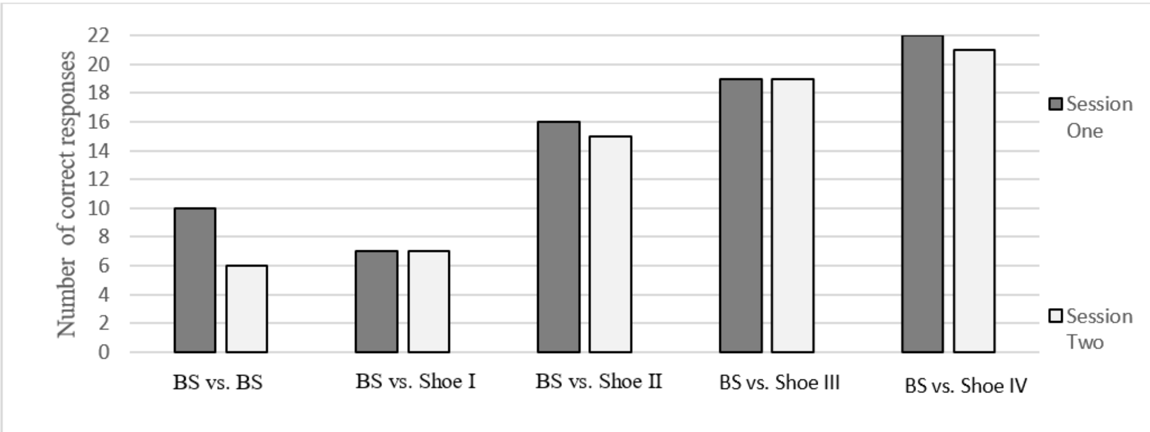


Figure 2.5.3 Number of subjects correctly perceived additional mass in all possible pair tests with baseline shoe (n = 22)

2.1.5.4 DISCUSSION

In our study, the sensory threshold of perceived mass is distinguished by the additional weight of +150 g ($P < .05$). An odds ratio higher than 1 only occurred when the additional shoe mass was greater and equal to +150 g. In our study—based on the Weber law—the just-noticeable difference (JND) is 150 g and the Weber fraction is equal to 0.53 (150: 283). Several factors have to be considered when relating the JND and Weber fraction measurements to results presented in the literature. Slade et al. (2014) determined that an increase of 140 g is the sensory threshold (JND) an6d the Weber fraction is 0.64 (the initial shoe mass is 220 g). The difference in the JND may be attributable to the methodologic difference in the original shoe masses of both studies, which is approximately 63 g. This difference may have affected the subjective threshold, where the Weber fraction decreases when the mass increases up to 200 g (Ross and Brodie et al. 1987). Findings of Hausler et al. (2016) determined that a JND of 100 g is required for the foot to consistently (and accurately) perceive differences between weighted running shoes. In addition to the methodologic difference of origin of shoe mass, a “sex-specific difference“may also have increased the foot’s activation threshold. Moreover, other factors that led to our results were the use of the BS and binary questionnaire. Runners are used to their shoes; thus, using the BS for a pair comparison with weighted shoes can reduce all potential biases (Saxton et al. 2020). The visual analog scale (VAS) was a main tool in past related studies (Slade et al., 2014; Hausler et al., 2016; Saxton et al., 2020). However, the complexity of this measure (ie, VAS) can influence the reliability of assessments (Hoerzer et al., 2016; Mills et al., 2010). In other words, the number of individuals who reliably assessed the footwear could be increased by reducing the complexity of the VAS measure to simple binary yes or no questions.

Another part of our findings determined that foot perception of shoe mass does not improve with practice. Our findings are in line with those of Greenya et al. (2014) In our study, repeating the sessions does not show any effect on perceived mass. The accuracy, from 72 % in session 1, dropped to 70 % in session 2. Greenya et al. (2014) determined that accuracy was 93% in the first trial compared with 92 % in the second trial for the hand test. They also showed that the accuracy of the feet did not differ significantly.

The findings from some of the aforementioned studies (Slade et al., 2014; Hausler et al., 2016; Saxton et al., 2020; Greenya et al., 2014) also determined that hands have a higher sensitivity—concerning the additional mass—than feet. While moving or holding mass by hand, both touch and muscle contribute to the perception of mass. In our study, the weight of the shoe could probably not be perceived by the touch sense in the skin of the feet. Thus, the shoe mass is mostly perceived with the degree of tension in the muscle when lifting weight (Brodie and Ross 1984). We speculate that acceleration of the feet in the swing phase of running influences the Golgi tendon organ and muscle spindle. By inserting up to 150 g additional mass, the Golgi tendon organ, which is a cluster of sensory receptors, will be stretched and depolarized significantly in the swing phase and send the signals through Ib-afferent synapses to the brain cerebellum and cerebral cortex (Parent et al., 1996). The sensitivity of tendon organ receptors to changes in muscle force suggests that the discharges arising from these receptors could provide the basis for judgments of force (Houk et al., 1980; Jones, 1986). This judgment may be related not only to perceived heaviness, as reported by Brooks et al. (2013) but also to perceived mass.

In addition, we generated (in early steps: pilot study) a running gait model using a multibody system simulation in the Simpack software to calculate the knee moment while running. The calculation of the knee joint moment in the swing phase in the sagittal plane—and with a velocity of 2.6 m/sec—is increased by 7.5% when additional shoe mass is 150 g (in comparison to the BS). This increase in knee moment (product of moment of inertia and angular acceleration) could contribute to the threshold of mass perception in running in a major way. Future study is necessary to investigate such a simulation, and also the relationship between joint kinematics and kinetic variables of runners and their perception of mass with the BS and shoe II (150 g).

2.1.5.5 CONCLUSION

In our study, we developed a new methodologic framework for perceiving shoe mass. An increase of 150 g is the JND among other weighted shoes and the Weber fraction is equal to 0.53 (150 :283 g). The learning effect did not improve by repeating the task in two sessions on the same day. The accuracy of mass perception while running drops 2% from session 1 to session 2. Future study is necessary to investigate the relationship between kinematics and kinetic variables and perception of mass while using the BS and shoe 2 (150 g).

2.1.6 SECTION 6

Additional Shoe Mass and Perception (II)

2.1.6.1 BACKGROUND

Perception of additional shoe mass during short-term running was investigated in the previous section (Section 5). The current section (Section 6) is an extension of that study to evaluate the effect of additional mass on perception in longer-term running.

2.1.6.2 METHODS

In total, 21 male runners (24.5 years old (± 3.1) and 77.1 kg (±7.9)) participated voluntarily. Four sizes of running shoes ranging from EU42 to 45 were selected. (*Shoe model*: Victory Performance, *company name*: Deichmann GmbH, Germany). Each shoe size was categorized as one Baseline shoe (without additional mass) and the other shoes were fitted with four additional masses: +50g (Shoe I = 1.18 × BS), + 150g (Shoe II = 1.55 × BS), +250g (Shoe III = 1.92 × BS) and + 315g (Shoe IV = 2.16 × BS). The average masses of shoes in the different sizes were 283g (±21g) and the mass deviation for each shoe size was (±3g). All wore identical socks – Falke KGaA – (45% Polypropylene, 35% cotton, 20% Polyamide) to minimize sensation variation. To find the most comfortable size, all subjects were asked to try alternative sizes. Participants ran with their desired velocity which has been explained in Chapter One. Participants ran on a motorized treadmill with the BS for 2 minutes. Immediately after this, the participants changed to one of the weighted pairs of shoes and then ran for 8 minutes. Afterward, they put on the BS and ran for 2 minutes before putting on another pair of weighted shoes and running for 8 minutes. They repeated this cycle for all weighted shoes and the BS (see Table 2.6.1). Participants took 100 seconds of rest after tests 2, 4, 6, and 8. Borg's RPE scale was completed five times in Tests 2, 4, 6, 8, and 10 (Table 1.6.1) and at the following intervals (including the warm-up duration): 20´30´´, 33´10´´, 45´50´´, 58´30´´, and 71´10´´ (in the final phase). The whole experiment including warm-up and rest duration lasted 73´20´´.

Table 2.6.1 Running with different shoe masses during the two-day testing protocol (example with random order for weighted shoes)

#Test	ID	Running Duration	TNCS**	#Test *	ID	Running duration	Rest
1	BS	2min	60 seconds	*2*	shoe 2	8min	100 seconds
3	BS	2min	60 seconds	4	shoe 4	8min	100 seconds
5	BS	2min	60 seconds	*6*	shoe 3	8min	100 seconds
7	BS	2min	60 seconds	8	shoe 1	8min	100 seconds
9	BS	2min	60 seconds	*10*	BS	8min	-

After each paired tests, participants responded to a comparative binary question, 'Is the second shoe perceived to be heavier?'The answer is binary, where "Yes" means 'the second shoe was perceived to be heavier' and "No" means 'the second shoe was not perceived to be heavier'. This question has been fully asked 5 times during all paired tests with BS. For further statistical analysis, correct responses were coded as '1' while incorrect responses were coded as '0'.

2.1.6.3 RESULTS

Data from this section was analyzed using IBM SPSS Statistics (Version 26). The finding of the Cochran Test determined that there is a significant difference between pair tests as follows: x^2 (df = 4) = 16.5 p = 0.002. In addition, A McNemar pair test showed an increase of additional mass 250g and 315g differed significantly in comparison to baseline shoe with respective p value=0.039, and p value = 0.008.

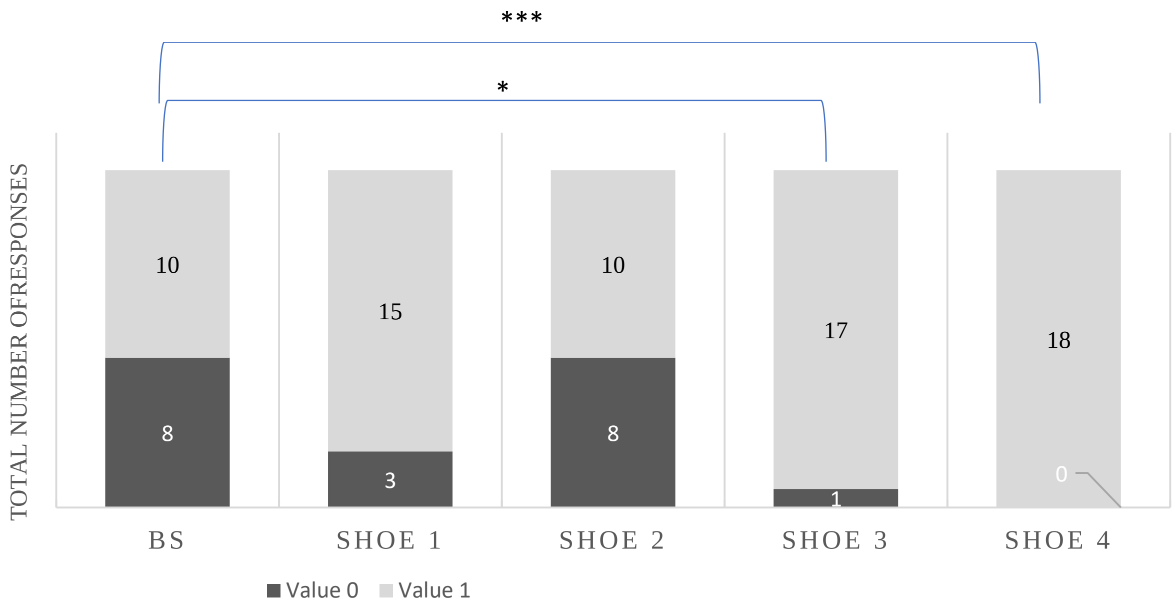


Figure 2.6.1 Total responses to the questionnaire (value 0 = incorrect and value 1=correct): (*: p < 0.05, *** : p < 0.001)

2.1.6.4 DISCUSSION

Our findings determine that an additional mass of up to +250 g, which is approximately the mass of the baseline shoe, represents the just noticeable difference (JND). This additional mass results in Weber's fraction equal to 0.88 (250 g : 283 g). In this study, the Weber fraction is higher than in previous sections or previous studies (Slade et al., 2014; Greenya et al., 2014; Saxton et al., 2020). This difference might be related to how sensory adaptation occurred, particularly in terms of kinetic adjustments, specifically regarding vertical ground reaction force (VGRF). To better understand these findings, it is necessary to refer to the findings of Section One in this chapter. Changes in VGRF in three phases were analyzed at the beginning of this chapter (Section 1). While running for 8 minutes, in phases 1 and 2 (as indicated by the results of vertical ground reaction force), the minimum additional mass that could influence VGRF was 150g. However, in phase 3, no significant difference (in VGRF) was found between the baseline and the weighted shoes. Since in this phase (phase 3), there were no observable adaptations in VGRF with additional mass (as indicated by the results of Section 1), it can be assumed that the participants' threshold level increased after adaptation in phases 1 and 2. An analogy can be made when a person dips their finger in a bowl filled with warm water; after some time, they will no longer be able to feel the same warmth because their senses have adapted to the stimulus. In this example, sensory adaptation occurs when sensory receptors can no longer perceive the stimulus due to continuous exposure. A similar process happened in the first and second phases, where participants perceived the minimum mass of 150 g and had kinetic adjustments within phases 1 and 2 (increasing VGRF, please see Section 1). Since there was no kinetic change (VGRF) in phase 3, we assume that the minimum additional mass threshold of 150 g (detected in phases 1 and 2) increased to 250 g by the end of the 8-minute run. This is the phase in which participants could only perceive 250 g but did (and could) not adjust their running kinetics for such a change in shoe mass.

2.1.6.5 CONCLUSION

Our findings determined that an increase in wearing time did affect the Just Noticiable Difference. Additional mass increased the Weber fraction to 0.88. Adaptation to the shoe mass and related kinetic adjustments occurred in Phase 1 and Phase 2, but not in Phase 3. Future studies should investigate whether or not the perception of additional shoe mass comes from the final phase (phase 3), or is a result of adjustment that occurred in phase 1 or phase 2.

2.2 SUMMARY

Shoe technology is intended to have positive effects on performance, comfort, and injury risk. Understanding footwear construction and its implications for running biomechanics can improve clinical reasoning and communication with patients or athletes when selecting appropriate running footwear. Running shoe construction has undergone significant changes over the past 55 years, with particularly rapid advancements in the last decade (Davis, 2014). This research aims to investigate the effects of additional shoe mass on kinetic, joint motion, spatiotemporal parameters, and perception, contributing to a better understanding of human interactions when subjected to such perturbations. I. The findings in the first section indicate that the addition of a minimum shoe mass of 150 g results in an increase in vertical ground reaction force (VGRF) and impulse. Specifically, in our study, during 8 minutes of running with a Weber fraction of 0.53, we observed a 1.86 % increase in VGRF. This increase in VGRF with an additional 150 g of mass was significant within the first and fourth minutes of running. II. In the second section of our research, we investigated the electromyographic (EMG) activity of runners when wearing additional masses (+150 g and +315 g) on their running shoes, comparing it to when they were running with baseline shoes. EMG recordings provide an additional source of information for cognitive-behavioral processing, aiding our understanding of added mass intervention studies. Our findings indicate a significant increase in the mean and integrated EMG activities during the swing and stance phases with the weighted shoes. This increase in muscle activity, involving the Tibialis Anterior, Gastrocnemius lateralis, and Gastrocnemius medialis, occurred when the Weber fraction was 0.58 (165 g: 283 g), corresponding to a total shoe mass of 433 g. III. In the third section, we measured peak knee flexion/extension in the swing and stance phases while wearing baseline shoes and weighted shoes (+50 g, 150 g, and 315 g). Our findings showed no significant difference between baseline shoes and weighted shoes in the stance phase. However, a significant difference was observed between baseline shoes and weighted shoes (+150 g and +315 g) only in the swing phase. IV. In the fourth section, we investigated the effect of additional shoe mass on temporal parameters during a 30-minute running performance conducted on two different days. We evaluated two variables, contact time and stride frequency, at five-time points, starting at the 3rd minute and continuing with 7-minute increments. Our findings indicated that running with shoes that had an additional mass of 400 g (total mass of 633 g) did not show significant differences in either variable when compared to baseline shoes (originally 233 g). V. In sections 5 and 6, we explored the perception of shoe mass in two subjective measurements: short-term (2 minutes) and long-term running (8 minutes). In both measurements, paired tests of additional shoe masses (+50 g, +150 g, +250 g, and +315 g) were compared to baseline shoes. Our findings revealed that in short-term running, the just noticeable difference (JND) was 150 g, corresponding to a Weber fraction of 0.53. However, during long-term running (8 minutes), the JND increased to 250 g, with a Weber fraction of 0.88. The addition of a minimum mass of 150g in the first 2 minutes of running was associated with a 2 % increase in vertical ground reaction force (VGRF) and increased muscle activity (see section 2). Joint motion data showed no changes in knee angles during the short-term run (2 minutes) in the stance phase and negligible changes during the long-term run (8 minutes). Moreover, temporal parameters, including stride frequency and contact time, remained unchanged.

Our findings suggest that when the total shoe mass exceeds 400g, kinetic variables begin to change while joint motion and temporal parameters remain relatively constant. These findings sought to same results in the previous studies that have used skin or shoe-mounted markers to measure limb motion while manipulating shoe inserts, yielding similar results (Eng and Pierrynowski, 1994; Nawoczenski et al., 1995). Additionally, Wilson et al. (1996) proposed a concept called the 'minimal resistance movement path' for lower extremity joints, suggesting that skeletal movement in running changes minimally when altering shoe or insert conditions. Nigg et al. (2017) found that such interventions usually fail to produce significant changes in joint kinematics. Instead, it seems that the neuromuscular system tends to maintain a specific, individual kinematic pattern, referred to as the 'habitual joint (motion) path,' guided by an individual's anatomy and passive tissue properties (Trudeau et al. 2019). According to our overall findings, we report that the mass intervention did not change the kinematics of the knee joint (at least in the stance phase) nor temporal parameters but did change some kinetic variables such as mean EMG muscle activity and vertical ground reaction force. The neuromuscular system seems to prefer a specific, individual kinematic pattern in which when external interventions interfere with this pattern, the neuromuscular system counteracts, e.g. with modified muscle activation and VGRF, such that the original movement pattern is preserved.

The following part of the conclusion should be taken cautiously and requires further research: Regarding those neuromuscular alterations (mentioned in the last paragraph), the additional mass to a shoe has the potential to modify both the length and tension of muscle fibers. Muscle spindles, responsible for detecting shifts in muscle length, may be influenced by this added mass, potentially reshaping the base length and stretch attributes of muscles. Consequently, this can affect the firing rate of muscle spindle afferents. Furthermore, the supplementary mass might also affect the rate and scope of sensory adaptation, which, in turn, can influence the proprioceptive feedback provided by muscle spindles. The information conveyed through proprioception by these muscle spindles plays a pivotal role in governing motor control and coordination. It is crucial to recognize that the particular consequences of introducing extra mass to muscle spindles can fluctuate based on variables such as the location and distribution of this additional mass, the type of muscle fibers involved, and the nature of the movement or task being executed.

Additionally, the central nervous system assumes a vital role in assimilating sensory input, including proprioceptive feedback, and subsequently regulating motor responses. Therefore, it can be speculated the addition of 150 g of mass could potentially exert an influence on muscle activity, amplifying the tension and force generated by the muscles. This escalated tension is then transmitted to the tendons, resulting in heightened activation of the Golgi tendon organs. These Golgi tendon organs possess sensitivity to alterations in tendon tension, and the additional mass can induce increased activation of these sensory organs. This sensory information received from Golgi tendon organs not only governs muscle force but also contributes significantly to motor control. Furthermore, additional mass can stimulate adaptations in the sensitivity and responsiveness of Golgi tendon organs over time. That would be consistent with our finding that during extended periods of running, the threshold for the perceived just noticeable difference (JND) increased to 250 g (total mass of 536 g), as opposed to the JND of 150 g (total mass of 433 g) observed during shorter running sessions.

Consequently, forthcoming research endeavors should be directed toward exploring the ramifications of introducing extra mass during prolonged running sessions, as well as its potential effects on sensory perception and neuromuscular responses.

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CHAPTER TWO

(CUSHIONING)

3.1 AIM of the STUDY

The early version of a running shoe with a rubber sole was presented in the 1870s (known as the *plimsoll*) which promoted cushioning as a main consideration for runners. Nowadays, various midsoles and insoles are on the #market with different materials, designs, and technologies to reduce injuries through cushioning. The Oxford English Dictionary (1989) defines a cushion as “A case of cloth, silk, etc. stuffed with some soft, elastic material used to give support or ease to the body when sitting, reclining or kneeling”. The Random House Webster’s College Dictionary (1995) defines cushioning as “…anything similar in form or function, such as a pad to prevent excessive pressure […] something to absorb or counteract shock, jar, or jolt,” or “to lessen or soften the effects of; to cushion a blow”. Running exposes the body to repeated collisions resulting from rapid decelerations at initial ground contact. These cumulative effects are considered to increase the likelihood of overuse running injuries. (Yang et al., 2012). However, a reduction of skeletal shock waves by shock-absorbing shoe sole material has resulted in clinical improvements in terms of injury prevention and reduction (Voloshin et al., 1981 and 1982; Warden et al., 2014).

The significance of cushioning in mitigating sports injuries underscores the necessity to explore individuals' perceptual responses to varying cushioning conditions. An array of methodological approaches, encompassing Likert scales, visual analog scales (VAS), and binary questionnaires, has been employed to examine the perceived qualities of cushioning (Isherwood et al., 2021; Pisciotta et al., 2018). Similarly, these techniques have been utilized in assessing comfort perceptions related to sports equipment (Böhm et al., 2008). Of particular note, binary questionnaires, relying on paired comparisons, present respondents with two options per judgment, offering a nuanced exploration of participants' preferences. This methodological framework facilitates the systematic evaluation of stimuli, enabling participants to select the item that best aligns with specified choice criteria. The binary questionnaire method serves as a valuable tool in elucidating the participant's ability to discern nuances in cushioning alterations, denoted as just noticeable differences (JND). Detection of such perceptual thresholds contributes significantly to the comprehension of muscular adaptation mechanisms following external stimuli (Robbins et al., 1989; Robbins et al., 1990). Historical research endeavors have extensively investigated the mechanical attributes of running shoe midsoles and insoles through diverse methodologies. Notably, durometer measurements, assessing midsole and insole hardness via resistance to indentation, have been conducted in accordance with ASTM D2240-05 standards (2005). Functional mechanical impact assessments, simulating foot strikes during running, have been employed to gauge running shoe impact attenuation (Frederick et al., 1984; Henning et al., 1993; Milani et al., 1997; Maropoulos et al., 2017). Mechanical testing data further serves as a predictive tool to discern potential correlations between material characteristics and perceived cushioning (Pisciotta & Shorten, 2018; Salzano et al., 2021; Isherwood et al., 2021).

Post-1990, the evolution of investigative methodologies witnessed the introduction of intricate biomechanical analyses. Specifically, the assessment of pressure distribution within the shoe emerged as a critical determinant for preventing orthopedic and biomechanical injuries, illuminating the nuanced interplay between body and material comfort (Chen et al., 1994; Miller et al., 2000; Mündermann et al., 2001; Kryger et al., 2017). The efficacy of this measurement in discerning differentiating characteristics of shoe cushioning establishes it as a promising avenue for the nuanced prescription of running shoes. The comprehensive integration of cushioning perception, mechanical testing, and biomechanical measurement has been notably lacking in prior investigations. While shoes inherently function as mechanical entities, their transformation into biomechanical systems during interaction with athletes underscores the imperative for a nuanced understanding from diverse perspectives. A dearth of studies has systematically addressed this triad of considerations. For example, the work by Isherwood et al. (2021) represents a commendable exception where a concerted effort was made to encompass all three dimensions. Conversely, some studies have selectively addressed elements of this triad. Notably, in the seminal works by Benno Nigg et al. (1987 and 2001), an insightful exploration into running biomechanics was conducted; however, the evaluative dimension of perception was regrettably omitted. This oversight led to a failure to elucidate whether runners were cognizant of changes, particularly perturbations in cushioning. Similarly, in another investigation by Pisciotta et al. (2018), a commendable inclusion of mechanical testing and subjective measurements for perception evaluation was observed, yet the absence of biomechanical testing limited the depth of insights into the holistic dynamics of sports footwear. This dissertation endeavors to bridge this gap by rigorously examining the interplay between mechanical attributes, biomechanical responses, and perceptual evaluations, thereby contributing to a more comprehensive understanding of the intricate dynamics within sports footwear.

The scarcity of research integrating all three fields is evident in the literature, highlighting the tangible necessity for such comprehensive investigations. This chapter is dedicated to advancing our comprehension of how athletes perceive and adapt to cushioning perturbations introduced in their footwear. This investigation will unfold across three distinct domains: subjective assessments and laboratory-based evaluations. In the initial section, the analysis focuses on the midsole of football boots, scrutinizing its influence on vertical ground reaction force. Other sections (2 and 3) delve into the exploration of running shoes, with a specific emphasis on the impact of diverse insole densities on running gait dynamics. The flowchart for this chapter is as follows:

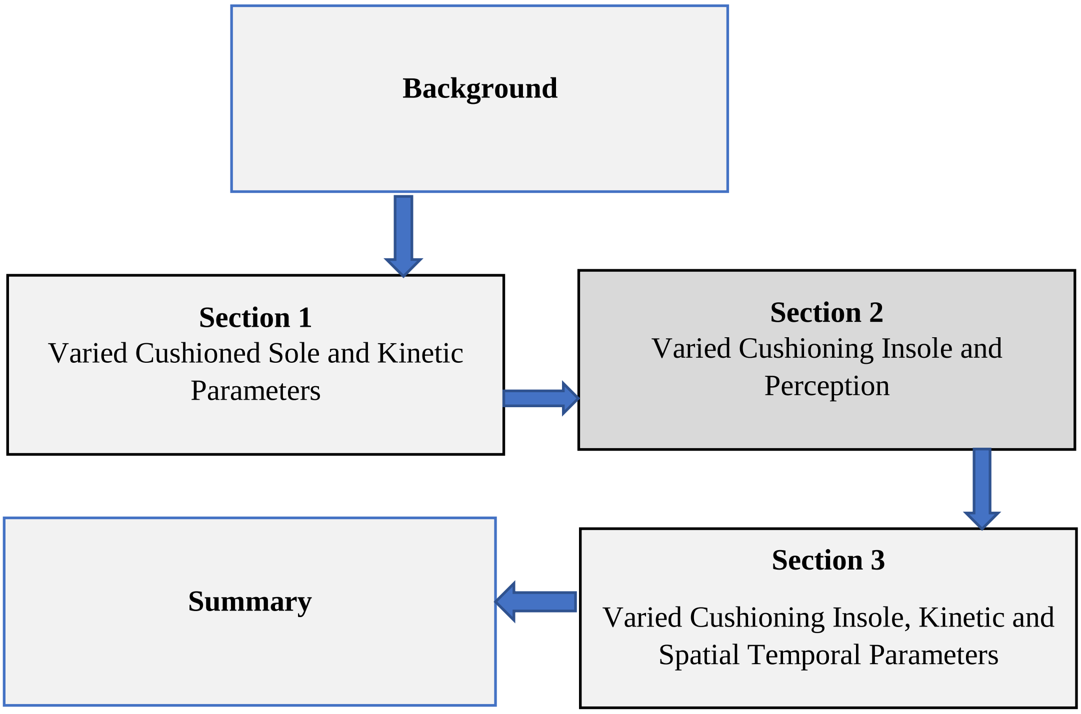


Figure 3.1. Flowchart of three sections used in this chapter. The dar gray box denotes published paper.

3.1.1 SECTION 1

Varied Cushioned Sole and Kinetic Parameters

3.1.1.1 INTRODUCTION

Mechanical loading on a human locomotor system in *sports activities* can be defined as constructive or destructive (Nigg BM, 1988). Evidence for destruction comes from the response of bone cells, depending greatly on excessive repetitive impacts (Cole et al., 1996; Eils et al., 2004; Meijer et al., 2006; Queen et al., 2008) with surface properties (Ford et al., 2006; Nigg et al., 1987), boot properties (Bentley et al., 2011; Smith et al., 2004), and movement types (Eils et al., 2010; Wong et al., 2007). This excessive destructive mechanical loading can cause micro or macro trauma and result in overuse injuries (Nigg BM, 1988; Clements et al., 2001; Kibler et al., 1992) which include around a quarter of all sports injuries (Yang et al., 2012).

A study of sport boots focusing on the midsole component revealed that highly cushioned running shoes increased the peak vertical ground reaction force (VGRF) more than normal running shoes during running (Kulmala et al., 2018). Nin et al. (2016) determined that forefoot peak VGRF in a 'soft shoe' was higher than in a 'hard shoe' during shot-block landing. In contrast, other studies have revealed that there is no effect of highly cushioned midsole shoes and normal running shoes on VGRF peaks (impact and active peaks) (Aminaka et al., 2018; Hannigan et al., 2020). Previous research has produced mixed results concerning the effect of the midsole on ground reaction forces, thereby requiring more studies to reach a common consensus. However, to our knowledge, in some sports where the midsole is equipped with a studded outsole, the effect of the midsole on peak VGRF has been poorly understood.

In football, boots are broadly classified as natural grass shoes (NGS) and artificial turf shoes (ATS). These are different in stud configuration, which alters peak VGRF (Queen et al., 2008, McGhie et al., 2013) and peak pressure (Orendurff et al., 2008). However, the boots in these studies are not only different in stud configuration, but midsole cushioning and other impact factors also vary. Since these studies cannot show that the differences occur due to midsole or outsole changes, more research is needed to investigate the effect of two midsole cushioning properties – ATS and NGS – while taking into account the configuration and characteristics of the outsole. In addition, another feature of football boots, namely a 'good fit', which is a prerequisite for boot comfort (Hagen et al., 2010), can be influenced by different construction of boots (upper shape, boot lace, heel flare, heel cup) (Luximon et al., 2003). Paying less attention to good fit while comparing midsole cushioning, makes the final results ambiguous as to whether significant biomechanical changes or no changes to midsole hardness influence comfort in terms of good fit. It has been suggested that the comfort variable can be measured by a different type of questionnaire, such as a visual analogue scale (VAS) (Mills et al., 2010).

Football teams are more likely to frequently switch between pitch surfaces during training or in match venues. Following the development of artificial turf in football, the demand for this surface in training and competition has increased. A 12-year study by Fujitaka et al. (2017) showed that switching football surfaces from natural grass to artificial turf increases the incidence of injuries. Accordingly, Stiles et al. (2007) state that '…past studies showed that the increased use of artificial surfaces in place of natural surfaces in sport has led to a higher prevalence of overuse injuries (Nigg et al., 2003; Nigg et al., 1987)'. However, current research comparing natural grass with artificial turf generally finds that there is no significant difference in injury incidence and severity (Ekstrand et al., 2011; Fuller et al., 2007; Loughran et al., 2019). Mc Murtry et al. (2019) assessed injury incidence to quantify risks in different playing surfaces and determined that surface cannot account for cumulative injury mechanisms such as medial tibial stress syndrome, chronic ankle sprains, and cartilage degeneration. Reducing the magnitude of the external load (which affects internal structures) is one option for preventing overuse-type injuries. External load can be affected by a variety of boundaries, including surface. Girard et al. (2007) found that ‘harder surfaces (e.g. greenest)’ compared to ‘softer surfaces (e.g. clay)’ increased plantar loading in specific movements. McMurtry et al. (2019) revealed that in football, VGRF parameters differed between two surfaces: natural and artificial grass (there is no information available about the compliant level of surfaces). However, their study included participants with prosthetic legs, which may exaggerate effect sizes in comparison to able-bodied participants. Other kinetic parameters such as peak pressure in concrete vs natural grass surfaces (Wang et al., 2012), peak pressure in natural grass vs artificial turf (Ford et al., 2006), and vertical peak acceleration in concrete vs woodchip (Boey et al., 2017) have determined that harder surfaces are correlated to the aforementioned kinetic parameters. In contrast, Weijie Fu et al. (2015) revealed that there is no difference in peak pressure in running movement comparing both concrete and grass surfaces nor in peak acceleration. Inconsistency of past results prevents researchers from reaching a consensus about the effect of surface stiffness on kinetic parameters in football, e.g. peak VGRF. To measure peak VGRF, some studies, Simonsen et al. (2000), Orloff et al. (2008), Katis et al. (2010), Blackburn et al. (2003), and Bencke et al. (2000), inserted force plates into the pitch to measure specific movements (i.e. cutting and kicking). However, inserting force plates under a *natural* football pitch was criticized by (Clarke et al., 2017; Akins et al., 2015) who are of the opinion that this results in changing soil characteristics. Thus, to avoid such a change, some researchers have used in-boot measurement systems in different sports as an alternative to force plates (Bentley et al., 2011, De Berardinis et al., 2021, Stöggl et al., 2017). The purpose of the present investigation was to characterize and compare VGRF on two different midsoles, two surfaces, and two specific football movements using an insole measurement system. Three hypotheses were investigated:

1) Midsole conditions have a significant influence on VGRF

2) Surface conditions have a significant influence on VGRF

3) Different football-specific movements have a significant influence on VGRF

3.1.1.2 METHODS

Participants. Seventeen male participants were recruited from football clubs (third and fourth Bundesliga) in Munich (Germany) and the surrounding area based on having had a minimum of 5 years experience playing football but were excluded if they had had a lower extremity musculoskeletal injury within the past six months or a lower extremity reconstructive operation within the past three years. The mean age, height, and weight were 26 (±6) years old, 181(±7) cm, and 79 (±11) kg. Participants gave written, informed consent before the experiment. The consent form declares the confidentiality of the objectives, risks of the study, and protection of personal data through appropriate procedures for anonymization. In addition, the consent form assures participants they are free to withdraw from the research at any time without giving a reason and without penalty for not taking part. This research was conducted according to the ethical standards of the World Medical Association Declaration of Helsinki (2013).

Protocol of the study:

Footwear. Two boots of different football classifications (see Table 3.1.1), *Adidas Neoride III* (ATS), and *Nike Hypervenom, (NGS)* were chosen based on the high demand for these boots at the time of the experiment. Before the start of the study, a Type A shore durometer (Kern & Sohn GmbH, Germany) was used to determine the material hardness at the forefoot and rearfoot regions of the boots. Each reading was taken as the average of five measurements at one location following Sterzing et al. (2013) (see Figure 3.1.1)

Table 3.1.1 Characteristics of the two boots' sole shore hardness used in the experiment.

	Shore hardness		*HT* (mm)*	*Mass (g)*	*FT** (mm)*
	Forefoot	*Heel*			
ATS	*58.5 (1.8)*	*53.4 (1.6)*	*200*	*320*	*100*
NGS	*82.4 (1.7)*	*72.4 (1.3)*	*100*	*238*	*50*

Note: Average Shore Hardness after five trials. HT: Heel Thickness, FT**: Forefoot thickness*

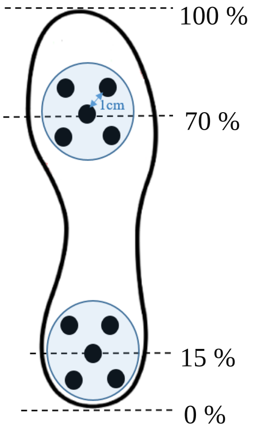
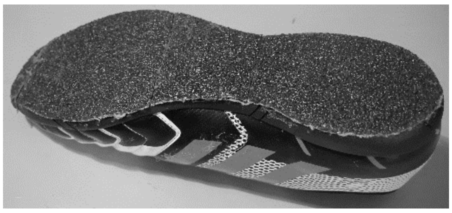


Figure 3.1.1 Shore hardness measurement locations using shore durometer.

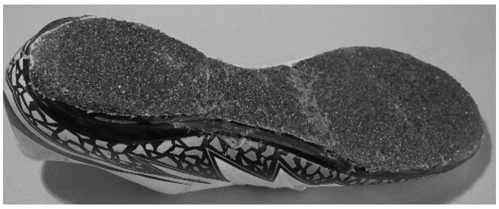
To ensure the outsole was identical, the studs of both boots were flattened and smoothed with a belt-grinding Machine. Abrasive paper (grit size, 80) was then glued to the outsole of the boots (see Figure 3.1.2). The grit size was chosen in a pilot test among three participants and the main criteria was to avoid slippage while performing sprinting, kicking, and cutting movements on natural grass and artificial turf.

2a ATS before flattening

2b NGS before flattening



2c ATS(after flattening)



2d NGS (after flattening)

Figure 3.1.2 Two football boots before and after flattening (ATS=artificial turf shoe, NGS=natural grass shoe).

Mechanical surface Testing. To distinguish the characteristics of the surfaces (resilience variable), a Light Falling Weight Deflectometer (LFWD) was used for every experimental test day on two common football pitches: artificial turf (sand/rubber infill, 100% Polypropylene) and Natural grass. The test determines the dynamic modulus of soil deformation in different ranges (ASTM E2583-07 (2015)). The LFWD consists of a steel loading plate with an accelerometer sensor, a falling weight, and an electronic measuring unit. The procedure involves dropping a 10 kg weight freely from a height of 72 cm, along the guide rod, to hit an installed dashpot unit in the middle of the steel loading plate. The loading plate has a diameter of 30cm and a thickness of 20 mm and weighs 15 kg (Tawfik et al., 2015). The dynamic soil resilience modulus (Evd) is determined as the ability of a material to absorb energy when it is deformed elastically and release that energy upon unloading. It is a measure of material stiffness and provides a means to analyze the stiffness of materials under different conditions, such as density and stress level. It is measured in MN/m^2 (MPa) using the following equation 1, German standard TP BF-StB (2003).

(1) $$Evd=1.5\ r\ \left(\frac{\sigma}{s}\right)$$

Where r = radius of the load plate, σ = stress under the load plate, s = mean settlement of the load plate. In our experiment, the dynamic soil resilience modulus of the artificial and natural surfaces was measured with the average of second and third drops of the four trials on a random spot in the experimental field.

Mechanical Midsole Characterization. A dynamic shock absorption test (Keshvari et al., 2020) was used for the measurement of cushioning properties. The pneumatic impactor device consists of a sphere with a 5 cm diameter and total weight of 4.3 kg. The machine operates by allowing the weight to fall onto the tested material from a defined height of 7 mm. A single-axis accelerometer sensor (range of – 50 g-Max to 50 g-Max) was attached to the impactor to quantify the shock absorption with a measuring frequency of 5000 Hz. The two locations on the boot midsoles tested by the impactor were in the forefoot and heel, following Sterzing et al. (2013) (see Figure 3.1.3). Twelve trials were conducted using the impactor but only trials two to ten were taken into account.

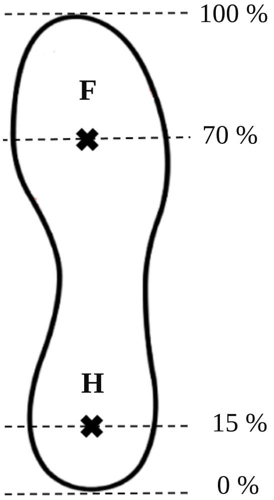
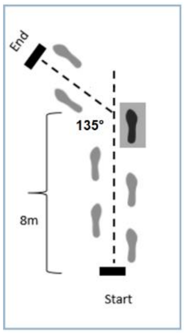


Figure 3.1.3 Schematic of impact test spots (F = forefoot, H = Heel).

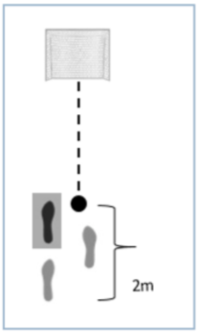
Material of the Study: Insole Sensor Pressure. To measure vertical force and plantar pressure distribution, the insole pressure sensor system Opengo (Moticon GmbH, Munich Germany) was used in this experiment. Opengo consists of 13 capacitive pressure sensors, covering more than half of the insole area (52%). The peak pressure is specified from 0 to 40Ncm^{-2} and also was computed at 50 Hz. No external devices or cables are needed to operate the system. To differentiate pressure in the plantar, 13 insole sensors were divided into 6 plantar sections. HxT= hallux and other toes, CFF = central forefoot, LMF = lateral midfoot, MMF = medial midfoot, LRF = lateral rearfoot, and MRF = medial rearfoot.

Main Test: Pre-Test. At the beginning of the experiment, all participants undertook a warm-up program (*FIFA 11*) (Nuhu et al., 2021), with their own private (desired) boot. *Step 1.* Participants randomly chose a pair of boots, NTS or AGS, into which the insole-pressure sensors were inserted with identical football socks. *Step 2.* They carried out two movements, cutting (135°) and penalty kick with two boots on one of the surfaces. For each condition, participants performed three trials; the first and second trial was to make players familiar with the boots and surfaces and the inserted insole, and the third trial was recorded and used for the measurement phase. *Step 3.* After performing the test on one of the surfaces (randomly), participants took up to two minutes rest and then followed the same procedure in step 2 but on another surface. As in *Step 2*, for each condition, participants performed three trials; the first and second to make players familiar with the boots and surfaces and the inserted insole, and the third recorded and used for the measurement phase.

In the cutting movement (135°), participants were asked to use an 'open' technique, which involves the athlete using the foot on the opposite side to the direction they want to turn. In this movement, the VGRF of the feet which changed direction was collected. In another movement, the penalty kick, the data of the supporting foot was collected during their kicking of the ball at goal (see Figure 3.1.4).



4a



4b

Figure 3.1.4 Schematic of two movements (measurement foot in black). **4a** Cutting 135° movements. **4b** Penalty kick.

Furthermore, the discomfort level of the feet based on *footwear fit*, which can potentially be caused by pressure-related skin lesions on the edge of the feet was measured using a Visual Analogue Scale (VAS, 100mm) after warm-up. Poor fit (= high discomfort) is denoted by 0, and best fit (=low discomfort) by 100.

Methodology of Data Analysis. Data of the participants (n = 17) on two surfaces, two boots, and two movements (N = 136) was analyzed with a repeated measure ANOVA (2×2×2) with an Alpha level= 0.05. The effect size was used for all three variables expressed as Cohen's d—0.2 denoting a minor effect, 0.5 denoting a moderate effect, and 0.8 denoting a major effect (Cohen, 1988).

3.1.1.3 RESULTS

Comparing peak vertical force. The repeated measures ANOVA (2×2×2) determined that cutting vs penalty kick, natural grass vs artificial turf, and finally, artificial turf boot vs natural grass boot differed significantly (see Figure 3.1.5). The effect size of three variables on peak force, movement, boot midsole, and surface are respectively 0.72, 0.78, and 0.28. There was no significant interaction effect between variables (P > 0.05).

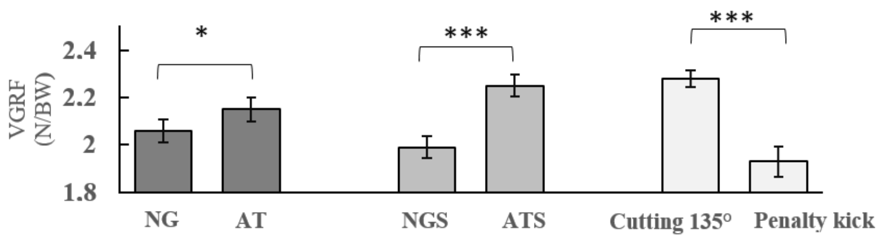
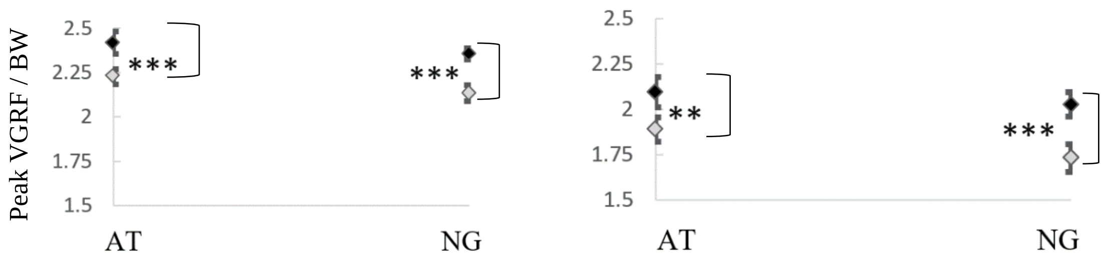
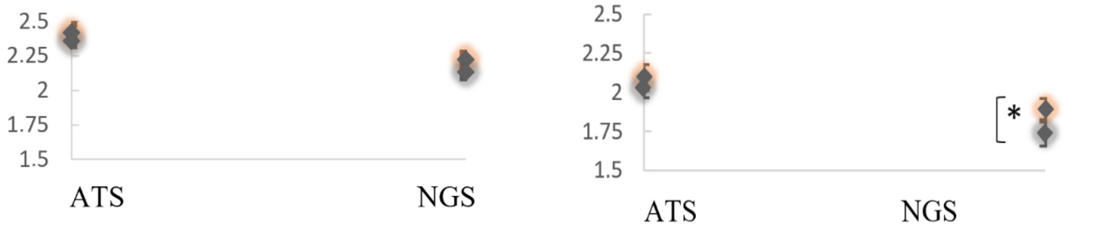


Figure 3.1.5 The peak force on different surfaces (NG and AT), different boots (NGS and ATS), and two movements (* = p < 0.05, *** = p < 0.001, AT= artificial turf, NG = natural grass, NGS = natural grass boot, ATS=artificial turf boot). Note: VGRF denotes vertical ground reaction force N/BW denotes body weight / newton. The standard error is based on SEM (+/-2).

The effect of the two different boots, ATS and NGS, on natural grass and artificial turf are shown in detail for each movement in Figure 3.1.6. The boots in all four conditions differed significantly from each other. The artificial turf boot in all four conditions had a significantly higher peak VGRF than the natural grass boot.



6a. Cutting (135°) **6b.** kicking



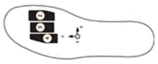
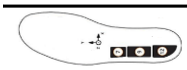
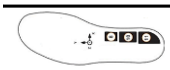
6c. Cutting (135°) **6d.** Penalty kick

Figure 3.1.6 Comparing peak vertical ground reaction force (VGRF) force in different conditions (p-value < 0.05=*, p-value < 0.01= **, p-value < 0.001 =***)

◆ **Artificial-Turf Shoe** ◇ **Natural-grass Shoe** ◆ **Artificial Turf** ◆ **Natural Grass**

However, peak VGRF on the two surfaces, NG and AT, only differed significantly when NGS was used. The supporting leg in the penalty kick had a higher peak VGRF on artificial turf in comparison to natural grass. To characterize pressure distribution, pressure mapping was classified into 6 sections (see Table 3.1.2). Peak pressure (average of both movement and surface) in ATS was significantly higher in the hallux, central forefoot, and lateral heel.

Table 3.1.2 Peak pressure (N/cm^2) of the plantar concerning the three variables.

	Movement		Surface		SHOES		**Effect size**	**SMEI** [8]
Plantar mapping	**C**[2]	**K**[3]	**AT**[4]	**NG**[5]	**ATS**[6]	**NGS**[7]		
 HxT	7.99 0.13[s]	5.79 *** 0.39	7.21 0.22	6.58 * 0.30	7.06 0.24	6.73 * 0.23	M=0.7 Sr=0.27 Sh=0.24	M, Sr; Sh
 CFF	9.60 0.22	6.20 *** 0.50	8.46 0.25	7.34 *** 0.40	8.28 0.30	7.52 *** 0.31	M=0.74 Sr=0.42 Sh=0.59	M, Sr; Sh
 LMF	6.31 0.24	7.02 0.49	7.18 0.27	6.16 * 0.36	6.58 0.26	6.76 0.28	Sr = 0.34	Sr
 MMF	7.75 0.45	4.98 * 0.54	5.99 0.47	6.74 0.44	6.11 0.44	6.63 0.40	Sr= 0.19 M= 0.59 Sr×M=0.28	Sr , M Sr×M
 LR	2.01 0.17	4.66 *** 0.40	3.18 0.32	3.49 0.26	3.93 0.27	2.74 *** 0.28	M=0.8 Sh=0.73	M, Sh
 MR	2.36 0.17	3.96 *** 0.37	2.9 0.28	3.33 0.24	3.32 0.31	3 0.23	M=0.52	M

[S] standard error, [1] different percentage of vertical GRF in each group. [2] C=cutting. [3] K=penalty kick. [4] AT=artifical surface. [5] NG=natural grass .[6] ATS= artificial turf shoe. [7] NGS= natural grass shoe. [8] SMEI=significant main effects and /or interactions. [9] M=movement. [10] Sr=surface. [11] Sh=boot. *p-value < 0.05, ***p-value < 0.001

Accordingly, peak pressure in AT was significantly higher in the hallux, central forefoot, and lateral midfoot. The comparison of peak pressure between the two movements determined that the peak pressure differed significantly in all sections except the medial forefoot. The central forefoot was affected by all three variables.

Soil hardness. The hardness of the natural and artificial surfaces was measured during the dynamic load plate test. The last two trials for each surface before the beginning of the experiment were collected. The mean and standard deviation of both surfaces' hardness over the five test days were 4.67 MPa (SD = 0.02) and 2.95 MPa (SD = 0.16), for AT and NG respectively. The Friedman non-parametric test revealed that there was a statistically significant difference between the two surfaces' dynamic soil resilience (E_{vd}), χ^2 (1) =10, p = 0.002. Artificial turf was consistently harder than natural grass during the experiment.

Midsole cushioning. Peak acceleration and energy loss (hysteresis) of peak impact force were measured for fore- and rear-foot sections in the ATS and NGS (see Table 3.1.3). Due to the thickness difference between the rear section of the two boots, there was a greater gap between peak acceleration and energy loss in comparison to the forefoot of the boots. In general, peak acceleration and energy loss in the NGS were higher than in the ATS.

Table 3.1.3 Midsole thickness and peak acceleration from 7 mm impactor drop.

	Forefoot		Rearfoot	
	Peak AC[1]	Energy loss(J)	Peak AC	Energy loss(J)
ATS[2]	23.8 *(1.61)*	0.44 (0.05)	11.2 (0.39)	0.31 (0.08)
NGS[4]	28.35 (0.53)	0.57 (0.04)	23.18 (0.47)	0.44 (0.04)

[1]=peak acceleration unit is g. [2]=artificial turf shoe. [3] =standard deviation. [4] =natural grass shoe.

Discomfort. The data of the visual Analogue Scale (100 mm) was analyzed with a non-parametric Friedman Test. The average discomfort for NGS and ATS, respectively, was 7.9 (Standard Error = 0.58) and 7.1 (Standard Error = 0.46). The results revealed that there was no significant difference in footwear fit between the two boots, χ^2 (1) = 1.66, p = 0.197.

3.1.1.4 DISCUSSION

The outcomes of our investigation reveal the significant influence of three independent variables—midsole characteristics, playing surface, and specific movement patterns—on peak vertical ground reaction force (VGRF). Specifically, the study first elucidates that football boots with softer midsoles, as exemplified by the artificial turf boot, generate a notably higher VGRF (2.25 BW) compared to their natural grass counterparts (1.95 BW), representing a 15% increase. This aligns with prior research by Hreljac (1998), indicating greater VGRF in soft midsoles (1.59 BW) compared to hard midsoles (1.36 BW) during landing in tennis shoes, reflecting a 14% increase. Our results are in line with Nin et al. (15), who determined that forefoot peak VGRF in a soft boot was higher than in a hard boot during shot-block landing in basketball. Our findings also support those of Kulmala et al. (2018), who determined that VGRF in a high-cushioned midsole boot (2.91 BW) was significantly greater than in a low-cushioned boot (2.85 BW). The strength of the relationship between this independent variable and VGRF was shown with an effect size of 0.78, which was close to a 'major' size effect. The results of our study indicate that wearing soft midsole boots can result in increasing VGRF peaks. In addition, by changing the NGS to ATS for both surfaces, the VGRF also increased in both movements (see Figure 3.1.5). A possible interpretation is that players need an optimum traction coefficient – rotational traction for cutting, and translational traction for kicking – to perform the movement successfully. Since the midsole of the ATS had higher cushioning and lower shore hardness than the NGS, participants generated greater VGRF through the ATS midsole to reach the sufficient traction coefficient, required for minimizing the slippage risk. According to the results for peak pressure, shown in Table 2.1.2, the central forefoot is the main contributor to peak pressure in both movements. This result also coincides with and is supported by that of Hennig and Sterzing (2010), who determined that peak pressure is greater in the central forefoot for different football boot models.

The perceived discomfort of these two boots' fit did not differ significantly. The discomfort factor rated in VAS was 7.9 (Standard Error = 0.58) and 7.1(Standard Error = 0.46) for ATS and NGS, respectively. Therefore, this boot attribution cannot be argued to have had any effect on kinetic parameters such as VGRF. We believe that boot discomfort (fit) should be assessed in all research with the related questionnaire to establish the potential effect of this variable on biomechanical parameters. Another finding of our study was that the natural grass with lower resilience modulus (softer ground) results in lower VGRF than artificial turf. Since peak acceleration is correlated to peak VGRF (Hennig et al., 1993; Janz et al., 2003; Neugebauer et al. 2014; Rowlands et al., 2012), our study corresponds to the results of (Boey et al., 2017) and (García-Pérez et al., 2014), who showed harder surface can increase peak acceleration. Additional analysis of plantar pressure determined that peak pressure in harder surface (AT) was significantly higher than NG (in certain sections such as HxT, CCF, and LMF). These results are in line with Tessutti et al. (2012) and Ford et al. (2006), who revealed that the softer surface shows lower peak pressure. In our study, switching between surfaces from natural grass to artificial turf found a significant increase in VGRF (4%). Changing football surfaces can occur within football programs, e.g. training or competition. These excessive and repetitive impact forces can cause tissue damage (Ekstrand et al., 1989), resulting from poor adaptation due to the players' lack of perception of surface resilience (Fujitaka et al., 2017; Rago et al., 2019). Therefore, further research should also investigate the relationship between perceptions of switches between playing/training surfaces and overuse injuries.

In our study, the peak VGRF of two movements, cutting and penalty kick, were investigated with insole pressure sensors (Moticon, OpenGo). The peak VGRF of the cutting movement (2.28 BW) differed significantly from the penalty kick (1.99 BW). As far as we know, this study is the first to compare the peak VGRF of both two movements by insole pressure sensor. According to the findings of Blackburn et al. (2003), the VGRF in cutting movements (in football) is equal to 2.5 BW (n = 8) whereas Bencke et al. (2000) found that the VGRF in cutting (in Handball) is 2.8BW (sample rate of 1000 Hz). In another study, Simonsen et al. (2000), determined the peak VGRF with the sample rate of 500Hz is 2.78BW (n = 6). In kicking movements, Orloff et al. (2008) and Katis et al. (2010) determined that VGRF can reach up to 2.25 BW. A comparison between cutting and penalty kick movements in these studies revealed that the VGRF of the cutting movement was greater than the kicking movement. Accordingly, our study shows that the VGRF of the two movements differ significantly. The diverse values of VGRF in our study and these previous studies (especially in cutting) may be due to several intrinsic and extrinsic factors, such as different techniques used to perform the movements, type of surfaces and boots, the number of test subjects, and finally the use of different measurement tools, especially with a different sample rate. Besides the advantages of using insole sensor measurements, such as high reliability, the limitation of insole pressure sensors is that they provide a lower force and latency in force kinetics compared to the force plate during fast motions (Stöggl et al., 2017). This study suggests that boot developers should pay more attention to the midsole of boots when designing outsole configurations. However, the optimum design is likely to be when all three variables (midsole hardness, movements, and surface hardness) are taken into account by weighting them according to the effect size – as shown in our study – to further improve the design of studded football boots.

3.1.1.5 CONCLUSION

This study aimed to investigate the effect of three independent variables: midsole cushioning, surface hardness, and different movement types using a pressure measurement system. The findings determined that the ATS (with a softer midsole) generates greater VGRF (2.25 BW) than the NGS (1.95 BW). However, switching between surfaces from natural grass to artificial turf (greater hardness) was observed to have a significant increase in VGRF (4%). Finally, the peak VGRF of the cutting movement (2.28 BW) differed significantly from the penalty kick (1.99 BW). This study provides useful data for boot developers to weigh these three independent variables to further improve the design of studded football boots.

3.1.2 SECTION II

Varied Cushioning Insole and Perception

This article was originally accepted in October 2023 with the title "Perception of running shoe cushioning: Determining the correspondence between subjective assessment and in-vitro measurement" in the International Journal of Exercise Science.

3.1.2.1 INTRODUCTION

When running, the human body is exposed to repetitive impacts resulting from sudden decelerations (jolt) at initial ground contact. These accumulated impacts are considered a risk factor for the development of overuse running injuries (Clarke et al., 1983; James et al., 1978; Milner et al., 2006). To prevent excessive loads and related injuries, three concepts were proposed by Nigg and Segesser (1992), namely, cushioning, support and guidance

The cushioning system of shoes can be highly related to certain shoe components e.g. insoles. In the biomechanical measurement using an impact testing device, Chiu et al. (2007) determined that insoles can absorb impact energy up to 24–32 % (Chiu et al., 2007), and in the study of O'Leary et al. (2008), such a reduction of impact peak force using insoles reaches to 6.8% (O'Leary et al., 2008) This would indicate that insoles play an important role in cushioning properties of sport shoes under a low impact energy conditions. An increase in shoe cushioning does not always result in such an impact reduction. Kulmala et al. (2018) determined that increasing cushioning increased the slope of the loading rate, i.e., the speed at which forces impact the body. Kulmala et al. (2018) and Milani et al. (1997) stated that such excessive cushioning of a running shoe may be responsible for injuries by creating a perceptual underestimation of the actual impact severity. An increase or reduction of footwear cushioning may affect the natural frequency of soft tissue vibration, where muscle tuning then occurs (Boyer et al., 2006). The paradigm of muscle tunning determined that runners adapt their muscle responses with each step to minimize the transmission of vibrations through the soft tissues in their lower extremities. Such an adaptation (muscle tuning) raises the question of the extent to which a participant can perceive cushioning changes.

The science of psychophysics introduced a threshold as the minimum intensity a subject can perceive, known as the Just Noticeable Difference (JND) (Pisciotta et al., 2018; Isherwood et al., 2018). Detecting such a threshold can aid in better understanding how muscle adaptation originates when a stimulus occurs (Robbins et al., 1989; Robbins et al., 1990; Robbins et al., 1991). Other researchers, Hennig (2011) and Salzano et al. (2021) attempt to compare the results of subjective assessment with in-vitro physical properties of running shoes using the impact testing method (Hennig et al., 1996; Salzano et al., 2021). Their findings determined a poor correspondence between the cushioning measured in vitro using peak acceleration (g-Max) and wear tester ratings (subjective assessment). The method used in subjective assessments of these studies is the Likert-type scale (1–7). The complexity of this measure can worsen the reliability of assessments. In other words, the number of individuals who reliably assessed the footwear could be increased by reducing the complexity of the measure to simple binary Yes/No questions (Hoerzer et al., 2016; Mills et al., 2010).

Among the methods available for eliciting comparative judgments, binary choice is the easiest for respondents because it compares only two stimuli at a time (Hoerzer et al., 2016). In addition, the method of paired comparisons uses the inherent familiarity with and ability to make comparisons. Furthermore, the effect of *different insole materials* on cushioning attenuation has been investigated in only a few studies (Nigg et al., 1988; Che et al., 1994; Ruano et al., 2009). Insoles with different types of material can react to local pressure peaks in different ways. *One way* is the absorption of impact energy by compression of the material. Closed cell foams and insoles with integrated air chambers produce a counter-force when compressed (the inner pressure of the cells rises) and can thus return part of the stored energy during unloading. Open-cell foams have a lower energy return capability. The *second way* of reacting to local pressure peaks is the evasion of the material, occurring in materials with some characteristics of liquids, for example, silicones and other viscoelastic. The material moves from an area with high pressure to surrounding areas with lower pressure and thus balances the pressure. Since the mechanism of impact attenuation in both types of material is different (affecting kinetic and kinematic changes in running), it is necessary to investigate the subjective parameters of cushioning at an early stage.

A parameter that has been investigated in past studies (Pisciotta et al., 2018; Hennig et al., 1996) is the perception of cushioning measured by rating 'hardness' or 'softness' of the material underfoot in a subjective assessment. However, in these studies, perception of cushioning defined how hard or how soft the material of the midsole is but this parameter cannot indicate in detail the c*omfortability of the attenuation mechanism (= cushioning comfort).* Past studies have not thoroughly investigated whether *cushioning comfort* in subjective assessment can be linked to the parameter of in vitro physical properties of running shoes using the impact testing method.

In addition to subjective assessments, mechanical testing can provide parameters, such as peak acceleration, for predicting the cushioning properties of insole materials (Hennig et al., 1996; Salzano et al., 2021). While peak impact acceleration offers insights into initial shock absorption, it may not offer a comprehensive evaluation of overall cushioning, as cushioning may encompass other factors like the rate of acceleration change (loading rate) which have been shown in the biomechanical measurement (Aguinaldo et al., 2003). The loading rate in the context of running gait refers to the rate at which the body experiences an increase in vertical ground reaction force during the initial contact and loading phase of running. However, such a parameter has been overlooked in the mechanical testing. For a more precise prediction of cushioning in running insoles, a holistic approach involving multiple factors and testing methods, including mechanical testing and subjective comfort assessments, should be taken into account. Consequently, this study aims to achieve two objectives: firstly, to determine the just noticeable difference in insole cushioning in running through subjective assessment, and secondly, to explore potential variables in mechanical impact testing to assess their potential as predictors for cushioning comfort.

3.1.2.2 METHODS

Participants. Nineteen (n = 19) male participants were recruited from the sports center at the Technical University of Munich with a mean age of 23.89 (SD = 2.31), weight of 73.52 kg (SD =3.08), and height of 178.84 cm (SD = 2.81). Participants gave written, informed consent before the experiment. The consent form declares confidentiality of the objectives, risks of the study, and protection of personal data through appropriate procedures for anonymization according to the EU General Data Protection, Regulation, EU (2016). In addition, the consent form assures participants they are free to withdraw from the research at any time without giving a reason and without penalty for not taking part. This research was conducted according to the ethical standards of the World Medical Association Declaration of Helsinki (2013) and the IJES Ethics Statement (Navalta et al., 2020).

Protocol. Three commercially available insole materials—(i) ethylene-propylene-diene-monomer rubber foam (EPDM), (ii) Ipocon-Gel, and (iii) synthetic viscoelastic urethane polymer Sorbothane©, and their combination were prepared in the pilot study. In total, 12 samples were tested using an impactor device. In the end, only four insole samples, one with the highest, one with the lowest, and the two materials with middle range magnitudes of the peak accelerations were selected to use in the subjective measurement. These were as follows: EPDM 6 mm = (E), Ipocon Gel = (IP.Gl), a synthetic viscoelastic (viscoelastic urethane polymer Sorbothane©) with a thickness of 3.2 mm (= SB_{tn}), and a similar synthetic viscoelastic but with a thickness of 4.8 mm (= SB_{tk}). The mean (and standard deviation) of impact peak of EPDM, IP.Gl, SB_{tn}, and SB_{tn} were 32.41 g-Max (± 0.47), 11.54 g-Max (± 0.25), 12.6 g-Max (± 1.13), and 7.2 g-Max (± 0.2), respectively (please see Table and Figure 3.2.1). The drop height was 6 mm and the weight of the impactor was 4.3 kg.

Table 3.2.1 Samples were tested in the pilot test with an impactor device (drop height of 6mm and weight of 4.3 kg).

ID#	Material	TT	**ID#**	Material	TT
1	E1	6	*7*	E+ IP.Gl (2mm+3mm)	5
2	E2	9	*8*	E+ IP.Gl (3mm+4mm)	7
3	IP.Gl -1	3	*9*	E+ IP.Gl +E (2mm+3mm+2mm)	7
4	IP.Gl -2	4	*10*	E+ IP.Gl +E (2mm+4mm+2mm)	8
5	IP.Gl -3	6	*11*	SB_{tn}	3.2
6	IP.Gl-4	8	*12*	SB_{tk}	4.8

EPDM = E, Ipocon Gel = IP.Gl, Sorbothane© thin= SB_{tn}, Sorbothane© thick= SB_{tk}

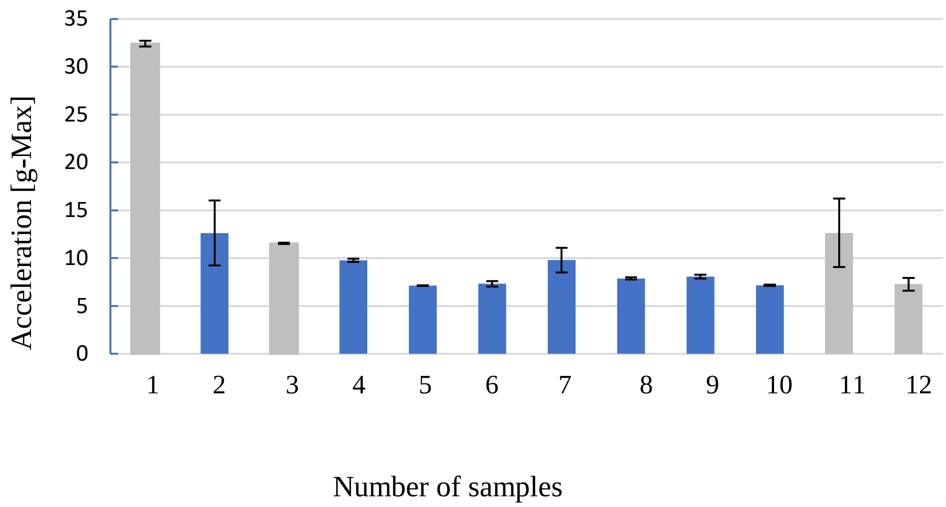
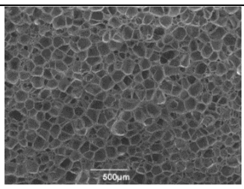
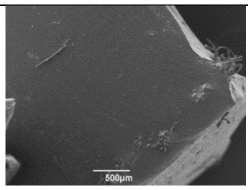
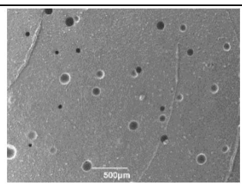
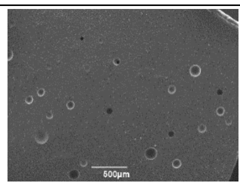


Figure 3.2.1 Peak acceleration, g-Max, of twelve samples: four samples are highlighted in gray used in the main experiment (Error Indicator: 95%-Confidence Interval).

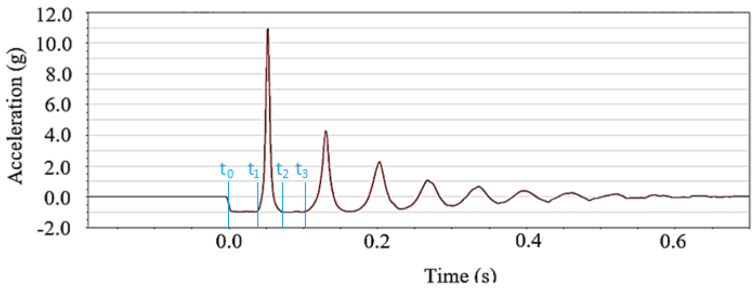
Cross sections of these four samples were prepared for scanning electron microscope through JEOL - Model JSM-6390. Before the start of the study, a type A shore durometer (Kern & Sohn GmbH, Germany) was used to determine the material hardness. The samples are shown in Figure 3.1.2. Four samples, EPDM, IP.Gl, SB_{tn}, and SB_{tk} were designed and cut manually to replace the original insoles of shoe size 42 - 47 (Scott running shoe, model: Palani).

	EPDM[1]	IP.Gl[2]	SB_{tn} [3]	SB_{tk} [4]
SEM *				
SH **	5.6	23. 2	13. 2	17. 8
Thickness	6mm	3mm	3.17mm	4.7mm

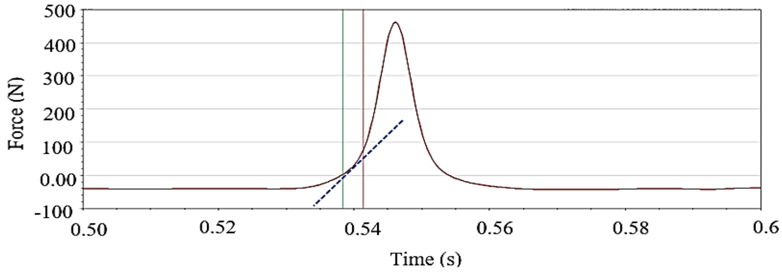
*SEM:Scanning Electon Microscope Machine **SH =Shore hardness [1]EPDM= ethylene-propylene-diene-monomer rubber foam; [2]IP-Gl: Ipocon-Gel; [3] SB_{tn} = thin synthetic viscoelastic urethane polymer Sorbothane©; [4] SB_{tk} = thick synthetic viscoelastic urethane polymer Sorbothane©

Figure 3.2.2 SEM frames and mechanical characteristics of four samples

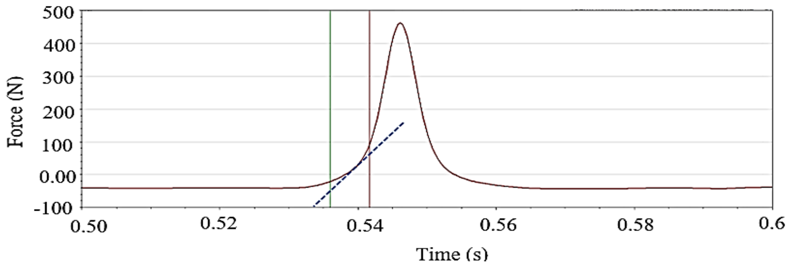
A dynamic shock absorption test was used for the measurement of cushioning properties. The pneumatic impactor device consists of a 5 cm diameter indenter and 4.3 kg weight. The machine operates by allowing the weight to fall onto the tested material from a defined height of 7 mm (potential energy of 0.3 J). A single-axis accelerometer sensor (range of −50 to 50 g-Max) was attached to the impactor to quantify the shock absorption with a measuring frequency of 5000 Hz. The impact tester provides the acceleration-time graph shown in Figure 3.2.3 (using Gaussian smoothing, Sigma = 5 which corresponds to 1 millisecond at the measuring frequency of 5000 samples per second). Two parameters, g-max, and the Jolt $_{\alpha, \beta}$ were evaluated: g-max is the peak of acceleration of the first impact (Figure 3.2.3). The Jolt was calculated as the slope between 5-20% in the inertial force-curve, and the Jolt $_{\beta}$ is the slope from 0-80 Newton in the inertial force-curve below.



a.



b.



c.

Figure 3.2.3 Overview of peak acceleration, Jolt $_{\alpha}$ and Jolt $_{\beta}$. **a** schematic of peak acceleration with the Impactor test device (at t_0 the device is triggered, from t_0 to t_1 and from t_2 to t_3 the impactor is in free-fall). An acceleration value is reported in gravitational units (1 g = 9.80665 m s $^{-2}$). **b** The Jolt $_{\alpha}$ was calculated as the slope between 5-20% of the inertial force graph. **c** The Jolt $_{\beta}$ was calculated as the slope between 0-88 Newton of the inertial force graph.

Subjective assessment. Four samples were compared with pairwise methods. This method on 'n' abilities requires several n (n − 1) / 2 different combinations, which must be tested by each tester (Böhm et al., 2008). With paired comparisons, respondents select the stimulus, or item, in each pair that had the greater magnitude on the choice dimension they were instructed to use.

Pair comparison test (PCT) is a binary test that requires a simple decision between two alternatives. In our study, participants performed six pairwise comparison tests with shoes which were equipped with one of the four abovementioned insoles in a random order, i.e., (EPDM vs. IP.Gl); (EPDM vs. SB_{tn}); (EPDM vs. SB_{tk}); (IP.Gl vs. SB_{tn}); (IP.Gl vs. SB_{tk}); (SB_{tn}vs. SB_{tk}). At the beginning of the subjective test, the plantar sensitivity of participants was evaluated with 3.61 monofilament = grade 4 (Semmes Weinstein Monofilament Examination), which is equivalent to 0.4 g of linear pressure (Jeng et al., 2000). Nine defined points (distal great toe, third toe, and fifth toe; first, third, and fifth metatarsal heads; medial foot, lateral foot, and heel) (Costa et al., 2022) on the participant's foot plantar (random foot) were tested three times to detect peripheral neuropathy.

*Main experiment***.** In the first, participants ran five minutes on the treadmill at their desired velocity with an experimented shoe (Scott- model: Palani) with the original insole. They then performed six movements: quad piriform walk, hip opener, arm circles, leg crossover, and inchworm (following the experimenter's instructions). Participants ran on the treadmill for 3 minutes with a randomly selected insole, and immediately afterward, changed into the shoe with a different insole, running for a further 3 minutes (in pairwise comparison tests). After accomplishing the first pair test, the following two questions were put to the participants:

1. Is there any difference between the cushioning features of the two running shoes? Possible answers: Yes/No.
2. Which shoe did you perceive as "…having a better cushioning comfort"? Possible answers: " The first shoe "/ " The second shoe ".

After a one-minute rest, participants repeated the same procedure until all pair tests had been accomplished. Participants were informed that the meaning of cushioning is protection against force or shock. Accordingly, cushioning comfort was defined as 'How comfortable the procedure of reducing shock while running'.

In our study, a number of factors were controlled to determine biases in the experiment. Subjects were blind to any information about the selected shoes (type of the shoe and type of the insoles). The experimenter untied and removed the shoes from the subject's feet. Participants were given the option of changing the shoe in the early phase of the experiment (warm-up) when they were not convinced of the comfort of the fit.

Based on our previous study (Keshvari et al., 2022), participants mostly chose a running pace between 10 – 12 (2.7 – 3.3) km/hour (m/s). Hence, in our study, the running pace was firstly set at 10 km/h (2.7 m/s), however in case participants could change it according to their desired velocity. To avoid any biases, all participants wore identical socks during the experiment.

3.1.2.3 RESULTS

The results of the monofilament test determined that the minimum plantar sensitivity of participants (lower range) was 75% and, the maximum (upper range) was 92%. The average plantar sensitivity was 80.3 % (SD = 6.8 %). These results are based on 513 tests among 19 participants, nine plantar spots, and each of three trials.

In-vitro measurement – Impactor. Peak acceleration, Jolt $_{\alpha}$, and Jolt $_{ß}$ were measured with the impactor and analyzed with self-designed software in Visual Basic (Table 3.2.2). The differences in peak acceleration between samples were as follows: (IP.Gl vs. SB_{tn}) = 0.7 g-Max, (SB_{tk} vs. IP.Gl) = 3.2 g-Max, (SB_{tk} vs. SB_{tn}) = 3.9 g-Max, (IP.Gl vs. EPDM) = 5.4 g-Max, (SB_{tn} vs. EPDM) = 5.7 g-Max and (EPDM vs. SB_{tk}) = 9.6 g-Max. The impactor results determined that EPDM had the lowest Jolt $_{\alpha}$ equal to the lowest inclination and then STK, IP.Gl, and SB_{tn}, respectively. In Jolt $_{ß}$, EPDM, SB_{tk}, IP.Gl and SB_{tn} also showed the lowest to highest inclination (see Table 3.2.2).

Table 3.2.2 Peak acceleration and Jolt $_{\alpha}$ and Jolt $_{ß}$ in 7mm drop height

	Jolt $_{\alpha}$ (N/S)	Jolt $_{ß}$ (N/S)	g-Max
	5%-20%	0N-88N	Peak Acceleration
EPDM[1]	23961	34011	17.9
sd	*2224*	*2781*	*1.4*
SB_{tk} [2]	34562	52030	8.3
sd	*2867*	*2756*	*0.3*
IP.Gl[3]	39297	54897	11.5
sd	*1831*	*2166*	*0.4*
SB_{tn} [4]	50535	69162	12.2
sd	*3469*	*4485*	*0.6*

[1]EPDM= ethylene-propylene-diene-monomer rubber foam; [2]IP-Gl: ipocon-Gel; [3]St_n = thin synthetic viscoelastic urethane polymer Sorbothane©; [4]St_k = thick synthetic viscoelastic urethane polymer Sorbothane©

The inertial forces of insoles varied from 349 N, 485 N, 514 N, and 754 N respectively for EPDM, SB_{tk}, and IP.Gl and SB_{tn}. These were calculated using the peak acceleration of each insole multiplied by the weight of the impactor. All participants chose a velocity of 10 km/h and maintained such a velocity in their pair-comparison tests. In all 114 pair tests (n =19 × 6 = pair test), participants were able to distinguish the cushioning difference. The just noticeable difference between the two samples is indicated when participants compared IP-Gl and Stn. The just noticeable difference between the two samples' g-Max was 0.7. The four insoles were evaluated using a 4 × 4 square decision matrix for each tester. A decision matrix for all testers is shown in Table 3.2.3. It was normalized on the number of subjects in Table 3.2.4. The transformation of normalized rank data in Table 3.2.4 to ratio scaled preference z values was calculated according to the assumption of a standard normal distribution justified by the law of comparative judgment (Hull, 1992) (see Table 3.2.5). A pair comparison test of samples in scaled values is shown in Figure 3.2.4.

Table 3.2.3 The specific PCT of all testers. *Notice: Read vertically in the columns the number of rated preferences. Example: 19 subjects rated insole G as "…having a better cushioning comfort" than insole SB_{tk}, whereas 6 subjects rated insole S2 as being more comfortable than insole IP.GL*

	SB_{tk}	IP.Gl	EPDM	SB_{tn}
SB_{tk}		13	8	15
IP.Gl	6		2	10
EPDM	11	17		15
SB_{tn}	4	9	4	

Table 3.2.4 Normalized PCT matrix of all testers by dividing PCT from table 3.2.3 to 19

	SB_{tk}	IP.Gl	EPDM	SB_{tn}
SB_{tk}		0.68	0.42	0.79
IP.Gl	0.32		0.11	0.53
EPDM	0.58	0.89		0.79
SB_{tn}	0.21	0.47	0.21	

Table 3.2.5 Replacement of normalized PCT by corresponding Z-values

	SB_{tk}	IP.Gl	EPDM	SB_{tn}	*AVG*
SB_{tk}		0.47	-0.2	0.80	0.27
IP.Gl	-0.47		-1.25	0.06	-0.41
EPDM	0.19	1.2		0.80	0.56
SB_{tn}	-0.80	-0.07	-0.80		0.41

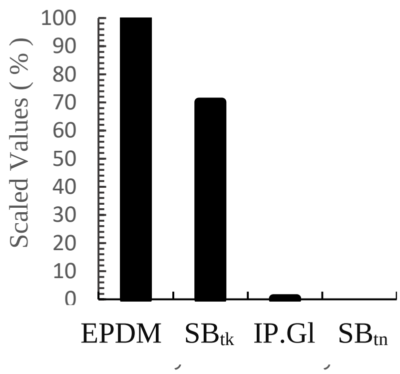


Figure 3.2.4 Pair-comparison test of samples in scaled values

To find a statistical difference between the four insole samples, coin theory was applied in the pairwise comparison test. Coin theory is a binomial distribution based on two possible outcomes heads and tails. In the case of coins, heads and tails each have the same probability of 1/2. In Python, we performed a binomial test using the binom_test() function from the scipy. stats library, which uses the following syntax: binom_test (x, n = None, p = 0.5, alternative='two-sided'), and effect size in binominal data is measured as follows: 1/p0 -pt. Our findings determined a significant difference (P value) in the following pair tests: (SB_{tk} vs. IP.Gl): 0.013, (IP.Gl vs. EPDM): 0.0007, (EPDM vs. SB_{tn}): 0.019 and (SB_{tk} vs. SB_{tn}): 0.019. However, the cushioning comfort of the two pair tests did not differ significantly: (SB_{tk} vs. EPDM): 0. 64, (IP. Gl vs. SB_{tn}): 1.0. To calculate the effect size, univariate ANOVA model used in IBM® SPSS® Statistics 26. The independent variable (insoles) showed a medium effect size, 0.43 on the dependent variable (preferred cushioning comfort).

The results from the second question were also analyzed by correlation coefficient. Spearman's rank correlation was calculated in Python using *the spearmanr() SciPy* function. There was a negative and significant correlation between Jolt $_{\alpha}$ and, perceived cushioning and, similarly, between Jolt $_{ß}$ and perceived cushioning r (10) = - 0.93, p = 0.00001. Our findings revealed that an increase of Jolt $_{\alpha}$ and Jolt $_{ß}$ resulted in a reduction in the perception of comfort. No correlation was found between peak acceleration and perception of cushioning (p = 0.1).

3.1.2.4 DISCUSSION

In this study, we evaluated the just noticeable difference in cushioning perception and also compared cushioning comfort with an in-vitro measure of impact attenuation parameters. In our series of 114 paired comparison tests involving 19 participants, it was observed that the participants were capable of detecting differences in insole cushioning stimuli. Notably, the just noticeable difference (JND) was determined between the SB_{tn} and IP.Gl insoles, amounting to 0.7 g-Max. In simpler terms, participants exhibited the ability to discern a minimal increase of 6% in cushioning properties, distinguishing between the 11.5 g-Max (SB_{tn}) and 12.2 g-Max (IP.Gl) conditions. This outcome is largely consistent with the results of Preciotta et al. (2018) and Isherwood et al. (2021), who identified JND values of 1.4 g-Max and 1.1 g-Max, respectively, in the context of distinct sole cushioning levels. The heightened precision of the JND observed in the present study, in contrast to the aforementioned investigations, is likely attributable to variances in the subjective assessment methods and experimental designs employed. The earlier studies (Preciotta et al., 2018; Isherwood et al., 2021) utilized a visual analogue scale (VAS) for subjective assessment, which lacks comparative reference values for intensity. The cognitive task of estimating attribute intensities within this framework may be more intricate compared to the straightforward comparison of one insole to another, where reference values are not available. Furthermore, differences in methodology, such as the duration of insole wear and its impact on somatosensory perception (Saxton et al., 2020), may yield varying results. In our study, participants engaged in uninterrupted running for two minutes, whereas participants in the earlier studies behaved differently, possibly contributing to the differential effects of wear time on somatosensory perception. Another noteworthy methodological variance concerns the potential energy expended in mechanical testing. In the studies mentioned above, the potential energy utilized was 8J, while in our study, it was restricted to 0.3 J, a choice made to serve the specific objective of assessing insole behavior. During a pilot study, we initially employed potential energy levels of up to 8 J to test the mechanical properties of all four insoles. However, preliminary findings indicated that such high potential energy levels could damage the insoles. Consequently, we defined the potential energy with a 7 mm drop height and a 4.3 kg weight, equivalent to 0.3 J. This limitation of potential energy resulted in peak impact accelerations ranging from 8.3 g-Max to 17.9 g-Max (refer to Table 3.2.2). Notably, the peak force was defined based solely on interactions with the insoles, in contrast to other studies, which also measured interactions with midsoles or a combination of midsoles and insoles. As per the data presented in Table 3.2.6, the current impact force observed in our study falls within a range that aligns with the findings of other studies. This supports the feasibility of conducting insole testing with the chosen potential energy level.

Table 3.2.6 Comparisons of Impact Testing Methods Used in Previous Studies and the Present Study.

	Material	Impact Mass (Kg)	Impact Energy (J)	Peak Acceleration (g)	m × a (N)
Henning et al. (1991)	m+i*	7.8	3.2	9.1	696
Henning et al. (1993)	m+i	8.5	4.2	11	917
McNair et al. (1994)	m+i	9	4.4	9.6	847
Milani et al.(1997)	M	7.3	3.6	9.6	687
Preciotta et al. (2018)	m+i	8.5	5	11.1	924
Chiu et al. (2007)	M+i	6.2	1.8 - 6	9.6 - 20	584-1229
Current study	i	3.4	0.3	8.3	349-754

m + i* = midsole +insole; i*= insole

These two insoles, which exhibited minimal differences in peak acceleration, belong to the same family of viscoelastic materials and demonstrate similar mechanical behavior (as opposed to EPDM). This discovery holds significance, as it suggests that participants can perceive such differences in peak acceleration (g-Max) not solely based on varying material characteristics, but rather due to their precise just noticeable difference (JND) discernment, even within materials from the same group

The secondary objective of this research was to enhance our comprehension of the connection between the mechanical characteristics and the cushioning comfort provided by various insoles. Our results revealed that participants rated the EPDM insole, which exhibited the lowest Jolt, as the most comfortable in terms of cushioning. Notably, a statistically significant correlation (r = 0.9) was observed between both $Jolt_{\alpha}$ and $Jolt_{ß}$ and the perceived cushioning comfort. Vibrations within the soft tissue compartments of the leg, including muscles, fascia, surrounding tissue, and skin, are initiated during the rapid deceleration of the leg upon landing (Nigg et al., 2001; Wakeling et al., 2002). These vibrations are subject to influence by three key factors: running speed, the hardness or flexibility of the running surface, and the vibration-damping properties of the footwear. For this study, the first two factors remained constant among all participants, with the only variation being the type of insole used. Drawing from the muscle tuning paradigm, it is hypothesized that as the input signal represented by $Jolt_{\alpha}$ (ranging from EPDM = 23 kN/s to SB_{tn} = 50 kN/s) or $Jolt_{ß}$ (ranging from EPDM = 34 kN/s to SB_{tn} = 69 kN/s) increases, it moves closer to the resonant frequency of the soft tissue package. In response, the muscles need to adapt and tune to the surface's hardness. This tuning may be most pronounced when using the EPDM insole, followed by the SB_{tk}, and IP.Gl, and SB_{tn} in descending order. Alternatively, when confronted with harder or softer surfaces, muscles may adapt by increasing damping to minimize resonance. Thus, both $Jolt_{\alpha}$ and $Jolt_{ß}$ serve as input signals, leading to an increase in the damping coefficient of the soft tissues and, subsequently, more intense muscle activity. These dual mechanisms work in harmony to reduce vibrations, potentially resulting in a more robust muscle response when utilizing insoles, ranked in descending order as SB_{tn}, IP.GL, SB_{tk}, and EPDM. This, in turn, may have implications for cushioning comfort and motor task performance.

Furthermore, it is plausible that participants employ these mechanisms as a protective response to mitigate the risk of injuries stemming from elevated soft tissue vibrations. Future studies should delve deeper into these assumptions to explore how runners adapt to such changes in a controlled laboratory setting.

3.1.2.5 CONCLUSION

In this study, the perception of cushioning and cushioning comfort of four insoles were compared in a pair comparison test among runners. The cushioning attenuation properties of these insoles were compared in vitro measurements using an impactor. Our findings show that runners could detect a minimum difference of peak acceleration of 0.7 g-Max. In addition, Jolt was shown to be a predictor of cushioning comfort. Any reduction in jolt results in a significant increase in cushioning comfort. In summary, these findings demonstrate that, under controlled conditions, runners were able to distinguish a 6% increase in peak cushioning in insoles while running. Perceived cushioning comfort was therefore consistent with mechanical impact test results. The findings and the methodological framework of this study can be used to enhance the development and design of shoes and prosthetics.

3.1.3 Section III

Varied Cushioning Insole, Kinetic and Spatial-Temporal Parameters

3.1.3.1 INTRODUCTION

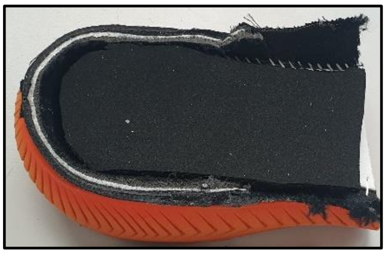
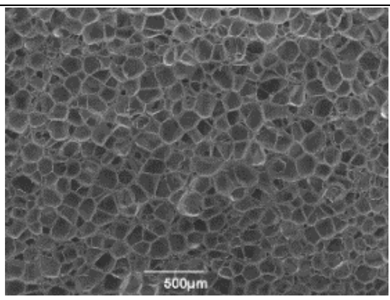
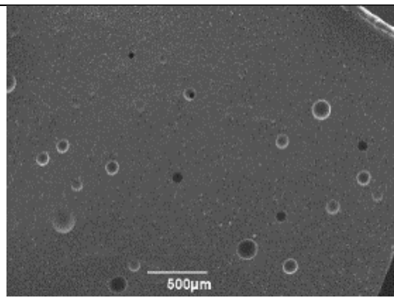
Comfort holds a pivotal role in the realm of footwear production. Nevertheless, defining and quantifying comfort presents intricate challenges. As asserted by Mills et al. (2010), comfort with footwear can be characterized as an ever-evolving perception influenced by various factors, including mechanical, neurophysiological, and psychological elements, as noted by Chen et al. (1995), Miller et al. (2000), Mündermann et al. (2001), and Nigg et al. (1999). What individuals deem 'comfortable' can vary significantly among different groups, driven by personal preferences and past experiences. Research options for measuring comfort are notably limited. Hoerzer et al. (2016) advocate for greater reliability in comfort assessment through simplification, such as employing a binary questionnaire instead of a visual analog scale (VAS). Specifically, the distribution of pressure within the shoe assumes paramount importance in orthopedic and biomechanical considerations. Individual-specific traits, encompassing aspects like anthropometry, skeletal alignment, foot sensitivity, and plantar pressure patterns, have been linked to individuals' perceptions of comfort in footwear, as highlighted by Chen et al. (1995), Miller et al. (2000), and Mündermann et al. (2001). It is proposed that a comfortable shoe should reduce pressure between the foot and the insole, averting localized irritations and maintaining pressure beneath a certain threshold on the plantar surface of the foot. Delving into the relationship between perceived comfort and the quantifiable parameter of pressure could enhance our comprehension of the science behind human-surface interaction during running, especially in the presence of perturbations. Krouskop et al. (1987) employed pressure measurements within an above-knee socket to gain insights into the comfort aspects of such sockets. In a separate study, Chen et al. (1995) found that changes in shoe comfort were detectable through pressure distribution between the plantar surface of the foot and the shoe. They observed that, during running, the pressure in the medial forefoot area significantly differed between the most and least comfortable insoles. In another investigation, Zhang et al. (2019) revealed significant differences in plantar pressure and perception characteristics between two shoes—one with harder and another with softer midsoles—although these findings were not consistently aligned. Past research underscores the challenge of reaching a consensus regarding the relationship between cushioning comfort and plantar pressure, necessitating further exploration.

The primary objective of the present study is to explore the impact of cushioning comfort on pressure characteristics during running.

3.1.3.2 METHODS

Participants. A number of 19 male runners were recruited for our experiment from sports clubs and associations in Munich. They had a mean (standard deviation) age, height, and weight of 24.5 years old (± 3.5), 184 cm (± 5.1), 78.7 kg ± (6.0). Runners were required to be injury-free six months before the time of testing. All participants gave written, informed consent before the experiment. The consent form assures the confidentiality of participant data, presents the risks of the study, and protection of personal data through appropriate procedures for anonymization according to the EU General Data Protection, Regulation EU, 2016. In addition, the consent form assured participants they were free to withdraw from the research at any time without giving a reason and without penalty for not taking part.

Material of study. All participants wore Scott Palani running shoes (EU 42, 43, 44, and 46 Scott Sports SA, Givisiez, Switzerland). For each of the sizes, there were two identical shoes. Two experimental insoles were used in the experiment: synthetic viscoelastic urethane polymer Sorbothane© (SB_{tk}) and ethylene-propylene-diene-monomer rubber foam (EPDM). These are similar types of insole used in the previous section (please see Figure 3.3.1 and Table 3.3.1). All participants wore identical running socks.

	EPDM [1]	SB_{tk} [2]
Schematic of insole and midsole		
SEM *		
SH**	5.6	17. 8
Thickness	6 mm	4.7 mm

*SEM:Scanning Electon Microscope Machine **SH =Shore hardness [1]EPDM= ethylene-propylene-diene-monomer rubber foam; [4]SB tk= thick synthetic viscoelastic urethane polymer Sorbothane©

Figure 3.3.1 Overview of two insoles used in the study.

Table 3.3.1 Data of peak acceleration of insoles and midsoles in d08ifferent drop heights

	Drop height- 7mm*	Drop height- 20mm
EPDM	17.9 (1.44)	-
SB_{tk}	8.3 (0.32)	-
EPDM + midsole	4.9 (0.03)	8.5 (0.06)
SB_{tk} + midsole ***	5.1 (0.05)	8.6 (0.04)

* peak acceleration while impact energy is 0.3 j
**peak acceleration while impact energy is 0.8 j
*** midsole was in this study belongs to Scott Palani-running shoe

Protocol of experiment. Firstly, runners were asked to warm up for 5 minutes on the treadmill (H/P Cosmos Pulsar 4P; HP cosmos Sports & Medical, Gmbh, Nußdorf, Germany) with a velocity of 2.2 m/s. They ran with the running shoe equipped with insole-sensor pressure, Pedar insole (Novel, Gmbh) (sample rate of 100 Hz). This system is meant to evaluate the center of pressure trajectory, peak, and mean pressure in running stance phases.

To capture spatiotemporal parameters (contact time, step length, and step frequency), the Optogait system (Microgate, Bolzano, Italy) was used in our experiment. This system is an optical measuring instrument composed of a sending and a receiving measuring bar, each with 96 LED lights. Two parallel bars of the device system are integrated into the side edges of the motorized treadmill at 3 mm over the contact surface so that no extra fixation and calibration is needed. The device is connected to a computer which is controlled by the experimenter. Data is sampled at 1000 Hz so that parameters such as contact and flight time are measured with a precision of 1/1000 seconds in the position of the interrupted LEDs with a space resolution of 1.041 cm. Stride cycles were captured from toe to toe. Contact time is defined as the time from when the foot contacts the ground to when the toes lift off the ground. Step frequency in this context is the number of ground contact events per minute (García-Pinillos et al., 2018). More than one project has proved the validity and reliability of measurements of spatiotemporal parameters with this tool. (Healy et al., 2018; Lee et al., 2014; Lienhard et al., 2013).

After the warm-up, the main experiment was divided into two phases, and each phase was into two sections (Sections I and II). The velocity of the treadmill for all four sections was defined based on a prior survey of participants in the training course. An average velocity of 10.5 km/hour was chosen for the main experiment, i.e. both phases. In Section I of the first phase, participants ran with a random insole for 10 Minutes. In the 8^{th} minute of running, they completed a Borg Rating of Perceived Exertion (RPE) questionnaire with a scale of 6-20. After accomplishing this section, the runners answered a Likert-scale questionnaire on the comfort fit of the feet (questionnaire#1) (See figure 3.3.2). Section II (of the first phase) followed a similar procedure as Section I with a similar questionnaire but wearing a different insole.

Question 1. How 'uncomfortable' is the fit of the foot for each area shown in the following areas of feet?

- Very comfortable
- Comfortable
- Neutral
- Uncomfortable
- Very uncomfortable

- **Forefoot**
- **Lateral**
- **Heel**
- **Medial**
- **Foot Arch**

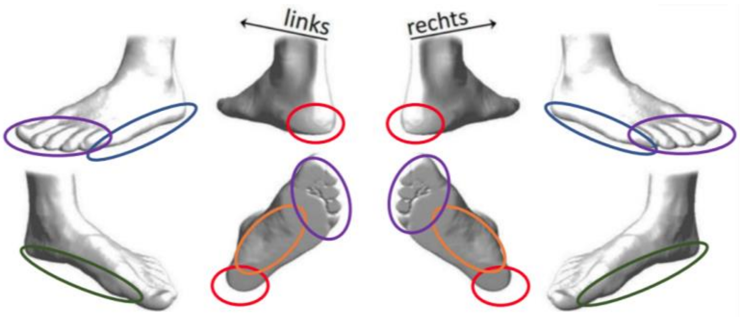


Figure 3.3.2 Foot classification into 6 sections (forefoot, lateral, heel, medial, foot arch, and total foot) to rate *comfort fit* using Likert scale (1-5)

Phase II followed a similar procedure as for Phase I with similar running duration and running velocity but a different questionnaire: Participants only answered two binary questions after Section I, which were as follows:

- ***Question 2.*** Which shoe has *the preferred cushioning comfort*?
  Possible answer: A or B

Cushioning was defined before the experiment as follows: 'the perception of reduced shock while running'.

- ***Question 3.*** *Which shoe is most comfortable in terms of sole softness?*
  Possible answer: A or B

In each section of the test, plantar pressure data was acquired via two consecutive 45s recordings at the following time intervals: 1´:30´´, 4´:30´´, and 8´:30´´. This corresponds to six 90s recordings per run. Peak pressure (PP) (kPa) and mean pressure (kPa) of all steps for both insole materials were calculated. Spatial-temporal (Swing time, Step length, and Step Frequency) for all 10 minutes was measured. For an overview of experiment steps, please see Figure 3.3.3.

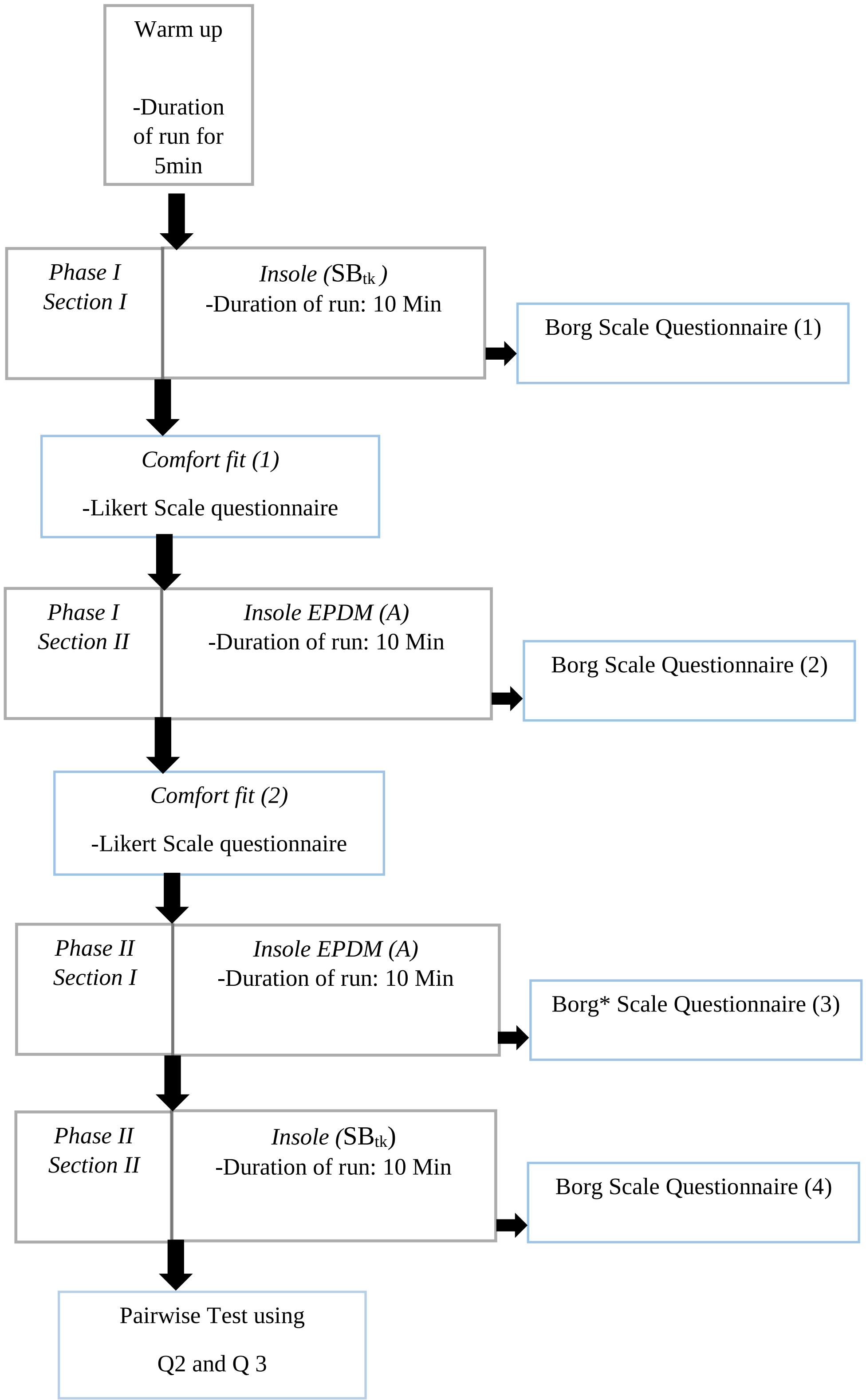
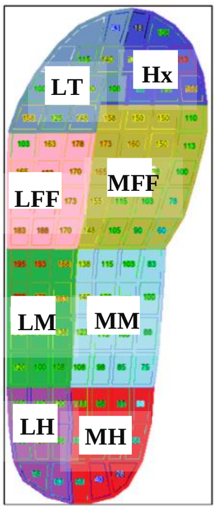
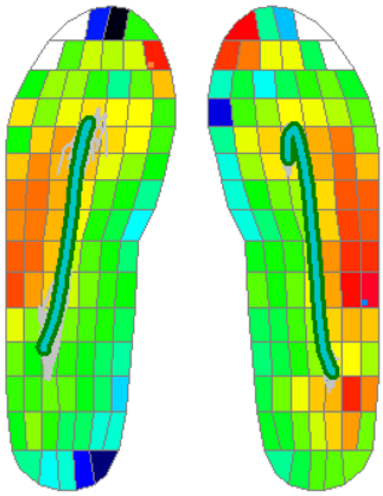


Figure 3.3.3 Schematic of the experiment procedure. Note: Borg scale questionnaires were asked in the 8th minute of all sections; Comfort fit question (#1) was asked right after Section 1 - Phase 1 and Section 2 - Phase 1; Pairwise test questions (Questions #2 and #3: Q2 and Q3) were asked right after phase 2-Section 2.

Methods of Analysis. The foot plantar was divided into 8 zones of classification as follows: Hallux (Hx), lateral toes (LT), medial forefoot (MFF), lateral forefoot (LFF), medial midfoot (MMF), lateral midfoot (LMF), and medial heel (MH) and lateral heel (LH). To define whether participants were forefoot or non-forefoot runners, the center of pressure (COP) in the early stance phase was evaluated during warm-up. If the COP was placed on the LMF and MMF or LH and MH, we refer to it as a non-forefoot runner. Moreover, the Pedar data during warm-up was used for defining the initial foot contact and thereby categorizing participants into forefoot and non-forefoot runners. The first step was to randomly select five steps for each participant with both insoles. The Pedar System can show the trajectory of the center of pressure on the plantar surface for each single step (Figure 3.3.4). It is possible to analyze which area of the plantar surface of the foot first touches the ground. If the initial foot contact area is LMF and MMF or LH and MH the runner is categorized as a forefoot (toe) runner or, conversely, a non-forefoot (heel) runner.



a.



b.

Figure 3.3.4.a An eight plantar classification Hallux (Hx), lateral toes (LT), medial forefoot (MFF), lateral forefoot (LFF), medial midfoot (MMF), lateral midfoot (LMF), and medial heel (MH) and lateral heel (LH)**.b** Center of pressure used in running in early steps.

Statistical Analysis Method. Two-way Repeated measures ANOVA (alpha level = 0.05) were used for the dependent variable (peak and mean pressure) and independent variable time and insole material. To analyze subjective data, in a solo pair test, coin theory was used. In addition, data of optogait has been analyzed both with parametric and non-parametric methods (Wilcoxon). C Some subjective data analyzed by coin theory. It is a binomial distribution based on two possible outcomes heads and tails. In the case of coins, heads and tails each have the same probability of 1/2.

3.1.3.3 RESULTS

Data of subjective test. The BORG scale is used to monitor the exertion level of the subjects while running. The subjects gave an average value of 12 (± 1.9) over all four laps, which is on the scale between ‘relatively easy (11)’ and ‘somewhat strenuous (13)’. The average effort in the first 10 minutes is assessed as relatively easy (BORG = 11.1 ± 1.8) and in the last 10 minutes is somewhat strenuous (BORG = 12.7 ± 2.4).

Question 1: Comfort fit. To assess the overall (dis)comfort of the two insoles examined the participants select a value from 1-5 on a Likert scale. (see Questions in the method section 3.1.3.3). Subsequently, a mean value of the (dis)comfort is calculated for all zones. Since each subject gave a comfort rating for both insoles, the samples are considered dependent samples. Additionally, the results of the Likert scale are ordinally scaled. Thus, both requirements for conducting a sign test are fulfilled. The overall (dis)comfort ratings between EPDM (Mdn = 2.17) and SB_{tk} (Mdn = 2.33) differ significantly from each other (p = 0.049, n = 19). A number of 13 (out of 19) participants perceive SB_{tk} as more uncomfortable, whereas a minority of four subjects rate EPDM as more uncomfortable. Two subjects evaluate both insoles as equally (dis)comfortable.

Question 2: Cushioning comfort. To analyze the results of the cushioning comfort, the probability for one of the two materials is calculated using the binomial coefficient. This can be used to determine if there is a significant difference between the choices for EPDM (B) (N=13) and SB_{tk} (N=6) in terms of subjective cushioning rating. The following hypotheses are as follows:

Null hypotheses H_0: P (B) = 0.5

Alternative hypotheses H_1: P (B) ≠ 0.5

Calculating P(B) using the binomial coefficient:

$$P\ (X = k) = \binom{n}{k} p^k\ (1-p)^{n-k},\ k \in\ \mathbb{N}$$

$$P\ (EPDM) = \binom{19}{13}\ 0.5^{13}\ (1-0.5)^{19-13} \approx 0.052$$

In the present case, the null hypothesis can be rejected (P (EPDM) ≠ 0.5). Accordingly, a significantly larger proportion of test persons rated EPDM as more damped than Sb.

Question 3: comfortable sole softness. A number of 15 participants who chose EPDM have greater comfortable sole softness in comparison to only 4 who voted SB_{tk}. To find out if there is a significant difference between the decision for EPDM (B) (number N=15) or Sb (A) (N=4) in terms of comfortable sole softness, the probability for one of the two materials calculated using the binomial coefficient. The following hypotheses are made for this:

Null hypotheses H_0: P (B) = 0.5

Alternative hypotheses H_1: P (B) ≠ 0.5

Calculating P (B) using the binomial coefficient:

P (X = k) = ( ) $p\ k$ (1 − p) n− , k ∈ ℕ P (B) = (19 *15) 0.5* 15 (1 − 0.5) (19−15) ≈ 0 .0074

Thus, the null hypothesis can be rejected (P (B) ≠ 0.5). This means that there is a significant difference between the number of decisions for EPDM (B) and SB_{tk} (A) in terms of comfort. EPDM (B) is thus perceived as more comfortable by the test persons.

*Biomechanical measurement***.** The mean and peak pressures of each plantar zone were measured for two insole materials using A paired sample t-test. As shown in Table 3.3.2, except the lateral heel, all other zones differed significantly in all zones except the lateral heel (alpha level = 0.05).

Table 3.3.2 Comparing pressure characteristics of two insoles (EPDM vs. SB_{tk}) using a pair sample T-Test, and related effect size and significance level.

Foot Plantar	Mean Pressure [kPa/kg]		Peak Pressure [kPa/kg]	
Zones	P value	Cohen's *f*	p	Cohen's *f*
M Heel	*> 0.001*	0.93	*> 0.001*	0.97
LHeel	0.152	0.35	0.167	0.34
MMF	*0.033*	0.54	0.02	0.6
LMF	*> 0.001*	1.18	> 0.001	1.27
MH	*>0.001*	1.45	> 0.001	1.45
LH	*>0.001*	1.12	0.001	0.9
HX	*0.009*	0.69	0.014	0.64
L toes	*0.002*	0.86	0.002	0.86

There was also a significant difference in mean pressure between EPDM (M = 299.1kPa SD = 68.7) and SB_{tk} (M =363.7, SD = 94.4), where the t (18) = -3.8, p = .001. In addition, there is a significant difference in peak pressure between EPDM (M = 352, SD = 79.8) and SB_{tk} (M =419, SD = 96.3); t (18) = -3.7, p = .001 (see Figure 3.3.5).

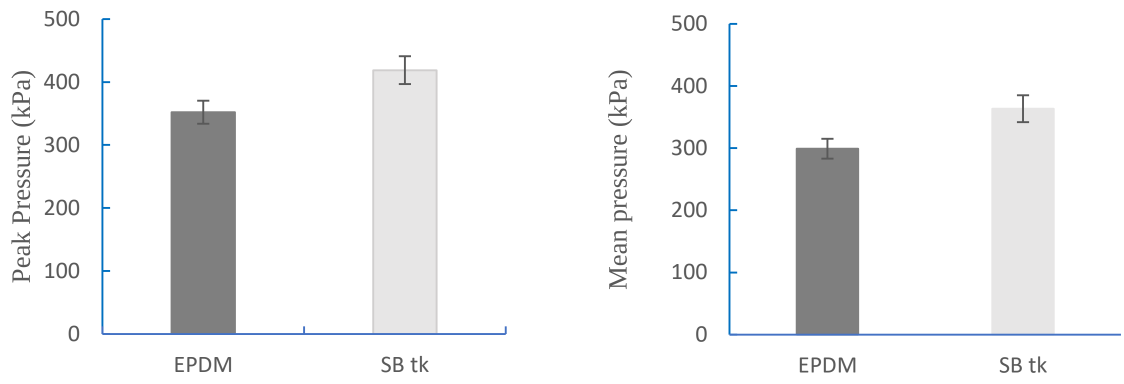


Figure 3.3.5 Peak and mean pressure of two insoles for all periods (SEM =standard error =+/-2).

Four spatiotemporal variables such as step length (cm), contact time (s), step frequency (step/min), and swing time (s) were measured for both insoles according to the protocol mentioned in Methodology (see Table 3.3.3). Except for variable contact time, other variables were not normally distributed. Using Paired Samples T-Test for the variable Contact time determined that there is no significant difference between two insoles (EPDM = 0.33, SD = 0.03 and SB_{tk} = 0.32, SD = 0.03): t (18) = 0.57, p = .58. Other parameters such as step length (cm), step frequency (step/min), swing time (s) analyzed by non-parametric test namely, Wilcoxon test, and determined that there is no difference between two insoles while running in our experiment.

Table 3.3.3 Comparison of two insoles with regards to different variables using the Wilcoxon non-parametric test

Parameters		Median	z-score	R**	Asymp. Sig.*
Swing time [ms]	EPDM vs. SB_{tk}	62 vs. 65	1.50	0.03	0.10
Swing time /contact time [%]	EPDM vs. SB_{tk}	20.2 vs. 20.5	1.50	0.03	0.12
Step length [cm]	EPDM vs. SB_{tk}	114.2 vs. 112.4	0.40	0.10	0.66
Step Frequency [Step/min]	EPDM vs SB_{tk}	153.6 vs. 155.8	0.10	0.03	0.87

**R = correlations coefficient ; *Asymp. Sig. = asymptotic significant- (2-tail)

3.1.3.4 DISCUSSION

The main finding of this study determined that the peak and mean pressure of the EPDM insole is significantly lower than SB $_{Tk}$ where the shore hardness of the insole was higher. Such findings are in line with Wiegerinck et al. (2009) who showed that more-softer shoes generate lower peak pressure in the toes (hallux and lesser toes), forefoot (medial, central, and lateral), and lateral midfoot regions than less-cushioned shoes in running. Our findings also agreed with those of Hong (2008) and Lam et al. (2016) who showed that a softer midsole caused lower peak pressure. In our study, an increase of insole shore hardness from 5.6 (for the sample, EPDM) to 17.8 (for the sample, SB_{tk}) was perceived by participants as a ‘less comfortable’ insole with higher peak and mean pressure. This is in line with Donato et al. (2015) who compared EVA midsole vs. Gel midsole. Using footwear equipped with a hard insole or midsole will tend to increase vibration frequency and tuning to active muscle is increased (Nigg et al. 2001). Therefore, there is an adaptation of runners’ techniques with harder footwear (midsole or insole) which avoids high impact rates (Hennig et al., (1996); Milani et al., (1997); Nigg et al., (1987). It can be speculated that this adaptation is related to kinetic changes in mean and peak pressure when shifting from soft insole EPDM to harder insole, SB_{tk}.In addition, since spatial-temporal parameters such as contact time, stride length, and stride frequency are maintained constant, it empowers the ‘preferred movement path paradigm’ defined by Nigg et al. (2017). Our findings also prove that when such a perturbation is imposed on the system, participant tends to stay on their running signature only by modifying their kinetic parameter.

3.1.3.5 CONCLUSION

In our perturbation study, two insoles were compared in running gait. Participants rated a running shoe as a higher cushioning comfort when it is equipped with insole EPDM. Such an insole defined as a soft insole (in mechanical testing) also resulted in lower peak and mean pressure in comparison to SB_{tk}. Differences in mean and peak pressure are significant in all plantar pressures except lateral Heel.

3.2 SUMMARY

In this chapter, the change of cushioning was investigated as a continuation of the series of perturbation studies. The first section centered around the investigation of how various cushioned midsoles, categorized as either hard or soft football shoe midsoles, may impact vertical ground reaction forces. Our research has demonstrated that artificial turf shoes with lower shore hardness (softer sole) and reduced impact peak acceleration, as observed in mechanical tests, produce higher VGRF and peak pressure compared to natural grass shoes during specific football movements such as cutting and penalty kicks. In the context of this study, the successful execution of these movements requires players to achieve an optimal traction coefficient – rotational traction for cutting and translational traction for kicking. Given that the midsole of artificial turf shoes offers greater cushioning and lower shore hardness than natural grass shoes, participants generate higher vertical ground reaction forces through the midsole of artificial turf shoes to attain the necessary traction coefficient for minimizing the risk of slippage. This study serves as a gateway to subsequent sections that aim to comprehend how athletes adapt and adjust their movements during continuous activities in controlled conditions when cushioning levels vary.

Our next section investigates the effect of different insoles on cushioning perception. The choice of insoles for sections 2 and 3 was based on their specific mechanisms. In general, insoles composed of various materials can respond to localized pressure peaks in diverse ways depending on the type and composition of the material they are manufactured from (See Figure 3.4.1). One approach involves the absorption of impact energy through material compression. Closed-cell foams and insoles featuring integrated air chambers create a counter-force when compressed, increasing inner cell pressure. This mechanism allows them to return a portion of the stored energy during unloading. In contrast, open-cell foams have a lower energy return capacity. Another way to address localized pressure peaks is the displacement of the material, a characteristic observed in materials with liquid-like properties. Examples of such materials include silicones and Sorbothane insoles used in section 2 and section 3.

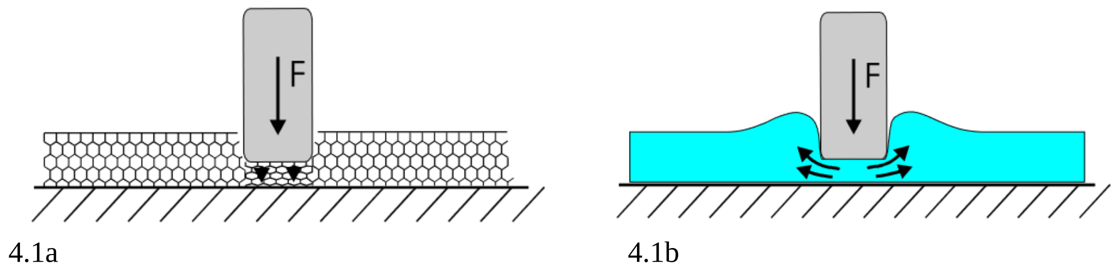


Figure 3.4.1 Schematic of two types of insoles used in this chapter **1. a** Compression of foam under load, schematic **1. b**. The behavior of a viscoelastic material under load, schematic.

In Section 2, four insoles (SB_{tk}, SB_{tn}, EMPD, and IP.GL) from two different types mentioned above were chosen from various samples and measured in the mechanical tests, and also evaluated in the subjective test. Our findings in Section 2 determined that participants (male runners) could fully distinguish the difference between all four insoles in all pair tests. In all 114 tests, participants (n=19) were able to detect the insole cushioning stimuli. The just noticeable difference in peak acceleration was detected between SB_{tn} and IP.Gl, which is 0.7 g-Max (acceleration). In other words, participants were able to distinguish a minimum increase of 6% in cushioning properties between 11.5 g-Max (SB_{tn}) and 12.2 g-Max (IP.Gl). Interestingly, such a difference existed between two insoles, namely IP.GL vs. $SB_{tn,}$ where both materials have the same behavior and are part of the second group (Viscoelastic material). When such a perturbation is imposed on the runner (different cushioning characteristics), results show that EPDM with a low cushioning feature was chosen as an insole with the highest cushioning comfort and there is a significant correlation between each $Jolt_{\alpha}$ and $Jolt_{ß}$ and cushioning comfort.

Given the rapid deceleration of the leg during landing, vibrations in the soft tissue compartments (muscle, fascia, surrounding tissue, and skin) of the leg are initiated. These vibrations can generally be influenced by three factors: the speed of running, the hardness or flexibility and vibration properties of the running surface, as well as the damping properties of the footwear. It may be speculated that increasing $Jolt_{\alpha}$ (EPDM = 23kN/s, SB $_{tk}$ = 34 kN/s, SB $_{tk}$, IP.Gl = 39 40 kN/s and SB $_{tn}$ = 50 kN/s) or $Jolt_{ß}$ = (EPDM = 34, SB $_{tk}$ = 52, IP.Gl = 54 kN/s and SB $_{tn}$ = 69 kN/s) as input signals would move toward the resonant frequency of the soft tissue package. However, another way muscles react (adapt) to harder or softer surfaces (insole) is to increase damping to minimize resonance (Wakeling et al., 2002). In this manner, $Jolt_{\alpha}$ and $Jolt_{ß}$ might be considered as the input signal. We thus assume that an increase in $Jolt_{\alpha}$ and $Jolt_{ß}$ will result in increasing the damping coefficient of tissues and therefore more intense muscle activity. Future studies should investigate muscle activity while running with these two varied insoles.

In the third section, the effect of two insoles (Foam vs. Viscoelastic) on running gait and perception was tested in laboratory conditions. EPDM insole (lower shore hardness) shows the highest cushioning comfort in comparison to SB_{tk} but with lower peak and mean pressure. Spatial-temporal paramenters such as stride length, frequency, and contact time did not differ between the two insoles in running. We finally speculated that SB_{tk} as a harder insole tends to increase vibration frequency, and potentially a tuning to the active muscle is increased (Nigg et al., 2017; Nigg et al., 2001)). In addition, there is an adaptation of their techniques with harder footwear which avoids high impact rates over the heel. It can be also speculated that adaptation in our study is related to kinetic changes in mean and peak pressure when shifting from soft insole EPDM to hard insole, SB_{Tk} to prevent muscle fatigue (Hennig et al., 1996; Milani et al.,1997). In the context of diverse cushioning perturbations, the perpetuation of spatial-temporal parameters—namely, contact time, stride length, and stride frequency—at a constant level aligns with the theoretical framework posited by Benno Nigg et al. (2017), denoted as the ‘preferred movement path paradigm.’ This paradigm postulates that individuals, external perturbations notwithstanding, tend to keep to idiosyncratic running patterns. In instances where perturbations manifest as alterations in cushioning, participants consistently adhere to their individualized running signatures. This primarily manifests through the slight but significant adjustment of kinetic parameters, while spatial-temporal attributes remain unaltered. Álterations in cushioning produce discernible modifications in peak pressure, influencing force distribution dynamics during running. Among these discernible changes in force dynamics, kinematic parameters – encompassing contact time, stride length, and stride frequency – exhibit notable constancy. This observation underscores the robustness of kinematic parameters in the face of alterations in cushioning. In summary, response to perturbations in running mechanics is profoundly shaped by the inherent biomechanical attributes distinctive to each individual. The constancy of spatial-temporal parameters, demonstrates the adaptive ability of individuals to modulate their kinetic patterns according to the specific perturbation, e.g., changes in cushioning, while at the same time maintaining their preferred movement path. This delicate interplay underlines both the complexity and adaptability of the biomechanical system.

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CHAPTER 3

(Traction)

Football played barefoot, Olympic Games, London, 1948.

The majority of the Indian football team members played the 1948 Olympic match barefoot as they were not used to playing with boots. This photo taken in dry weather depicts some Indian athletes practicing barefoot in Richmond Park, London. National Media Museum, (2012).

4.1 AIM of STUDY

Despite many positive health benefits, sporting participation as a professional footballer carries a high risk of injury (Ekstrand et al., 2011). In such field sports, contact injuries arise when players collide during ball challenges, aerial duels, or accidental impacts. These injuries encompass sprains, fractures, concussions, and muscle strains stemming from direct physical contact between players. In contrast, non-contact injuries occur when a player sustains an injury without direct physical contact with another player. These injuries are often caused by overuse, awkward landings, or making sharp movements. Common non-contact injuries in football include muscle strains, ligament tears, and tendonitis (Bezuglov et al., 2020). While football is a physically demanding sport that involves contact, the rate of non-contact injuries can indeed be relatively high due to the nature of the game (López-Valenciano et al., 2020). Players often need to perform quick and dynamic movements, which puts stress on their bodies and increases the risk of non-contact injuries, especially if they are fatigued or not adequately conditioned (Mc Call et al., 2014). A common football movement prone to such injuries is a rapid change of direction, known as "cutting" (Hootman et al., 2007; Griffin et al., 2000). In a recent comprehensive study investigating the physical demands of team field sports, it was revealed that football exhibits the most frequent occurrence of cutting movements, with players executing up to 800 cuts per game (Taylor et al., 2017).

In general, a player's ability to change direction is largely influenced by the available rotational traction (RT) between the football boot and the playing surface (Pedroza et al., 2010; Sterzing et al., 2009). RT refers to the rotation of the foot around a point of contact on the boot sole (Frederick, 1993). Although RT is often important for athletic performance, a high level of traction increases the risk of injuries such as foot fixation, anterior cruciate ligament (ACL) injuries, and joint injuries (Wannop et al., 2015; Hewett et al., 2005; Sharma et al., 1998; Shin et al. 2009; Stefanyshyn et al., 2006), while low RT increases the slipping risk (Serensit et al., 2014). Thus an optimum rotational force is required in a rotational manner (Thomson et al., 2022). One of the traditional and classical methods to evaluate RT is using player testing. This method may include some limitations such as: 1. It doesn't typically appeal to athletes as it can disrupt their training (Clarke and Carre, 2010); 2. During player testing, the laboratory environment might influence movement patterns, potentially affected by variables such as lighting conditions or the time intervals between trials (Kirk et al. 2007); 3. Inconsistent repeatability among players may arise from difficulties in managing loading conditions and performing identical movements (Clarke and Carre 2010), and subjects may change their stride length or adjust their movement so they fit the force plate which may also affect the results collected (Kaila, 2007).

To overcome such limitations, a number of researchers (e.g. Milner, 1972; Bonstingl et al., 1975; Lambson et al., 1996; Livesay et al., 2006; Wannop et al., 2012; Grund et al., 2010, Thomson et al., 2022) have attempted to measure RT in laboratory conditions by establishing a traction tester or mechanical prosthetic feet which can replicate specific football movements. However, it has been widely recognized that it is not sufficient to solely perform mechanical testing (Potthast et al., 2010). Laboratory tests with apparatus cannot reflect the response of the human body, especially the perception of traction. The latter is the result of sensory information derived from mechanoreceptors in skin, muscles, tendons, and joints which can facilitate players' adaptation to whether to changes in shoe-surface interaction. Changing the boot outsole's configuration (e.g. number of studs on the outsole and/or stud geometry) or change in surface characteristics (different hardness levels of grass/turf) play a role in the perception of traction (Brinkmann et al., 2017; Cole et al., 2023). Such a change in the perception of RT could affect the adaptation (Cai et al., 2023) and result in non-contact injuries (Fujitaka et al., 2017). Past studies poorly incorporated subjective (e.g. perception of traction) and objective tests (measuring traction with related mechanical prosthetic foot) and failed to provide an explanation of 'preferred', or/and 'excessive' or/and 'poor' traction mean and to what extent traction is noticeable.

This chapter aims to provide a methodological framework to provide a better understanding of shoe-surface interaction, especially traction. Section 1 of this chapter provides supplementary data for replication in the next Section. Section 2 describes the replication of the field test in the laboratory. In Section 3, the subjective measurement concerning perception traction is measured. Finally, in Section 4, we present a model to facilitate customizing individual outsole configurations based on artificial neural networks. The flowchart is as follows:

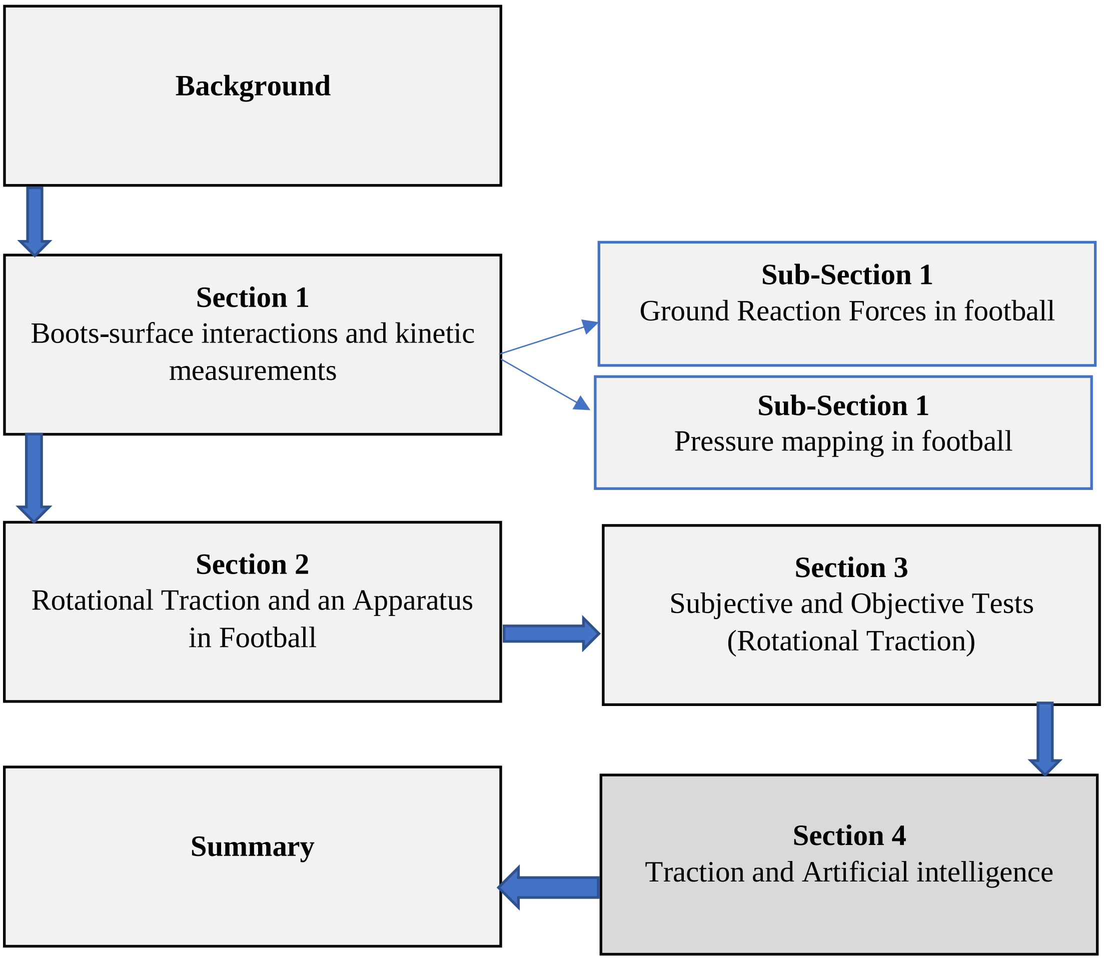


Figure 4.1 Flowchart of all sections in this chapter. The dark gray box denotes the published paper.

4.1.1 SECTION 1

Boot-Surface Interactions and Kinetic Measurements

4.1.1.1 BACKGROUND

Among several traction testing apparatuses developed since the early 1970s (e.g. Milner, 1972; Bonstingl et al., 1975; Lambson, et al., 1996; Wannop et al., 2012), few can replicate the realistic conditions of a specific football movement. However, Grund et al. (2010) established an apparatus with a mechanical prosthetic foot (TUM Tracktester) that can replicate a specific football movement based on kinetic and/or joint motion parameters. Therefore, the aim of this section was to obtain the necessary kinetic data e.g. pressure mapping and ground reaction forces, for human athletes performing cutting movements (135°) on natural grass and artificial turf under laboratory conditions.

This section includes two studies. The first study was designed to evaluate the pressure distribution on artificial turf and the second study was designed to measure ground reaction forces on natural grass. These subsections were pilot studies preparing the way for the subsequent sections describing the laboratory tests.

4.1.1.2 METHODS (STUDY 1)

A total of 11 male football players with a mean (standard deviation = ± SD) of age, body mass and height of 25.18 (± 3.0) years old, 78.8 (± 7.20) kg, and 180.0 (±5.0) cm participated in our experiment. Ten players had a preferred right kicking foot and only one had a left preferred kicking foot. Participants gave written, informed consent before the experiment. The consent form explained the confidentiality, objectives, risks of the study, and protection of personal data through appropriate procedures for anonymization. In addition, the consent form assured participants they were free to withdraw from the research at any time without giving a reason and without penalty for not taking part. This research was conducted according to the ethical standards of the Helsinki Declaration (World Medical Association Declaration of Helsinki, 2013).

A pair of turf boots – Adidas ACE 17.3– were used in this experiment in the sizes of 42.5 and 44 EU. All the original studs of the boot were flattened and smoothed with a belt sander. Three types of conical-shaped (see Figure 4.1.1) studs varying in size were made of the thermoplastic material Polyoxymethylene (POM) and produced by a Computerized Numerical Control machine (Fräsmaschine Deckel, FP 4A). These studs used for three sections (forefoot, middle, and heel) in the outsole are shown in Figure 4.1.1.

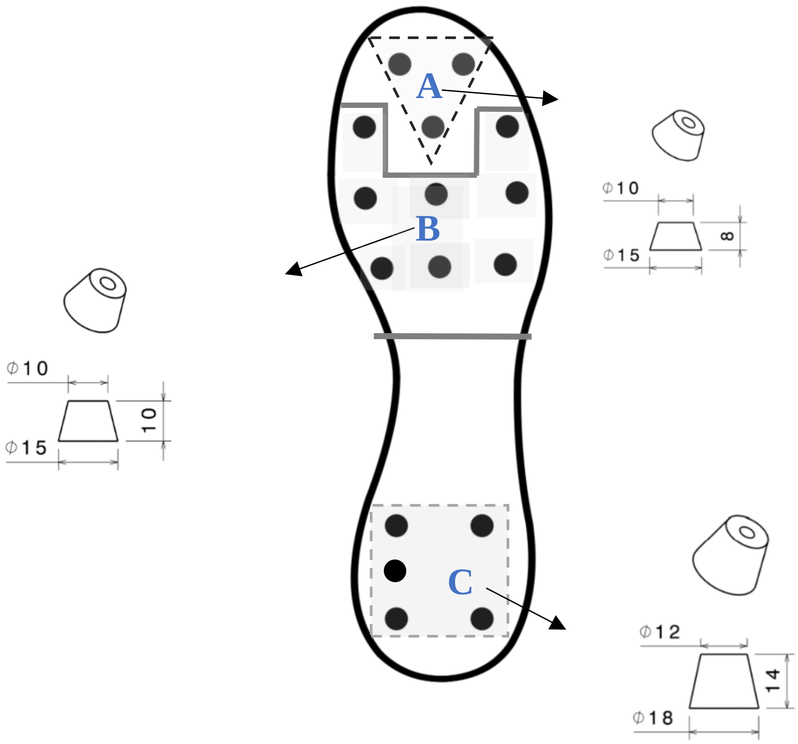
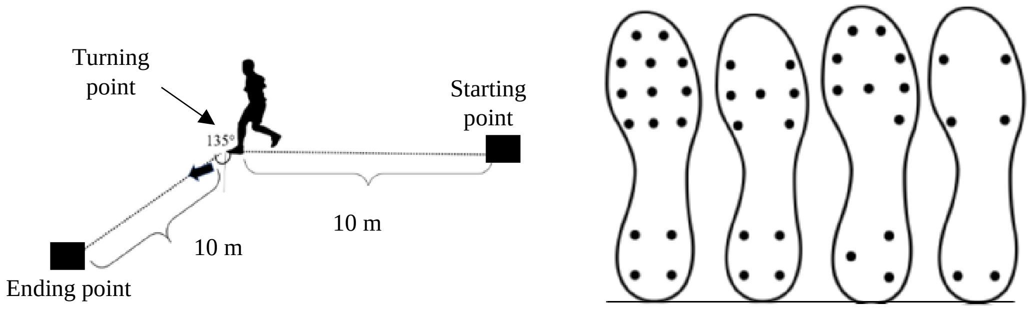


Figure 4.1.1 An outsole design with all possible stud positions and three types of studs in three sections (A, B, and C) (measurement in millimeters).

To measure the pressure distribution of all outsole configurations during cutting movements (135°), an insole sensor pressure was inserted into the boot. The Pedar Mobile system (Novel GmbH, Munich) was used to collect plantar pressure distribution in the field test. The system consisted of two flexible insoles (each containing 99 sensors in a matrix design) and a portable data logger for data storage. The sampling frequency was 100 Hz. The experimenter placed football boots on each participant to blind them to the stud configuration. Participants were asked to complete a 10-minute warm-up (five minutes of running and stretching, according to the protocol explained by the experimenter). Participants were asked to perform 135° cutting movements (See Figure 4.1.2.a) with four outsole configurations namely: full pattern, H shape, edge design, and soft-ground boots (Figure 4.1.2.b). Participants performed three trials with each outsole. Only two successful trials were randomly collected and used in the analysis phase. Participants were asked to do cutting movements with their preferred leg. For the left foot subject, the direction of the cutting movement changed. They were asked to run at 75 % of their maximum speed in all trials. The football surface used in this study was 2-star artificial turf (sand/rubber infill 100 % polypropylene) that had been used for 4 years. The surface construction consisted of a 30 mm elastic base layer (DIN V18035/7). It was covered by a fiber carpet with a pile length of 10 mm filled with sand (18kg/m^2) and rubber particles (0.5 - 1.7 granulation diameter, 5kg/m^2). The total length of the pile and carpet was 50 mm. A Clegg hammer impactor 2.25 kg (SD Instrumentation Ltd.) was used in the field test to evaluate the hardness of the surface. Ten trials were performed at three spots (starting, turning, and ending points). The compliance date of the Clegg hammer allowed replication of the same playing surface condition in the laboratory tests that comprise subsequent chapters. The unit of surface hardness is ‘g-Max’.



a b

Figure 4.1.2.a Sketch of cutting movement used in the experiment. **b** Four outsole configurations used in the experiment (from left to right: full pattern, H shape, edge design, and soft-ground boots).

Before the study, the sensory perception threshold of the participants’ plantar foot surfaces was evaluated using Semmes-Weinstein monofilaments. Nine areas of each subject were evaluated during the examination (Thaniyath et al., 2021; Barbic et al., 1985) for both the right and left feet (see Figure 4.1.3). The monofilament was pressed on defined spots until the bending point against the area of examination formed a c-shape, exerting a force equivalent to the monofilament specification. Patients without neuropathy should be able to sense the monofilament with the evaluator size of 3.61 which is equivalent to 0.4 g of linear force (One-g force is equal to 10 N) (Tanenberg et al., 2008). Moreover, to obtain peak and mean pressure data, insole-sensor pressure, the Pedar Mobile system (Novel GmbH, Munich) was divided into ten sections following Wong et al. (2007) (See Figure 4.1.4).

Figure 4.1.3. Nine Monofilament test locations.

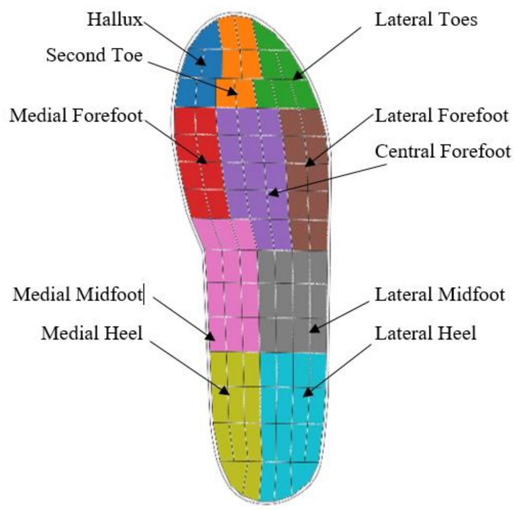


Figure 4.1.4. Plantar pressure sections were used for Data analysis.

4.1.1.3 RESULTS (STUDY 1)

The pressure mapping, mean and peak pressures (kPa) of the preferred foot were measured for all participants regarding the ten plantar sections. (See Figure 4.1.5 and Table 4.1.1). Across all four outsole designs, the medial forefoot and hallux sections showed the highest peak and mean pressures in the fore and midfoot zones. In the rear zones, the medial heel exhibited higher peak pressure in comparison to the lateral heel. In addition, the heel section pressures were lower in comparison to the medial forefoot and hallux sections.

The monofilament data revealed that participants could discriminate pressure with high accuracy (right foot 93 % accuracy (SD = 11.4 %) and left foot 95 % accuracy (SD = 6 %). The plantar sections that exhibited lower sensitivity in the Semmens Weinstein monofilament test were generally in the heel sections. This may be due to thicker plantar skin which may players develop over time. In addition, the hardness of the surface measured by the Clegg hammer impactor 2.25 kg was equal to 155g (SD = 4.9).

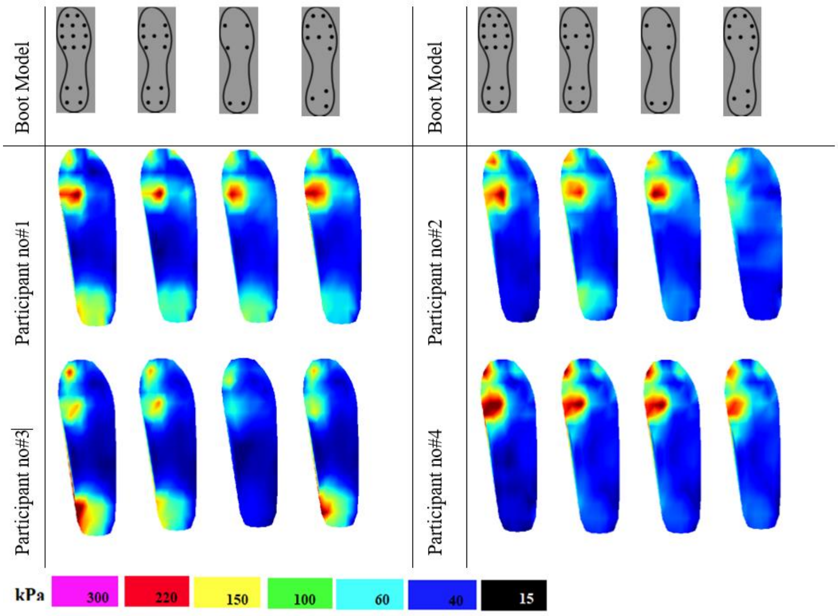
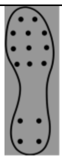
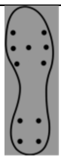
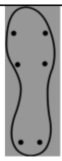
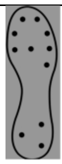
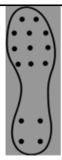
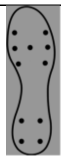
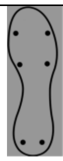
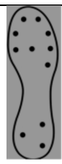


Figure 4.1.5 Morphology of pressure mapping for four outsole boots (random four participants).

Table 4.1.1 Mean and peak pressure of four outsole configurations in cutting 135°

		Mean Pressure (kPa) Boot Model				Peak Pressure (kPa) Boot Model			
***Abbreviation**	**Section**								
FF	Hallux	199.94 (43.19)**	193.59 (52.08)	196.69 (44.14)	192.94 (16.80)	278.85 (53.83)	261.43 (59.70)	303.43 (60.04)	265.50 (30.26)
FF	Second Toe	111.58 (19.59)	92.54 (21.54)	93.50 (15.14)	96.97 (15.68)	157.62 (34.32)	135.52 (36.73)	146.19 (29.29)	142.32 (30.42)
FF	Lateral toes	77.54 (24.15)	75.97 (18.16)	74.50 (17.84)	74.08 (21.76)	102.50 (35.67)	101.35 (23.59)	110.30 (25.32)	103.75 (30.72)
FF	Medial forefoot	207.45 56.94	224.23 (48.78)	154.50 (45.02)	232.35 (35.09)	278.49 81.64	291.96 (54.08)	269.73 (52.31)	300.21 (41.27)
FF	Central forefoot	125.55 (28.28)	120.28 (26.99)	99.64 (28.28)	95.28 (38.81)	178.07 (42.19)	165.69 (32.96)	162.20 (39.86)	145.08 (44.41)
FF	Lateral forefoot	66.53 (37.69)	86.28 (28.93)	56.90 (16.14)	66.54 (32.96)	113.14 (50.87)	121.09 (37.10)	106.14 (28.06)	102.96 (41.89)
MF	Medial arch	35.02 (21.75)	27.04 (16.00)	38.97 (11.44)	34.66 (15.09)	59.79 (19.66)	45.26 (16.45)	68.75 (15.64)	60.54 (14.07)
MF	Lateral arch	35.22 (20.08)	32.85 (13.24)	35.87 (7.70)	33.68 (12.43)	55.95 (23.31)	52.53 (13.31)	60.07 (9.73)	55.12 (16.88)
RF	Medial heel	69.70 (41.10)	50.85 (27.48)	54.43 (21.15)	78.06 (38.10)	113.07 (58.53)	108.80 (64.25)	136.82 (64.93)	146.20 (70.01)
RF	Lateral heel	40.06 (15.59)	32.46 (15.09)	42.19 (11.65)	46.45 (12.12)	79.98 (49.02)	69.17 (44.85)	108.61 (41.62)	92.98 (37.46)

*FF= forefoot, MF= midfoot, RF= rearfoot. ** standard deviation
Highlighted rows determined remarkable changes in the forefoot (Hallux and medial forefoot) and the rear foot (medial heel).

4.1.1.4 METHODS (STUDY 2)

Participant. A male football player from the regional SV Igling (Liga B-Klasse) league participated in the experiment. With over three years of football experience, he was a right-footed player, 19 years old, standing at 178 cm tall, and weighing 64 kg. The football boots used in this study were the Nike® *Phantom Vision 2 Elite Dynamic Fit FG,* size 42.5 EU.

Equipment. To measure plantar pressure distribution, an insole sensor pressure system, Pedar Mobile (Novel GmbH, Munich), was inserted into the right and left boots. The system consisted of two flexible insoles (each containing 99 sensors in a matrix design) and a portable data logger for data storage. Ground reaction forces were measured with a force plate (Kistler 9286 AA: Kistler Instruments AG, 2008). An appropriately sized hole was dug into the ground and then a level concrete foundation was poured into it (Figure 4.1.6). A steel plate (600 mm × 400 mm × 15 mm) with a mass of 40 kg was attached to the concrete foundation. To avoid slippage of the force plate, the sand paper was affixed to the steel plate. Finally, the Kistler force plate was set on top of the steel plate according to Figure 3.1.2.1. Before the study, four "pitch boxes" (570 × 370 × 70 mm) were filled with soil and planted with grass seed. Over 15 days, the grass was watered, and exposed to daylight and the grass was allowed to grow to a typical pitch condition. These boxes were made of polypropylene and each had a volume of 15 L. The total height of the pitch boxes (soil plus grass) was 60 mm. Each box was placed on top the of force plate (See Figure 4.1.6) per trial. In addition, the surface hardness was measured by A Clegg hammer impactor (SD Instrumentation Ltd). It had a 2.25 kg missile with an accelerometer at the end that was dropped once through a guide tube from a height of 45 cm to measure deceleration on impact reported as gravities (g). Clegg hammer data was measured on 5 spots on a pitch box at the beginning of the experiment: four trials in the four corners and one in the center. The components of the pitch box are shown in detail in Figure 4.1.7.

Main experiment. The participant was equipped with insole-sensor pressure Pedar X-system (Novel GmbH, Munich Germany). After 10 minutes of warm-up, the participant was asked to run and perform 135° cutting movements. After each trial, the grass pitch box was changed out and replaced with a new pitch box. The insole sensor pressure was sampled at 100 Hz and the force plate sampling rate was 1000 Hz. In total, the participant had two successful trials obtained by the experimenter. In addition, the velocity was 3.5 m. s $^{-1}$ and was measured and controlled by the SmartSpeed Pro timing gate system.

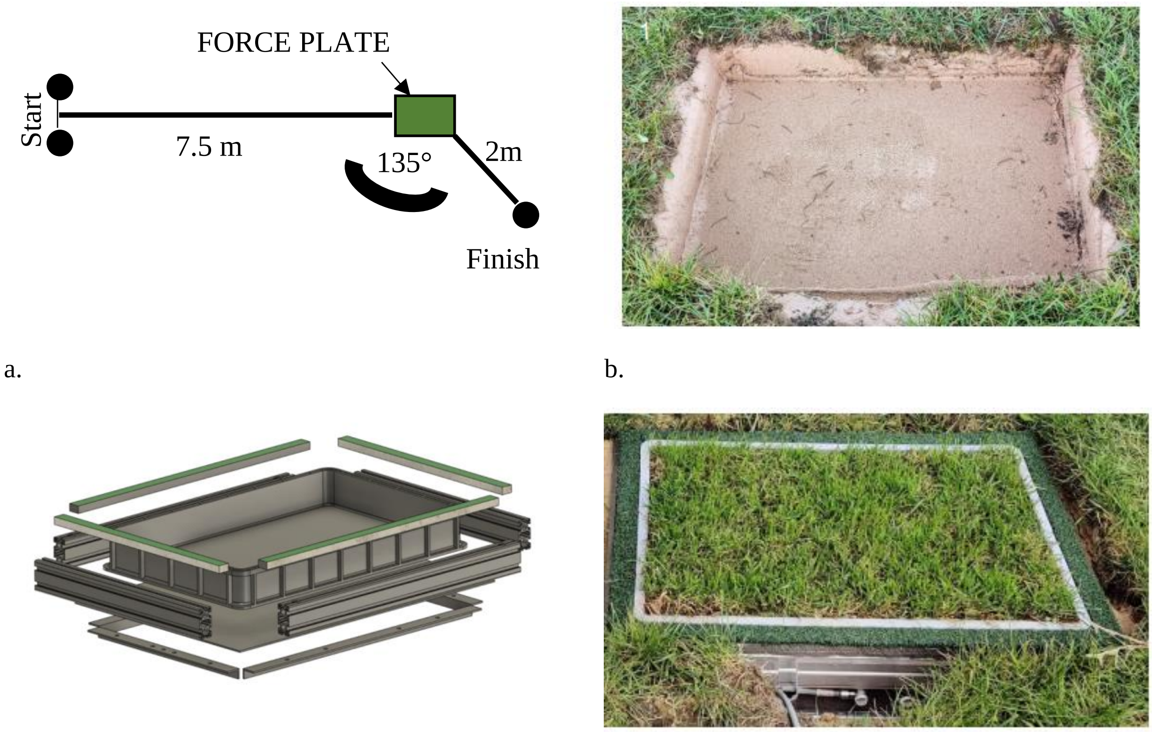
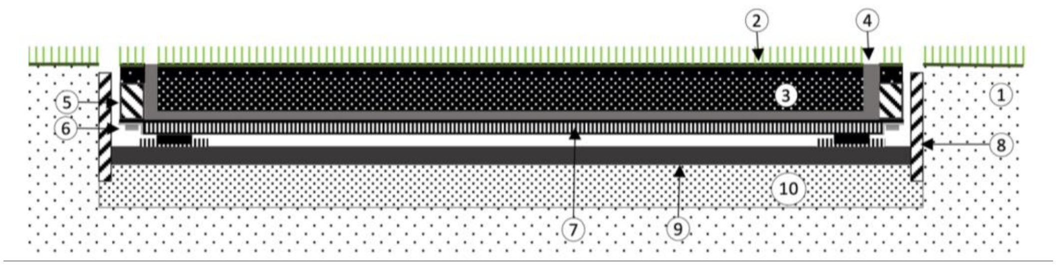


Figure 4.1.6.a Sketch of experiment movement (cutting Manouver 135°). **b** A tile to place the faceplate **c.** General overview of the box with outer frames **d.** Finally implemented force plate infilled with the natural grass.



1.	Natural grass	6.	Profile mounting screws
2.	Turf top layer	7.	Force plate
3.	Turf base layer	8.	Acrylic glass frame
4.	Plastic box	9.	Steel foundation plate
5.	Aluminium frame	10.	Concrete

Figure 4.1.7 Layering of the force plate construction.

4.1.1.5 RESULTS (STUDY 2)

Exemplary ground reaction forces data are shown in Figure 4.1.8. The peak vertical force components reached F_z = 4.12 BW, F_y = 2.39 BW, and F_x = 0.94 BW. The hardness of the ground was 71.2 g (SD = ± 4.9). When the peak vertical ground reaction force occurs during the cutting movement, the peak pressure data reveal that three sections on the plantar surface play a major role: the hallux, medial forefoot, and heel.

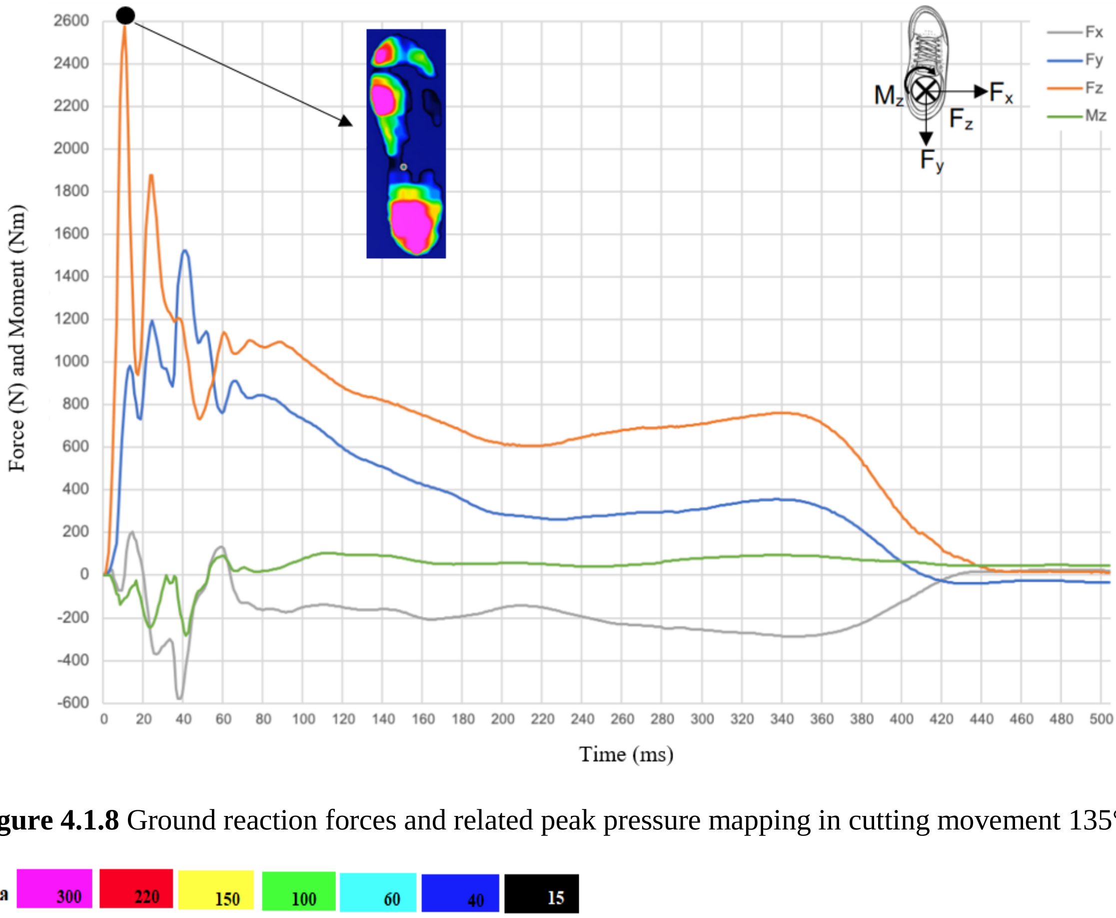


Figure 4.1.8 Ground reaction forces and related peak pressure mapping in cutting movement 135°.

4.1.1.6 DISCUSSION

Our findings of sub-section I showed that during a cutting movement, the medial forefoot and hallux sections have the highest mean pressure and peak pressure in the forefoot and midfoot zones. In the rearfoot zone, the medial heel section exhibited the highest peak and mean pressures in comparison to the lateral heel section. Similar to our study, Hennig et al. (2010), Wong et al. (2007), Eilse et al. (2004) and Orendurf et al. (2008) measured the plantar pressures in soccer-specific movements and those research groups also found that the pressures on the medial side of the foot plantar surface are greater and that may be reflected in football injuries. In all those studies, these three plantar sections exhibited higher peak and mean pressure in comparison to other sections.

However, there are some differences between the present findings and some of those studies. In the present study, the ranking of the three highest mean and peak pressure sections were the medial forefoot, hallux, and medial heel. In the study of Wong et al. (2007), the highest mean pressure rankings were: medial forefoot, hallux, and medial heel. In the study of Eilse et al. (2004), the highest peak pressure sections were: medial forefoot, hallux in forefoot and midfoot, and medial heel in the rearfoot zone. In the study of Orendurf et al. (2008), the highest pressures were: medial forefoot, hallux in forefoot and midfoot zones, and medial heel in the rearzone. Finally, in the study of Hennig et al. (2010), the highest peak pressure sections were the medial forefoot, hallux in the forefoot and midfoot zones, and medial heel in the rearfoot zone. Although these studies determined three common sections dominated the peak and mean pressures exerted, the ranks of these three sections differed somewhat. It is hard to find consensus on this matter, due to the number of extrinsic factors that may play a role including different training patterns, different cutting movements, different outsole configurations (Lv et al., 2019; Bentley et al., 2011), and effects of different surfaces (Tessutti et al., 2010). In the Subsection II study, the peak vertical forces reached: F_z = 4.12 BW, F_y = 2.39 BW, and F_x = 0.94 BW. These vertical forces are higher than Smith et al. (2004), Blackburn et al. (2003), and Simonsen et al. (2000) who determined it in the range of 2 BW to 2.6 BW. However, the present Fz is lower than Brock et al. (2014) who recorded up to 5 BW. It should be noted that the type of surfaces and types of outsole configurations differed among the abovementioned studies so comparisons need to be cautious (Lv et al., 2019; Smith et al., 2004). Changes in pitch surface and shoe outsoles may explain the different findings. Changes in surface hardness and stud length affect the velocity of stud penetration depth and accordingly rotational traction (Soe et al., 2022). Future studies should quantify and compare the rotational traction when using different boots and/or different surfaces.

4.1.1.7 CONCLUSION

Our findings indicated that during cutting movements, two-foot plantar sections exhibit the highest mean and peak pressures: the medial forefoot and the hallux. The medial heel section had the highest mean and peak pressure in the rearfoot zone. Our findings in Subsection II show that vertical ground reaction force can reach up to 4.1 BW. These actual field measurements on human athletes may be useful for evaluating the ecological validity of laboratory measurements that use mechanical prosthetic feet with actuation.

4.1.2 SECTION 2

An Apparatus and Rotational Traction

4.1.2.1 INTRODUCTION

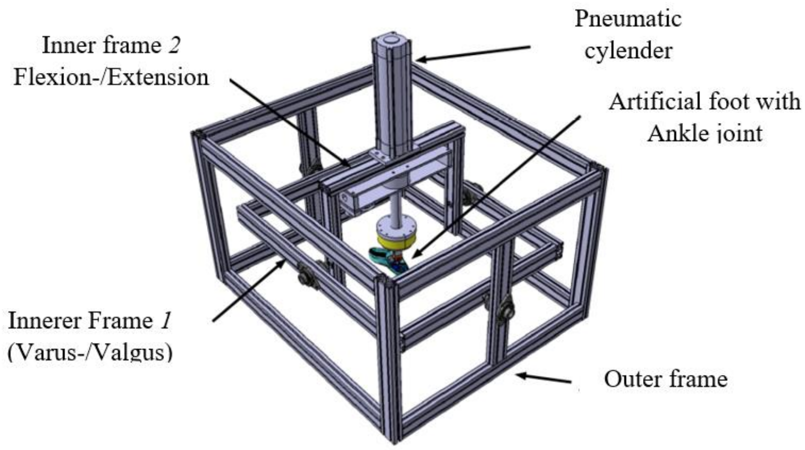
Most common injuries that occur during football matches come from torsion of the leg during rapid changes in direction (Waldén et al., 2011). Injuries occur frequently at the time of pivotal rotation when the foot is stationary, and range from damage to the knee cartilage and ligaments to fractures of the tibia bone (Torg and Quedenfeld, 1971; Yu et al., 2022). Past studies show that achieving optimum traction, not necessarily maximal traction, may reduce injuries and increase performance. (Stucke et al., 1984; Valiant, 1993).

Rotational traction is defined as the rotational force required to release the studs from the playing surface (Kent et al., 2021; Wannop and Stefanyshyn, 2016). Several extrinsic factors such as football boot design may affect rotational traction and accordingly the abovementioned injuries. Sufficient traction between a football boot's outsole and the playing surface greatly influences a player's ability to accelerate, decelerate, change direction, and perform other football-specific movements (Pedroza et al., 2010; Dowling et al., 2010; Schrier et al., 2014; Sterzing et al., 2009). An optimal range of rotational traction values may exist whereby performance is augmented yet the risk of injury is moderated (Frederick, 1993; Luo and Stefanyshyn, 2011; Wannop et al., 2013; Thomson et al., 2015; Stucke et al., 1984; Valiant, 1993). One method for measuring rotational traction is mechanical testing. It has the advantage of being a repeatable measure and can create objective loading conditions. Numerous researchers have attempted to measure lower extremity injury risk by measuring the rotational forces that occur between studs of football boots and playing with traction tester apparatus. In the early 1970s, Milner (1972) measured the traction developed at the shoe-turf interface by using a drag test apparatus. This apparatus utilized an Instron tensile test machine to measure the forces necessary to initiate and sustain translation when pulling a cleated shoe across the turf. In 1975, Bonstingl et al. (1975) claimed that "…if the maximum amount of dynamic friction between surface and shoe that could be safely tolerated was established and a shoe would release when the applied torque become hazardous, many knee injuries would reduce in severity". Accordingly, they established a testing device to simulate the shoe-surface traction system. It consisted of a simulated leg, a weighted pendulum and strain gauges mounted onto the leg to measure effective torques developed in the shoe-turf interface. Lambson et al. (1996) investigated the role of torsional resistance on the Anterior Cruciate Ligament among different football boots with different cleat designs. To reach this aim, they established an apparatus consisting of a prosthetic foot mounted onto a steel shaft and vertically loaded with 45.5 kg. A torque wrench, attached to the steel shaft of said machine, moved through an arc of 90 degrees in 1 second. Early in the 21st century, Livesay et al. (2006) fabricated a device to evaluate the torque developed at the shoe-surface interface. It consisted of a vertical shaft connected to the forefoot of the shoe and used to generate relative motion between the shoe and the surface. Compression load and torque were measured using an in-line torque/thrust sensor and rotation was measured using a rotary potentiometer. Relative rotation between the shoe and the surface was measured using a custom-built goniometer that generated a voltage potential proportional to the angle of rotation. Wannop and Stefanyshyn (2013) established the first portable traction tester for rotational and translational traction. For rotational traction, the vertical shaft was unlocked and free to rotate, with the rotational axis of the tester in line with the forefoot. By unlocking the vertical shaft, the moment generated was determined by multiplying the force measured by the force transducer with the moment arm of the shaft. The mechanical testing should be as biofidelic as possible, otherwise, the results risk being interpreted incorrectly (Fleming et al., 2011). To achieve biofidelity, laboratory mechanical testing with a traction tester apparatus should mimic the foot pressure patterns measured *in vivo* during specific football movements.

To the best of our knowledge, none of the abovementioned apparatus-traction testers have measured plantar pressure distributions. To overcome this barrier, the pioneering TUM *Tracktester* (Grund et al., 2010) was developed to replicate football-specific movements while measuring pressure distributions with ground reaction forces up to 1500 N. The data from the previous Subsections were used in this current section to replicate a realistic cutting movement (135°) and to calculate rotational traction for different shoe outsole designs. This study aimed to realistically measure rotational traction among three outsole configurations (full-stud, H-shape, and soft-stud designs) on natural grass in laboratory conditions.

4.1.2.2 METHODS

Equipment. An apparatus called TUM Tracktester was used in this section to replicate the cutting movement (135°) which was performed by an actual athlete in the field test (please see Subsection II). The TUM Tracktester is a custom-made rig developed to provide insight into the stud-surface interaction under realistic football game situations. The TUM Tracktester orientates (see Figure 4.2.1.a) an artificial leg shaft concerning an artificial foot and allows for foot plantar pressure mapping with an adjustable angular foot-leg position that might be observed during an injury situation in cutting movement. The TUM Tracktester emulates realistic human anatomy. The angular position of the shaft for the lower leg concerning the artificial foot model can be adjusted in the sagittal plane (flexion and extension) and the frontal plane (valgus/varus). The preload in this version can reach up to roughly 1500 N. The artificial leg comprises a vertical cylinder along the shaft, a pneumatic muscle (Muscle SIM, Festo AG & Co. KG, Esslingen, Deutschland, F_{Muscle} = Force responsible for rotational deflection) to induce rotation, and a 6-component load cell to record the three orthogonal force and moment components (ATI-IA Omega160, Apex, NC, USA). To mimic realistic pressure mapping of the plantar foot surface, a foot model made of steel and covered with silicon was used (see Figure 4.2.1b-c). This foot model mimics human skin and soft tissue compliance during rotational movements. Moreover, to monitor/replicate *in vivo* pressure distribution, an insole-sensor pressure Motion (Munich, GmbH) was inserted between the foot model and the boot. This insole contains 13 capacitive pressure sensors, and 52% of the insole is covered by sensors. The Moticon system can measure pressures up to 40 N/cm^2 at 50 Hz.



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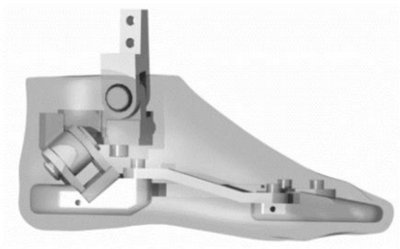
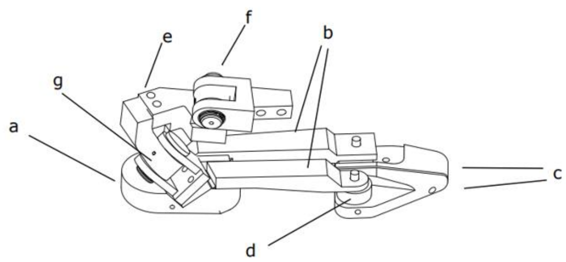




1. **ii** 1. **iii**

Figure 4.2.1.i TUM Tracktester and its components. **1. ii** Foot model with silicon upper layer, **1. iii** Foot model with its components: a-Rearfoot, b-Midfoot Struts, c-Forefoot, d-Robber bonded metal, e-Ankle, f- Uppr ankle joint, g-Lower ankle joint

Football boot preparation. To conduct the laboratory tests on the TUM Tracktester, the studs of the football boot, the Adidas Neoride III (size 42.5), were ground down flat with a belt sander. Then, 16 possible stud positions (see Figures 4.2.2 and 4.2.3) were created with boreholes on the outsole and equipped with threaded inserts. Three types of conical-shaped studs of varying lengths and cross-sections were created for the three plantar zones (forefoot, midfoot, and heel) on the outsole (see Figure 4.2.2 and 4.2.3). These studs were made of the thermoplastic material Polyoxymethylene (POM) and produced by a computerized numerical control (CNC) machine (Fräsmaschine Deckel, FP 4A). These studs were used according to the three outsole designs shown in Figure 4.2.4. The outsole material was Polyurethane.

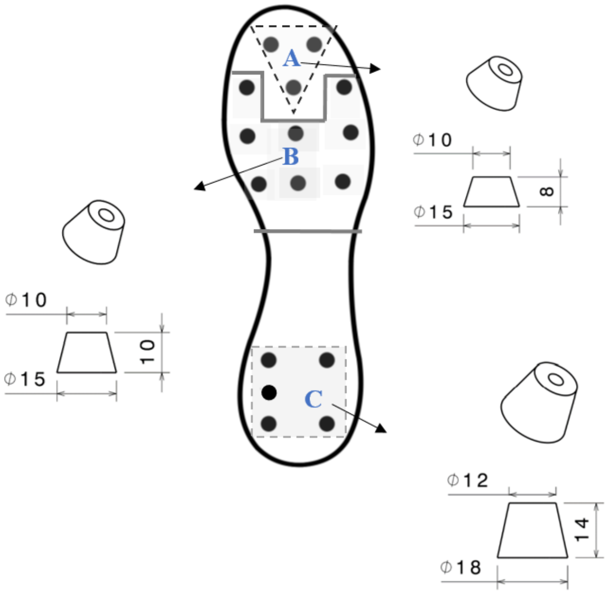


Figure 4.2.2 Three sections of football boot outsole A-B-C. Each section is defined with its stud geometrics (Adidas Neoride III, shoe size 42.5).

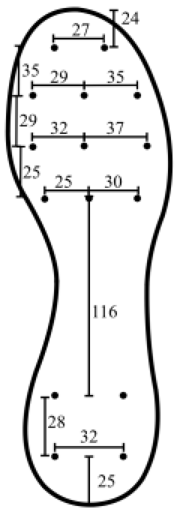
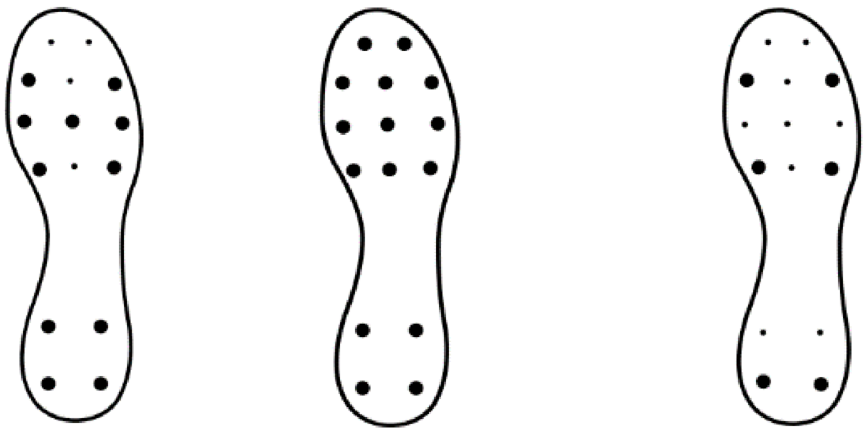


Figure 4.2.3 All possible stud positions and their distance from each other (Adidas Neoride III, shoe size 42.5)



Pattern A Pattern B Pattern C

Figure 4.2.4 Three studded outsole designs were used in the experiment (A H-shape pattern, B= Full-stud pattern, and C= soft-ground pattern.

Surface preparation. The surface used in this experiment was natural cold-season grass (Schwab Rollrasen ©). The pile height was trimmed to 25 mm and the combined thickness of the sand and grassroots was 35mm. The hardness of the surface was measured by a Clegg hammer impactor (SD Instrumentation Ltd.). The impactor incorporated an accelerometer that read in units of multiples of g (g = 9.81 m/s^2). The surface hardness averaged 59.2 g (SD = 1.3). The grass tile was laid over an experiment box in the size of 800 × 900 × 700 mm which was filled with soil and natural grass carpet with a thickness of 35 mm and grass length of 25 mm.

The combination of soil and grass was maintained in laboratory conditions for 3, days at 21° C and with proper maintenance (water, mowing, and cutting the grass). In the end, 15 boxes were prepared for the day of the experiment. Each box was fixed with clamps to minimize the deformation of the box walls. The grass tiles were changed after 3 trials (in case of severe damage, the grass tile was immediately changed).

Procedures. The football boot without studs was placed over the foot model. The outer layer and inner layer of the TUM Tracktester were adjusted to dorsal flexion at 30° and supination at 20°. That orientation produced the desired pressure pattern (see Figure 4.2.5). The baseline boot was replaced with one of three patterns (randomly chosen). Rotational tractions were measured for 5 trials under preloads of 700N, 1100 N, and 1400 N. All three boots were tested in 5 trials. This procedure occurred for all three shoes with a transformation matrix used for measuring RT.

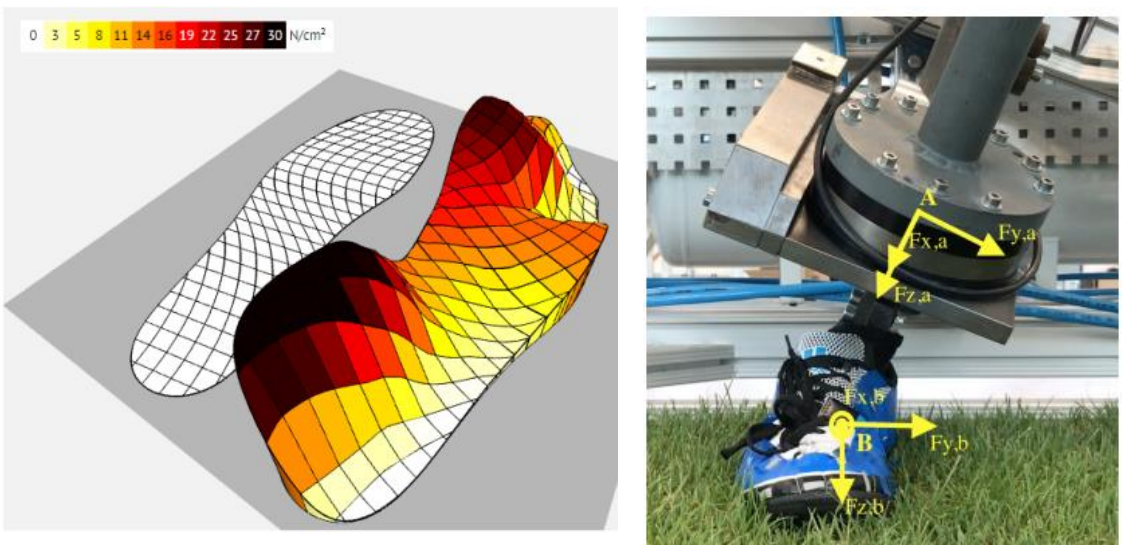


Figure 4.2.5 Left: Schematic of pressure distribution while performing cutting movement using TUM Tracktester; schematic of shoe surface interaction. **Right**: Point A is the measuring point of the actual load cell: Point B: is the coordination center of the spot, where the torsion is created.

4.1.2.3 RESULTS

The rotational traction was measured and calculated for all three boots (Table 4.2.1-3 and Figure 4.2.6). Our results show that the full-stud pattern had the highest RT on natural grass in all possible conditions (700, 1100, and 1400 N). Rotational tractions of three shoes were close to each other with a preload of 1100N. However, in the preload of 700 N and 1400 N, the differences were apparent. Rotational traction increased with increasing preloads in all boot designs tested. Our findings determined that the full-stud pattern boot had the highest rotational traction (7 Nm-10 Nm) among all boots tested at all three pre-loads. The H-shape stud pattern (pattern B) had the lowest RT for preloads of 700 N and 1100 N. With the highest preload, 1400 N, football boots with the soft ground stud pattern exhibited the lowest rotational traction (8 Nm).

Table 4.2.1 Rotational Traction measurement in boot pattern 1

	Pre-load			
PT1	#trial	700N	1100N	1400N
		RT (Nm)	RT (Nm)	RT (Nm)
	1	7.04	7.70	8.10
	2	5.42	7.90	8.60
	3	5.60	7.30	9.10
	4	5.83	7.42	9.09
	5	5.73	7.48	8.87
	Mean	5.92	7.56	8.75
	(SD)	*(0.64)*	*(0.24)*	*(0.42)*

Table 4.2.2 Rotational Traction measurement in boot pattern 2

	Pre-load			
PT2	#trial	700N	1100N	1400N
		RT (Nm)	RT (Nm)	RT (Nm)
	1	6.92	8.12	10.19
	2	7.20	8.25	10.11
	3	7.05	8.25	10.14
	4	7.17	8.06	10.11
	5	7.08	7.84	10.16
	Mean	7.08	8.10	10.14
	(SD)	*(0.11)*	*(0.17)*	*(0.03)*

Table 4.2.3 Rotational Traction measurement in boot pattern 3

	Pre-load			
PT3	#trial	700N	1100	1400
		RT (Nm)	RT (Nm)	RT (Nm)
	1	6.37	7.27	8.19
	2	6.78	7.58	7.78
	3	6.82	8.02	7.81
	4	6.75	7.96	7.96
	5	6.79	8.18	8.13
	Mean	6.70	7.80	7.97
	(SD)	*(0.19)*	*(0.37)*	*(0.18)*

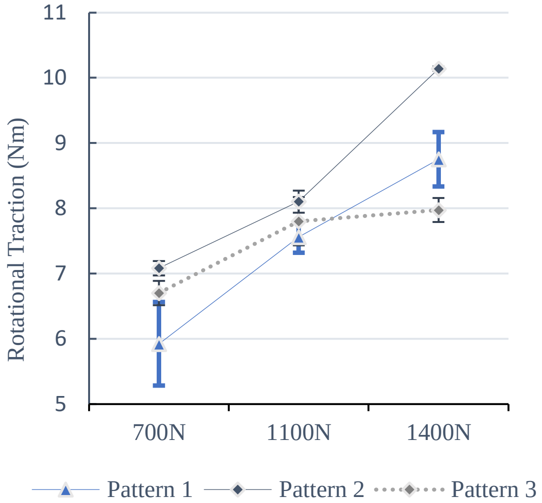
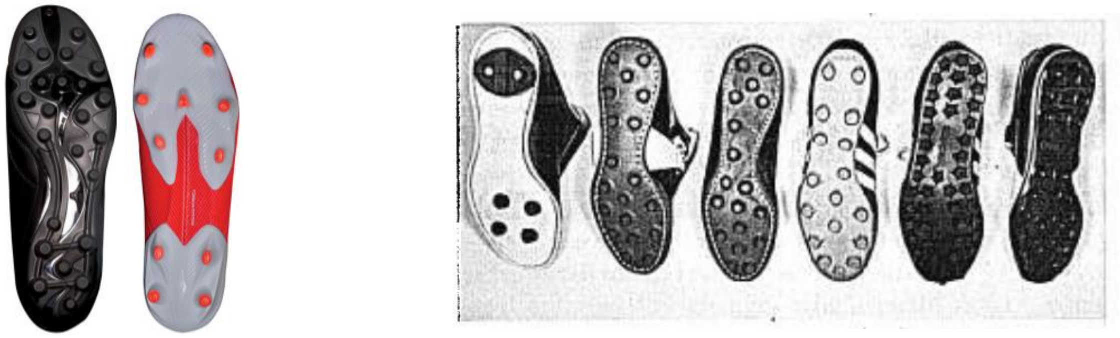


Figure 4.2.6 Rotational traction of three boots in three loads

4.1.2.4 DISCUSSION

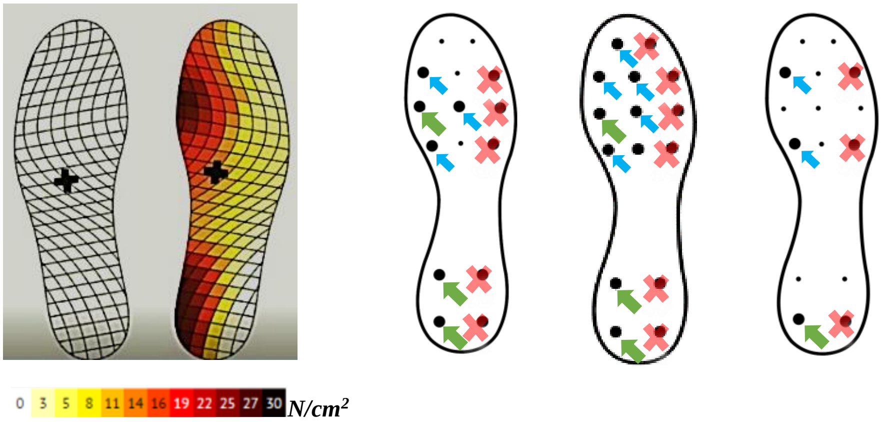
Overall, we found that different outsole configurations altered the mechanical traction properties. In detail, for the full-stud-pattern boot, an increase in the number of studs in the sections of the hallux, medial forefoot, and heel increased rotational traction. In our study, an increase in stud number increased traction up to 20 % with the preload of 1400 N. In the other preloads such as 700 N, and 1100 N, the traction increased up to 16 % and 6 % respectively. Such findings are in accordance partially with Bonstingl et al. (1975) and Lambson et al. (1996) who determined that increasing the number of stud results in higher rotational traction up to 16 % and 68 % respectively. In those two abovementioned studies, the preloads were set at 450 N and 900 N. However, our findings differ from those of Thomson et al. (2022). His findings determined that when changing the outsole pattern to a higher number of studs, the rotational tractions decrease which conflicts with the present findings, Bonstingl et al. (1975), Lambson et al. (1996) and those of Castillo-Domínguez et al. (2022). It is important to note that changing only one parameter, e.g. stud shape, can affect penetration depth (Dé et al., 2014). In the studies of Thomson et al. (2022) and Bonstingl et al. (1975) in comparison to my study, stud number, and stud position were also different noticeably, in addition to stud shape, (see Figure 4.2.7). These changes can influence penetration depth and penetration velocity and accordingly traction (Soe et al., 2022).



a.I a.II b.

Figure 4.2.7. a.I Two football boots the boot on the left a.I Boot in left Mizuno Monacrida AG. **a.II** The right side: Boot in left Adidas nemesis FG (Images adapted from Thomson et al. (2022)). **b** Types of boots with a similar shape tested in the study of Bonstingl et al. 1975 (Image adapted from Bonstingl et al.).

In our study, the number of studs that are involved actively in cutting movement was *six* in *pattern 1* (see Figure 4.2.8). Three of these six studs have a major influence on peak pressure and penetration, and accordingly on traction (green arrow where the peak pressure reaches 30N/cm^2) while the other studs (three blue arrows) had only a minor effect (lower than 25 N/cm^2). The number of studs that were involved actively in *pattern 2* was nine. Three of these nine studs had a major influence on peak pressure, penetration, and accordingly traction (green arrow where the peak pressure is 30 N/cm^2) while the other six studs (blue arrow) had minor effects. The number of studs that were involved actively in *pattern 3* was three. One of these three studs had a major influence on peak pressure, penetration, and accordingly traction (green arrow where the peak pressure is 30 N/cm^2) while the other two (blue arrow) had smaller effects.



a. b.

Pattern 1 Pattern 2 Pattern 3

Figure 4.2.8. A pressure mapping was obtained while cutting 135° using a TUM Tracktester. **b** Pressure mapping and stud patterns (green and blue arrows show the major and minor effects of studs in traction. The red cross shows poor contribution in rotational traction (poor penetration).

Increasing the preloads from 700 N to 1400 N causes a strong interlocking of active studs into the natural grass. Since the thickness of the grass was 30 mm, we assume that the studs indicated with the green arrow in Figure 4.2.8 penetrated the grass and locked in deeper layers (close to the root of the grass). This increased the rotational resistance and accordingly, affected the generated torque. This statement should be taken cautiously because the TUM Tracktester is a pseudo-static machine that does not fully replicate *the dynamic* movement of players. The present testing reached was a slow penetration of soil. Penetration depth is related to the velocity of impact force (Soe et al., 2022). Therefore, dynamic penetration depth in comparison to pseudo-static movement may indirectly affect soil cohesion, interlocking, and ultimately, rotational traction.

The current study illustrated how stud patterns affect rotational traction. However, it cannot indicate whether RT is excessive or suitable among those boots. Future studies should establish a field test using subjective measurement among these bespoke boots. Such a future test design can provide insight into perceived traction and its linkage to the traction tester. The subsequent section concentrates on that topic.

4.1.2.5 CONCLUSION

In this section, cutting movement (135°) was emulated using a TUM Tracktester to evaluate the rotational traction of three football boots. Our findings show that an increase in the number of studs increases rotational traction. The full-stud pattern showed the highest rotational traction with loads up to 1400N in comparison to other boots. Such a design (concerning the geometry used in this study) should be considered as a risky outsole on natural grass due to the increased grip which may result in an increased risk of ACL and other injuries.

4.1.3 SECTION 3

Subjective and Objective Tests (Rotational Traction)

4.1.3.1 INTRODUCTION

Successful completion of sensorimotor activities requires appropriately interacting with one's surroundings, whether it is a simple activity like closing the door or a complex activity like a rapid change of direction in football. These sensorimotor tasks necessitate not only generating desired movements but also accurate perception of movements (Cole and Waterman, 1995). While movement generation and perception can be identified as distinct elements of sensorimotor control, these processes do not occur independently of one another. That is, the generation of a movement is affected by the perception of what is happening; and, vice versa, the perception of a movement is affected by how the movement is generated (Sperry, 1950; Von Holst and Mittelstaedt, 1950). The perception of the force applied to a ground surface is derived from mechanoreceptors in the skin, muscles, tendons, and joints that respond to the consequences of torques applied to human tissue, though it should be noted there are no biological sensors devoted to torque. When sensory receptors are exposed to stimuli, a phenomenon known as sensory adaptation takes place (Stuck et al., 1984).

An important component of movement is the generation of forces around a joint, i.e. torques (Cai et al., 2023). In a complex movement like cutting in football, extrinsic factors such as surfaces and boots affect the torques required and generated (Scherier et al., 2014). Boots and surfaces (as environmental factors) were perceived by 25 % of players as a risk factor for inducing injuries (Zech et al., 2017). These extrinsic factors can change frequently between training and competition (Rago et al., 2019), or even during a game from the first half to the second half (because of variations in moisture level and surface hardness). Changes in boot-surface interaction require proper adaptation by a player, otherwise, the risk of injuries is greater (Damm et al., 2013). Fujitaka et al. (2017) investigated player adaption in a 12-year study among 397 football players. They investigated sports injuries on natural grass over 4 years and then on artificial turf for 8 years. They observed a significant increase in lower extremity injuries one year after players changed from natural grass to artificial turf. Such a scenario raises a question: How accurately can players perceive a change in sports equipment performance? Specifically, can football players perceive the ground reaction torque during cutting movements?

Past findings indicate that the amount of traction is linked to outsole configuration (Grund et al., 2010; Thomson et al., 2019; Stefanyshyn et al., 2010; Twomey et al., 2012; Villwock et al., 2009). Most of the abovementioned studies measured traction in laboratory conditions using an apparatus called in general traction tester. Complex models of traction testers exist; however, it remains unknown how well the mechanical measurements in the lab reflect player traction perceptions of football boots. Some researchers have investigated perceived traction in a subjective manner with questionnaires (Sterzing et al 2010). Such tests may aid in illustrating our understanding of traction, but the findings are incomplete without knowing if there is an actual traction difference. In addition, past investigations have not answered the question of how much traction is 'excessive'? Or how much traction is adequate? Such knowledge can be obtained via both test methods, subjective and lab tests. Despite the importance of interrelating these two tests, few studies measured actual traction and perceived traction (Cole et al., 2023). In addition, interrelating these two methods may aid in detecting noticeable differences in traction. Past studies of perceived traction have mostly studied manual grip with the hand(s).

Wu et al. (2011) showed that torque sensitivity is intrinsically responsive to the context of the task. Woodruff and Helson, (1967) found that the observed torque threshold increased from 4.4 % to 12.6 %, i.e. three times greater when the reference torque changed from 0.98 to 0.082 Nm (almost 9 times more). Studies related to the just noticeable difference (JND) have been performed on the upper extremities, and hand but far fewer studies have studied the foot, especially among athletes. Therefore, this study aimed to first design a methodological framework for evaluating perceived rotational traction, and next to investigate noticeable differences of rotational traction in a field test and relate those findings to laboratory tests.

4.1.3.2 METHODS

Participants. Seventeen football players (12 males and 5 females), aged 28.1 (SD = 11.5) years old, 71.2 (SD = 5.5) kg, and height 175 (SD = 4.8) cm, participated in the experiment. Their mean foot length was 26.1 (SD = 0.7) cm. Participants had at least 8 years of experience playing club football and in addition, averaged 2 - 4 hours of weekly training last year. They were injury-free for the 3 months before the experiment. This research was carried out in accordance with the ethical standards of the 1964 Helsinki Declaration (World Medical Association Declaration of Helsinki, 2013).

Materials. Three traditional outsoles (full-stud pattern, soft-ground pattern, and H-shape pattern) football boots were provided in the size 42.5 EU (three pairs from each pattern were provided). The football boots only differed in the stud configuration of the outsole whereas the upper layers and insole were identical for all boots. The outsoles configurations (number of studs) of the football boots were varied in the forefoot, midfoot, and rearfoot (see Figure 4.3.1). Stud shape and intra-stud distance are shown in Figure 4.3.2.

Figure 4.3.1.a Three-dimensional printed (outsoles of) boots, side view (upper photo). **b** Outsole configuration with different stud lengths (those shown with the arrow are 10 mm in length, and the rest are 12 mm in length)

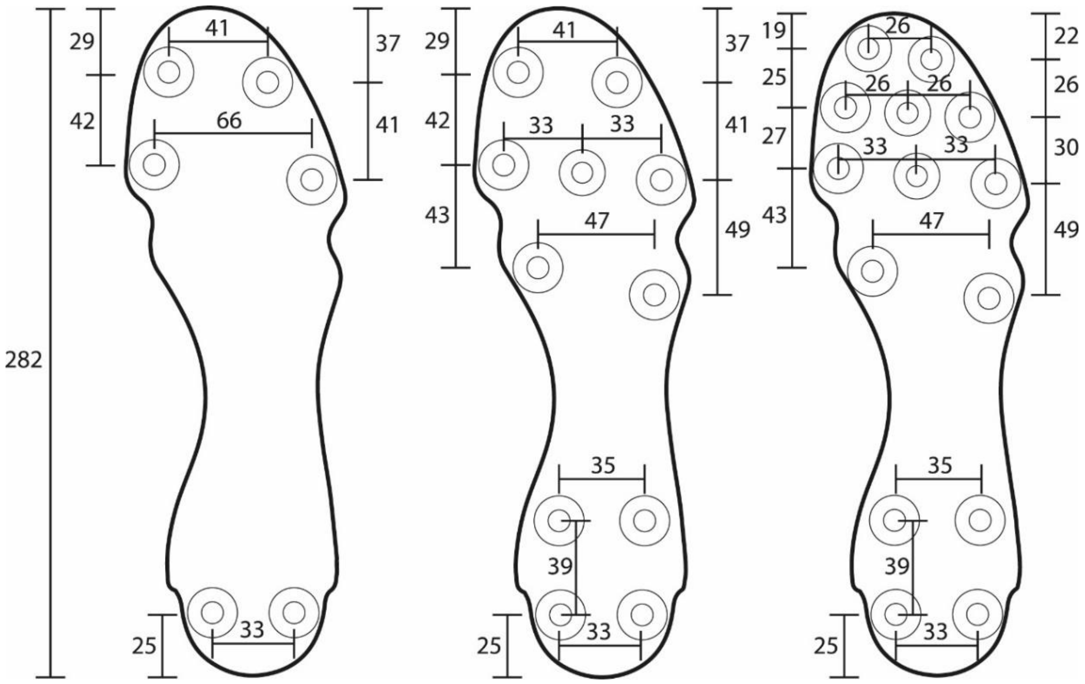


Figure 4.3.2 Three stud patterns and intra-stud distance (dimensions are in mm)

The outsole studs and outer sole were additively manufactured using the *Fused Deposition Modeling*. The outsole consists of the thermoplastic material polyamide 11 (PA11). This has a tensile strength in the XZ and YX directions of 51 MPa and the ZX direction of 47 MPa, an elongation at failure in the XZ and YX directions of 30%, and the ZX direction of 11%. The football boots used in this experiment had identical upper layers and lacing materials. The masses of the shoes are as follows: soft ground pattern =150 g, full stud pattern = 155g, and H-shape pattern = 153 g).

A natural grass football pitch (training and competition pitch in the sports complex at central sport at the Technical University of Munich) was used for the experiment. Surface hardness was measured with a Clegg hammer impactor 2.25 kg (SD Instrumentation Ltd) on three points (starting point, middle point, and random point on the corner). Five trials were conducted for each condition. A total of 15 Clegg hammer tests were obtained before the experiment for each participant. The soil moisture and soil temperature were measured with a soil instrument *(Easemind Technology Ltd)* before each experiment. The soil moisture is expressed by the soil measuring device by three criteria: I, dry, corresponds to a soil moisture of < 10 %, II, normal, to a moisture of 10-20 %, and III, wet, to a soil moisture of > 20 %. The experiment took place within 9 days starting in March 2021 and ending in May 2021.

Protocol of study. Participants were asked to warm up for 10 minutes according to FIFA 11 program (Nuhu et al. 2021). The warm-up program is divided into running exercises and strengthening exercises. All participants used their private shoes during the warm-up sessions and were given identical socks.

Comfort Fit. After the warm-up, in a random and blind order, participants wore football boots (boots A, B, and C) and performed a one-minute test. The test combined different movements such as jumping, cutting, and sprinting. Immediately afterward, participants filled out the first questionnaire related to comfort/fit. In the questionnaire, five locations within the boots were distinguished indicated in Figure 4.3.3. Participants rated the comfort/fit for each plantar location using visual analog scales (*150 mm).*

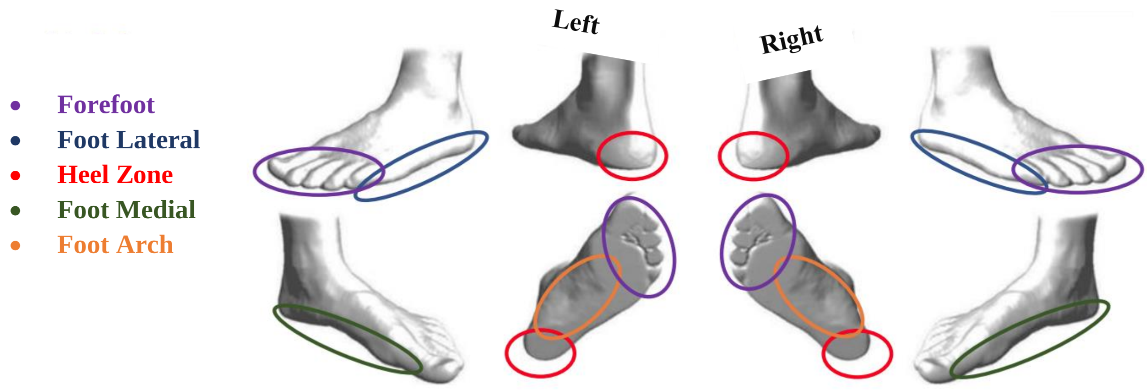


Figure 4.3.3 Foot classification into 6 sections (forefoot, lateral, heel, medial, foot arch, and total foot) to rate *comfort fit* using a 150-mm visual analogue scale.

Perceived Traction. Next, participants evaluated rotational traction (RT). This was defined for participants as: 'the interaction between the football studs and the surface where RT can increase grip. If it is excessive, it can cause foot fixation and if it is poor, it can cause slippage'. To evaluate traction, each player completed an individual evaluation session to obtain their sensory responses for boot traction assessment. Functional tests with and without football were designed (Figure 4.3.4). Participants started by running 12 m and then performed a cutting movement (135°). Immediately after, they ran a slalom course without a football for 7.5m. After reaching the starting point again, they performed a cutting movement (135°) to the right side and then ran a slalom course with a football for 7.5 m. Then, they returned to the starting point again. In the final movement, they performed a cutting movement (randomly either 45° or 135°) to the left or right side. During this section, an experimenter stood at point M and showed a left or right direction (randomly) when the participant was ~1m away from point M. Participants were asked to run at 75 % of their training intensity.

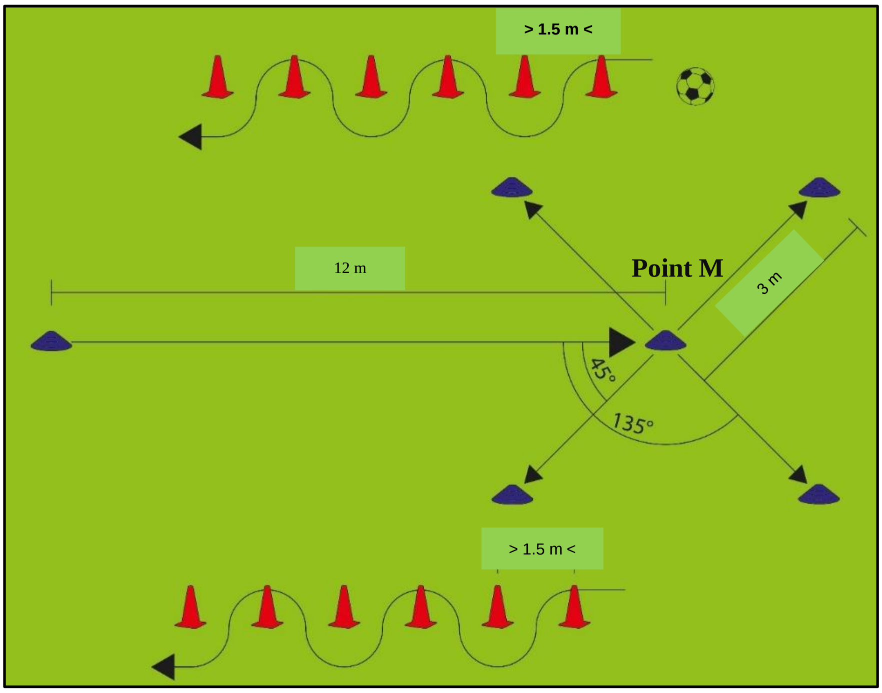


Figure 4.3.4 Schematic of functional traction test. It includes slalom with/without a ball, cutting 45° and cutting 135°.

Pair-wise comparison test. To evaluate perceived traction (rotational) in the field test, a method called pair comparison test (Thurstone et al., 1927: Böhm et al. 2008) was used. Three football boots *S06 = A*, *S11 = B*, and *S14 = C*, and a control football boot which is selected randomly among three football boots (A or B or C) used in 6 pair tests (see Figure 4.3.5). For example, for Participant 1, four shoes A, B, C, and control shoe B (control shoe was chosen randomly). In total, 6 paired tests were performed as follows: (A vs. B), (B vs. C), (C vs. A), (A vs. B), (B vs. B). Using a control boot helped to minimize bias in the experiment.

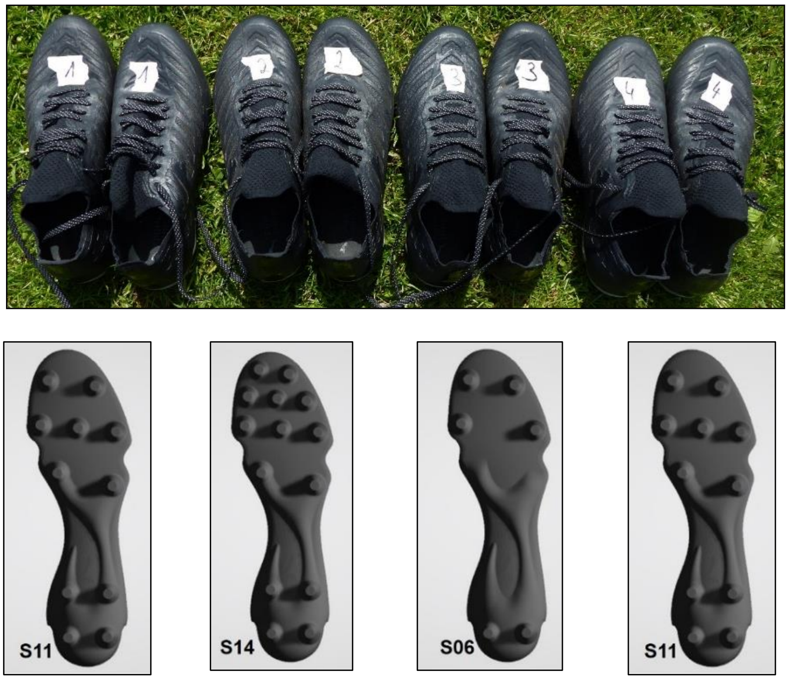


Figure 4.3.5 Image Above Top view of football boots used in the experiment boots (identical upper parts). The image below: outsole configurations used for a participant.

Participants performed functional traction movements for each pair test. This procedure occurred for all available paired tests. At the end of each paired test, participants answered three questions as follows:

Q1. Is there any difference between boots about traction?

Possible Answers Yes or No

Q2. Which boots have 'greater' rotational traction?

Possible Answers: boot 1 or boot 2

Q3. Which boot has 'preferred' rotational traction?

Possible Answers: boot 1 or boot 2

In addition, the exertion level was obtained using the Borg Rating of Perceived Exertion (RPE) scale (6-20), after the second, fourth, and sixth paired tests (Borg, 1998).

Objective test. Rotational traction between the boot and surface was measured using a commercially available portable traction testing device (S2T2, Exeter Research USA). The device consists of a prosthetic footform (size 10.5 US), on which shoes are fitted and positioned at 20˚ of plantar flexion to ensure only the forefoot studs are in contact with the surface (Serensits & McNitt, 2014; Wannop & Stefanyshyn, 2012; Thomson et al., 2019). The foot can be rotated to measure peak rotational traction (Wannop et al., 2013; Wannop et al., 2012). For rotational traction, a vertical load of 580 N (59.1 kg) was applied and the test foot rotated at a speed of ~ 90 °/s via a torque wrench (ETW-PR-100, Checkline, NY, USA) which is located at the top of the machine that was manually driven by a single operator. Wheels allowed us to move the Aspetar Traction Tester to multiple testing locations on the playing surface (figure 4.3.6). Each shoe model was tested at three separate locations on the playing surface (natural grass) for two trials. The surface hardness was measured with a Clegg hammer impactor (2.25 kg) on three spots randomly. Each spot was measured with 5 test trials. Several 15 tests were accomplished before the experiment per subject.

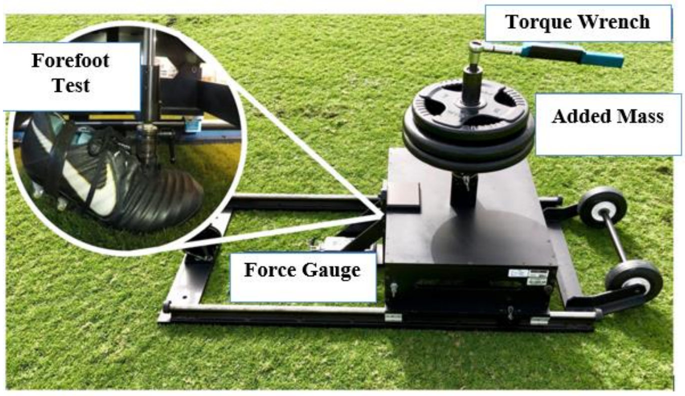


Figure 4.3.6.a Aspetar Traction Tester with their components (The image was reproduced from Thomson et al. 2022).

Figure 4.3.6.b Aspetar Traction Tester in Doha using full-stud pattern boot for test trial in this study.

Peak rotational traction was recorded in newton meters (Nm) by a digital torque wrench sampling at 500 Hz (ETW-PR-100, Checkline, NY, USA) with an accuracy of ±1% of indicated measurement in a range of 10 – 100 N.m. The normal load of 580 N (59.1 kg) allowed for comparison to previous studies using a similar methodology (Thomson et al., 2019; Wannop & Stefanyshyn, 2012; Wannop et al., 2013). The normal load of 580 N was deemed to cause an acceptable amount of damage for grounds-staff to manage on a high-use football surface and is a vertical load used for previous studies in American football (Wannop et al., 2013; Wannop et al., 2012).

Statistical Methods. Several statistical methods were used in this study. To analyze 'comfort fit' and 'Borg scale', a one-way *ANOVA* and *non-parametric Friedman test* respectively were used. To analyze the perception of traction in Questions 1, 2, and 3, the *pair test method*, *Chi-Square test*, and *coin theory* were used. In the end, linear regression analysis was used regarding surface moisture and hardness level.

To evaluate Questions 2 and 3 a pairwise comparison test (PCT) was used in this study (Böhm et al., 2008). The paired comparisons were evaluated using a 3 × 3 matrix. In the first step, the decisions of all subjects were summarized. If a subject chose a boot (highest traction or preferred traction), it was given a 1. If a subject had thus decided against a boot, with 0. These values were recorded in the first 3 × 3 matrix. In the next step, the values were normalized to the number of subjects and presented in a second 3 × 3 matrix. This was followed in the third step by a transformation of the normalized rank data into ratio-scaled preference values. This was done based on the assumption of normal distribution, which is justified by the law of comparative assessment (Thurstone et al., 1927; Böhm et al. 2008).

Coin theory a binomial distribution based on the PCT, was used to examine rotational traction for significant differences. When examining whether there is a significant difference, the probability of each stud configuration is calculated using the binomial coefficient. This is based on the fact that each result had a probability of 0.5.

$$P(X = k) = \binom{n}{k} p^k (1-p)^{n-k}, k \in \mathbb{N}$$

The evaluation was carried out with the software MATHLAB R2020a. The binomial test myBinomTest(s,n,p,Sided) from the MATLAB Central File Exchange was used (Nelson, 2022).

4.1.3.3 RESULTS

Environmental parameters were evaluated during 9 days of experiments and shown in Table 4.3.1 for 12 different days among 17 players.

Table 4.3.1 Environmental parameters within days of the experiment

Surface hardness [Gm]	Surface temperature [°C]	Ambient temperature [°C]	Ambient humidity [%]	Surface moisture	
				Dry <10%	Normal 10-20%
67 (SD = ±7.5)	10.1 (SD = ±5.7)	10.2 (SD = ±4.8)	57.1 (SD = ±13.9)	n =7	n =10

Surface hardness measurements were conducted on various days, as illustrated in Figure 4.3.7.

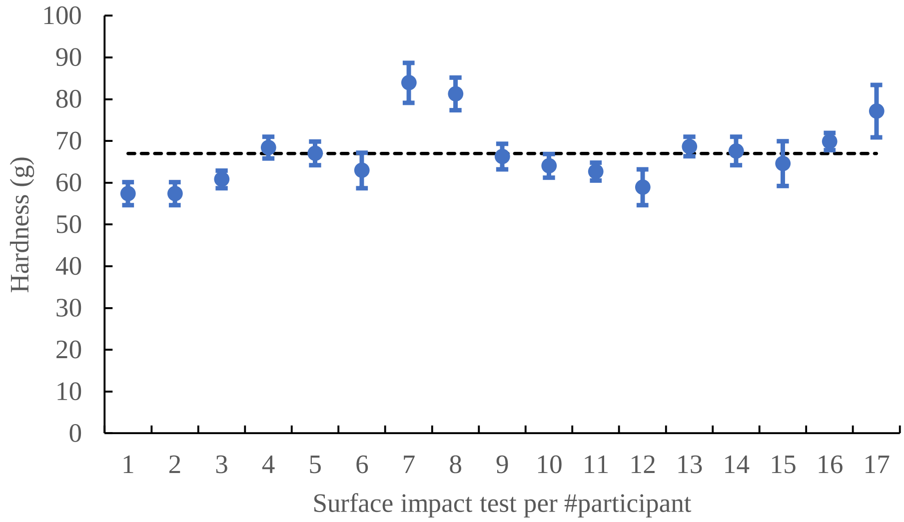


Figure 4.3.7 Mean surface hardness before the test of each participant.

Comfort Fit. Data of comfort/fit (Figure 4.3.8) among three boots is analyzed with one-way ANOVA. Our findings showed no significant difference between the three boots in overall plantar comfort/fit, F (2, 34) = 2.4, p-value=0.103. In addition, the comfort/fit of the four locations was compared with one-way ANOVA with Bonferroni post hoc correction. There was a significant difference between foot locations as follows: F (4, 204) = 4.22, p-value = 0.003. However, post hoc tests of Bonferroni correction showed that there was only a significant difference between two locations, the heel and medial forefoot (p = 0.003).

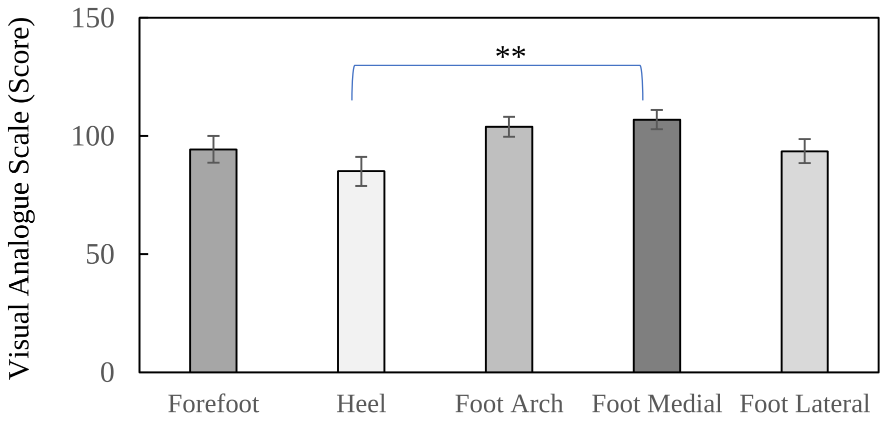


Figure 4.3.8 Comfort fit of Five sections for three boots using Visual Analogue Scale-150mm (SEM = +/-2)

Borg Scale. The Borg rating of perceived exertion (RPE) scale (6-20) results were analyzed with a non-parametric Friedman test. The findings showed that χ^2 (2) = 19.18, p = 0.001. Post hoc tests determine that there is a significant difference between *Part One vs. Part Two* (p-value < 0.001) and *Part One vs. Part Three*: p-value = 0.004. The mean rating was 12.05 (SD = 2.6). The medians of three parts (parts 1, 2, and 3) were 11, 13, and 13 according to the Borg scale referred to as ‘light =11’ and ‘somewhat hard = 13’.

Question One. To understand whether players could perceive traction while the outsole of the boots was varied, data from the binary questionnaire (Q1) was evaluated. A Chi-square test was used in our experiment for paired tests S06, S11, and S14; a total of 51 paired tests. Traction perception differed significantly between the three boots in all 51 pair tests: χ^2 (1) = 16.491, p-value = 0.0001. The binary data is showcased in Table 4.3.2 and visually represented in Figure 4.3.9.

Table 4.3.2 Traction perception of all pair tests for 17 participants

Traction perception in the pair test	Frequency	Percent	Test Percent
No	11	21.57	50
Yes	40	78.43	50

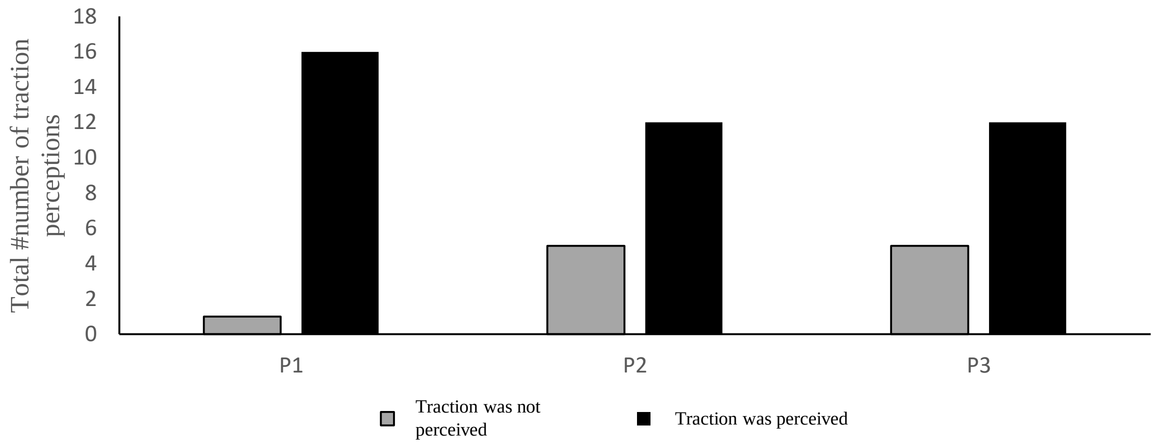


Figure 4.3.9. The responses of binary questionnaire-traction perception-per-pair test (P1 = S06 vs S11, P2 = S06 vs S14, and P3 = S11 vs S14)

Question 2. The initial dataset comprised the results of 17 participants across three paired tests, totaling 51 samples. However, participants unable to discern traction differences in the paired test, Question 1 (Q1) were subsequently excluded from the data analysis in this section. Therefore, in the end, 40 samples (out of 51 data samples) were analyzed in this section (See Table 4.3.3). The results of paired tests showed that there was no significant perceived difference between the three boots: Chi-square = χ^2 (2) = 0.43, p = 0.8

Table 4.3.3 Frequency of data sample among three boots

	Frequency	Percent	Test percent
S06	13	32.5	33
S11	15	37.5	33
S14	12	30	34

The results of Question 2 were also analyzed by using the Pair test method and in addition the coin theory (See Table 4.3.4 - 6 and Figure 4.3.10). According to the scaled values, the boot B =S11 showed the highest perceived traction among the boots. However, the coin theory showed no significant differences in the pair comparisons between the football boots. The comparison of *S06 vs. S11* showed p = 1.0, of *S06 vs. S11* showed p = 0.77 and of *S11 vs. S14* was also p = 0.77.

Table 4.3.4 The absolute value of the pair test

HT	S06	S11	S14	Sum
S06		8	5	13
S11	8		7	15
S14	7	5		12

HT denotes high traction.

Table 4.3.5 Relative value of pair tests

HT	S06	S11	S14
S06		0.50	0.42
S11	0.50		0.58
S14	0.58	0.42	

Table 4.3.6 Replacement of the normalized PCT by corresponding Z-values

HT	S06	S11	S14	Mean
S06		0.00	-0.21	-0.07
S11	0.00		0.21	0.07
S14	0.21	-0.21		0.00

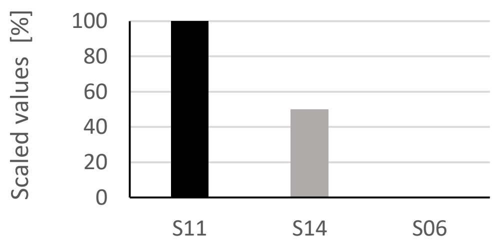


Figure 4.3.10 Scaled value of the paired test (High traction).

Question 3. The initial dataset comprised the results of 17 participants across three paired tests, totaling 51 samples. However, participants unable to discern traction differences in the paired test, Question 1 (Q1) were subsequently excluded from the data analysis in this section (See Table 4.3.7). In total, 40 samples were considered in this section. The results of pair tests show that Chi-square χ^2 (2) = 2.73, p = 0.25.

Table 4.3.7 Frequency of data sample among three boots

	Frequency	percent	Test percent
Shoe A	10	25	33
Shoe B	18	45	33
Shoe C	12	30	34

The results of question 3 were also analyzed by using the Pair test method and in addition, the coin theory See Table 4.3.8-10 and Figure 4.3.11). According to the scaled values, the boot B =S11 showed the best preferred perceived traction among the boots. However, the coin theory showed no significant differences in the paired comparisons between the different football boots. The comparison of *S06 vs. S11* shows p = 0.45, of *S06 vs. S11* shows p = 0.39, and *S11 vs. S14* is also p = 0.39.

Table 4.3.8 Sum of the PCT matrix of all 17 participants.

PT*	S06	S11	S14	Sum
S06		6	4	10
S11	10		8	18
S14	8	4		12

PT denotes preferred traction.

Table 4.3.9 Relative value of paired tests.

PT	S06	S11	S14
S06		0.38	0.33
S11	0.63		0.67
S14	0.67	0.33	

Table 4.3.10 Replacement of the normalized PCT by corresponding Z-values.

PT	S06	S11	S14	Mean
S06		-0.32	-0.43	-0.25
S11	0.32		0.43	0.25
S14	0.43	-0.43		0.00

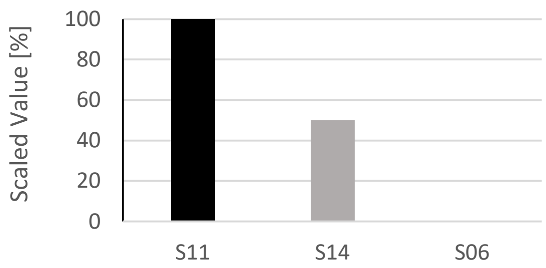


Figure 4.3.11 Scaled value of the paired test (PT).

Regression Model. The regression analysis was defined for Q2 and Q3 in regards to two variables ‘moisture’ and hardness (Table 4.3.11-12). Among these two questions, only question 1 (Q1) related to ‘high’ traction is determined to be significant by regression analysis as follows: χ^2 (2) = 2.73, p = 0.25.

Table 4.3.11 A goodness of fit test for regression models.

	Df	Chi-square	Pr > Chi-square
Moisture	2	4.98	0.08
hardness	2	6.90	0.03

Table 4.3.12 The regression coefficient for linear regression (Explanatory variables).

	Boots	Point estimate	95% Confidence interval	
			Low	high
Moisture	S06	1.2	0.16	9.60
	S11	0.03	0.001	0.86
Hardness	S06	0.96	0.84	1.09
	S11	1.21	1.02	1.43

S14 is considered a control boot

When surface hardness increases by 1 unit, the perception of 'high' traction of Boot A in comparison to Boot C decreased by 4 % (odds ratio = 0.96). Similarly, when surface hardness increased by 1 unit, the perception of high traction boot B in comparison to boot C increased by 21 % percent (odds ratio = 1.21). The formulation of the regression model is as follows:

$$\text{Prob (Boot = A) / Prob (Boot = C)} = \exp(2.5 + 0.2 \times \text{Moisture} - 0.03 \times \text{Hardness})$$

$$\text{Prob (Boot = B) / Prob (Boot = C)} = \exp(-11.6 - 3.3 \times \text{Moisture} + 0.2 \times \text{Hardness})$$

Moisture is coded 0 and 1. Hardness denotes its value. Additionally, in the aforementioned equation, S06 corresponds to boot A, S11 corresponds to boot B, and S14 corresponds to boot C.

Data for objective traction test. The mean of rotational traction for the three boots S06 = boot A, S11 = boot B, and S14 = boot C are as follows 52.5, 51.8, and 51.3 Nm (see Figure 4.3.12). A non-parametric Friedman test determined that there were no significant differences between three football boots regarding rotational traction (p-value > 0.05).

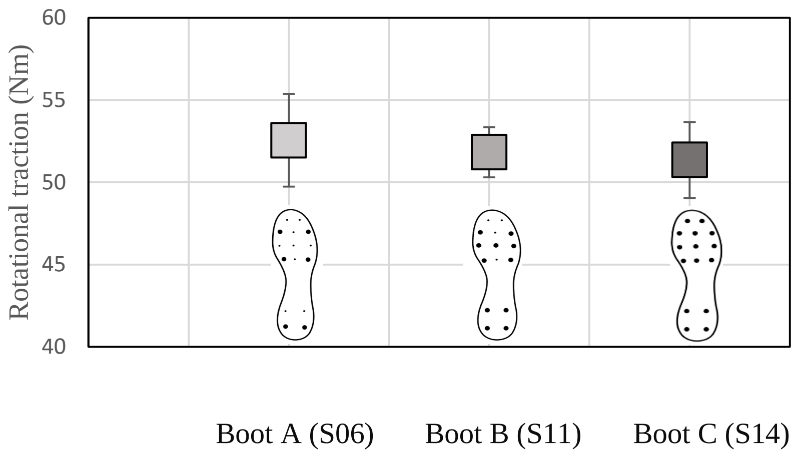


Figure 4.3.12 Mean and standard deviation of three boots' rotational traction using Aspetar Traction Tester Raw data

4.1.3.4 DISCUSSION

In this perturbation study, sensory response to rotational traction was evaluated with different outsole configurations during the rapid change of running direction with and without a football. The findings of our results determined that players were able to distinguish changes in the rotational traction when comparing three different football boots. Our early findings showed that participants detected changes in rotational traction when the boot outsole configuration changed. This has been shown in 78.4 % of all paired tests successfully. This finding is in line with those of Cole et al. (2022) and Müller et al. (2010) who showed that players could distinguish traction in regards to changes in surfaces and changes in boots respectively. In detail, our results determined that players were able to distinguish a difference of 0.9 % to 2.6 % in the rotational traction of three boots, full stud pattern (S14), soft-ground boot (S06), and H-shape (S11). However, in the study of Cole et al. (2023), the minimum perceptible change of rotational traction in laboratory tests reached 6 % and in Müller et al. (2010) results reached ~ 5 %.

In the second and third questions, participants could not perceive which boot was perceived as the 'highest' traction, and which boot they defined as the 'preferred' traction, respectively (although S11 shows the greatest subjective rate in both questions). One may ask how it would be possible that they detect minimum traction difference among boots but they cannot perceive the e.g. 'highest' traction among the boots. The answer to such a question is embedded in the complex nature of human perception and adaptation in the test. Due to variations of surface conditions (e.g. surface hardness –see Figure 4.3.7), every participant was faced with a unique boot-surface condition that was distinct (separated) from the other participants. Since perception and adaptation are the product of proprioception (ligament, muscle, skin, joint), we speculate that adaptation to a new boot and surface conditions affects the participant's paired test responses. A player who is using different football boots on a surface with a hardness of 60g requires different adaptations of the joint motions such as knee and ankle (Dixon, Collop, and Batt (2000)) and also different adaptations of muscle activity to reach 'high' / 'preferred' traction in comparison of a player who is using those three football boots on a surface with the hardness of 80g. Such a different adaptation magnitude between players results in different senses of effort (Cai et al. 2023) which can influence the perception of rotational torque.

However, it is still unclear whether such adaptation is towards the perception of 'high'/ 'preferred' traction or is an attempt to moderate the traction severity (to prevent injuries). It should be noted that the interaction between boot and surface is complex, and several variables contribute to the development of traction (Karamanidis et al. 2006). A better understanding of the effect of surface on traction perception emerges from the regression analysis (shown in the Results section) which determined that players' responses to higher traction are affected by increasing surface hardness. An increase of one unit (1g-acceleration) of surface hardness results in an increase of 21% in the related attribution = 'high traction perception' of the H-shape stud (boot B) when compared with the full stud pattern (boot C). On the other hand, an increase of one unit of surface hardness (1g) results in a reduction of 4% in the related attribution = 'high traction perception' of the Soft-ground boot (boot A) when compared with the full stud pattern (boot C).

Two points should be considered before the interpretation latter findings: 1. comfort/fit in our study was highest on the medial side of the boot, and 2. the medial forefoot, hallux, and medial heel (Figure 4.3.13) exert greater forces on the ground and a larger penetration of the cleat or stud under this part of the foot will occur, (Hennig et al. 2010). Thus, the cleats under the medial side (hallux, medial forefoot, and medial heel) are the most important outsole structures for providing the desired traction for better performance (See Figure 4.3.13). Therefore boot B (S11) showed higher traction due to the stud number and stud position, which suits the most important effective stud locations based on pressure mapping, including Hallux, medial forefoot, and medial heel). This aids in generating efficient penetration (Graf et al., 2012) and, accordingly, efficient peak pressure.

Boot S11 showed higher traction due to the stud position, stud number, and accordingly better penetration which suits the most important effective stud locations (Hallux, medial forefoot, and medial heel) to generate peak pressure. The relatively low traction perception of the boot (S06) might be attributed to its poor penetration (Graf et al., 2012).

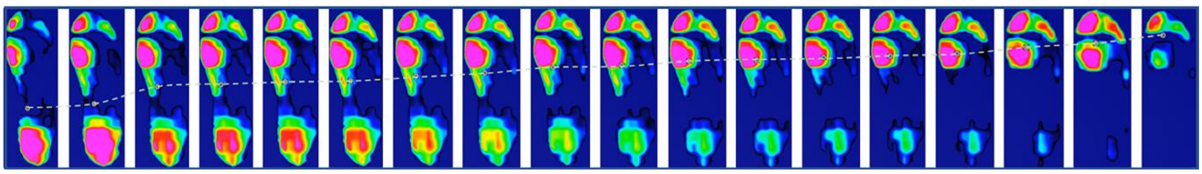


Figure 4.3.13 Pressure mapping in the cutting movement was obtained randomly from a participant in the experiment- the center of pressure changes from the bottom of the midfoot to the medial forefoot. Three zones exert the greatest pressure: heel, hallux, and medial forefoot.

The findings of this study should be taken in cautiously due to some limitations:

- The surface hardness used in objective measurement is different from the surface used in the subjective measurement. The mean hardnesses of the two surfaces in subjective and objective tests were 67 g and 72 g respectively. This should be taken into account that such a difference can affect penetration depth and absolute traction, however relative difference between the boot's traction may stay the same.

- The subjective measurement occurs for several weeks due to SARS Covid 19 pandemic condition. Performing the test during different phases of the season may help us to classify the hardness level. However, different surface hardness affects the homogeneity of results in subjective tests. As it has been mentioned before, traction preference could vary because of surface condition not only because of different outsole configurations.

- The traction tester used in this study only evaluates traction for the forefoot contact with the load 580 N. This may affect on noticeable difference of torque which has been defined early in this study. Future studies should replicate traction evaluation pressure concerning other cutting stance phases of pressure mapping.

- Both surfaces used in the subjective field test and objective field test were cool season grass, Rye grass species (Lolium perenne 'Perennial Rye'). However, in the objective test (Using Aspetar Traction Tester), Rye grass was overseeded on a warm season grass. This may change the cohesion and compression of the soil while testing. This might be another barrier to comparing the absolute results of two findings in subjective and objective tests.

4.1.3.5 CONCLUSION

Our findings determined that players could perceive a difference between the rotational traction of three boots (S06, S11, and S14) in the cutting movement. In addition, when the surface becomes harder by one unit (1g-Max), the boot which has a higher number of studs in medial sections (Hallux, medial forefoot, and medial heel) showed higher perceived traction. Future studies of traction comparison should break down the investigation of torque perception into two parts: first, assessing the ability to sense torque (just noticeable difference) and, second, examining the quality and quantity of understanding of the magnitude, direction, and rate of change of rotational forces under similar test conditions.

4.1.4 SECTION 4

Artificial Neural Network and Outsole Configuration

This section is based on the paper published in the Journal of Sports Engineering on October 2023: Keshvari, B., Lehoang, L. & Senner, V. Investigating the effect of outsole configurations on rotational and translational traction using a mechanical prosthetic foot. Sports Eng 26, 43 (2023).

4.1.4.1 BACKGROUND

Traction is the product of shear stresses resulting from the interaction between studded boots and the playing surface Valiant GA (1990) does not obey classical laws of friction. This means that any values measured are a combination of adhesion, cohesion, and material hysteresis and therefore show a non-causal behavior, which in turn leads to a distribution of measured values (ASTM Committee F08, 2012). Traction was consistently rated as a key consideration in the selection of sports footwear after comfort and stability in 1998, and second only to comfort in 2006 (Hennig, 2006; Sterzing et al., 2007). It was also reported that insufficient traction between studs and playing surfaces can affect performance in rapid change of direction, such as cutting or turning, or accelerating (decelerating) in sprinting (or kicking) (Pedroza et al., 2010; Dowling et al., 2010; Schrier et al., 2014; Sterzing et al., 2009). Accordingly, insufficient traction is linked to the risk of lower extremity injuries (Lambson et al., 1996; Olsen et al., 2003; Wannop et al., 2013). Insufficient traction can be defined as excessive traction which causes foot fixation, or as poor traction (low amount) which causes slippage. However, the safe zone of traction (between performance and injuries) has been poorly investigated.

Traction can be subdivided into rotational and translational traction. In the words of Thomson et al. (2022), translational traction "… is the horizontal force required to overcome resistance between the boot outsole (studs/ cleats) and playing surface in a straight line". They also define that "… rotational traction is the rotational force required to release the studs through the playing surface in a rotational manner". In their study, these two types of traction were independently assessed. An increase in rotational traction led to a significant increase in transverse and frontal plane joint loading at the ankle and knee joints, while increases in translational traction led to increases in frontal plane joint loading at the ankle and knee joints (Wannop et al., 2016).

Since the early 1970s, numerous mechanical test devices have been developed for evaluating the traction of studded boots (Wannop et al., 2016; Bonstingl et al., 1975; Livesay et al., 2006; Thomson et al., 2019). However, there are limitations to these devices. For instance, to evaluate rotational or translational traction, it is necessary to map the pressure of specific football movements and replicate the desirable stance phase (initial, mid, or final) of those movements under laboratory conditions using a mechanical testing device If the pressure mapping in a laboratory test applying a mechanical prosthetic foot was not according to those of players' performing movements, the results could be unrealistic. However, a sub-requirement is to have a more realistic foot model to replicate the movement via the test device.

Other than a study by Grund et al. (2010), pressure mapping from field tests (in football or other sports) has not been applied to mechanical tests in the laboratory. However, such pressure mapping is an important factor in calculating traction in a desired stance phase of specific football movements. It would therefore make sense to adjust the test device to better mimic the *in vivo* pressure mapping. Also, except for the mechanical prosthetic foot mentioned in the study by Grund et al. (2010), as far as we are aware, no measurements considering a varus/valgus position of the shaft relative to the foot have been reported in previous field tests, even though this is an important contributor to Anterior Cruciate Ligament (ACL) injuries. This current study addressed this by reconfiguring the available apparatus to measure traction by simulating pressure distribution using a realistic foot model.

During field tests in football, Smith et al. (2004) and Blackburn et al. (2003) used the implemented force plate technique to measure ground reaction forces. They determined that vertical ground reaction force regularly exceeded 2.5 BW (> 1500 N) although most of the abovementioned studies did not evaluate traction with preload exceeding 1500 N. This is likely due to mechanical problems that might occur while testing at higher preloads, which casts doubt on the validity of these studies. To predict translational and rotational traction in higher preloads, Kirk et al. (2006) suggest using an artificial neural network (ANN) to evaluate the interactive effects of boot and surface and the athlete who likely optimize their traction forces. Further, traction forces do not in all probability follow a linear relationship with applied load. In summary, there is a gap in the literature regarding the translation of traction in field tests to laboratory tests, especially under realistic pre-loads. Therefore, this study aims to establish a methodological framework within which to predict rotational and translational traction under a realistic preload. This would, in turn, facilitate the identification of a safe zone of traction to aid in the design of football boots.

4.1.4.2 METHODS

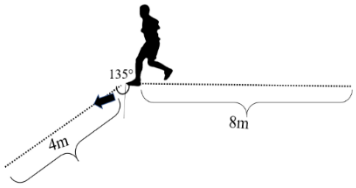
Phase 1. Field Test. Seventeen male participants were recruited from football clubs of Regionalliga Bayern (the fourth tier of the German football league system). Most of these clubs were close to Munich (Germany). The participants had at least 5 years of experience playing football. They were excluded if they had had a lower extremity musculoskeletal injury within the past six months or a lower extremity reconstructive operation within the past three years. The mean (and standard deviation) age, height, and mass of the participants were: 26 (±6) years old, 181(±7) cm, and 79 (±11) kg. Participants gave written, informed consent before the experiment. The consent form declared the confidentiality of the objectives, risks of the study, and protection of personal data through appropriate procedures for anonymization. In addition, the consent form assured participants they were free to withdraw from the research at any time without giving a reason and without penalty for not taking part. This research was conducted according to the ethical standards of the World Medical Association Declaration of Helsinki (2013).

Three pairs of Nike Hypervenom Phinish FG (size UK 9) soft-ground football boots were used in this experiment. All the original studs of the boots were flattened and smoothed with a belt grinder (see Figures 4.4.1a and 4.4.1b). To ensure appropriate grip, the football boot outsoles were covered with a layer of sandpaper, each of different thicknesses (coarseness: ISO 24 and ISO 50). The football surface used in this study (pilot test, field test, and laboratory test) was a 2-star AT (sand/rubber infill 100% polypropylene) used for 4 years. The surface construction consisted of 30 mm of an elastic base layer (DIN V18035/7). It was covered by fiber carpet with a pile length of 10 mm filled with sand (18 kg/m^2) and rubber particles (0.5-1.7 granulation diameter, 5 kg/m^2). The total length of the pile and carpet was 50 mm. Before the main field test, a pilot test was accomplished.

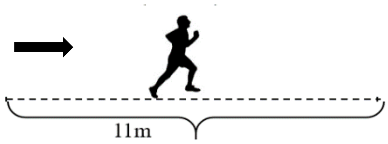
A pilot study was accomplished at the beginning of the experiment. Its purpose is to identify an appropriate sandpaper grip size for the main field test. Three participants were asked to accomplish two types of cutting movements (turning and cutting 45°) and sprinting. At the end of the test, they were asked to compare the boots based on their qualitative perception of traction on artificial turf (AT) and to report their perception of traction verbally. All three stated that the football boots covered with coarse ISO 24 - paper sand (Grit size = 80) provided sufficient traction to perform the required movements with appropriate grip and low risk of slippage

Field test. Participants were first asked to warm up for 15 minutes according to the FIFA 11 program (Nuhu et al., 2021). Then they were asked to perform four movements: cutting 135°, sprinting, turning, and penalty kick (see Figure 4.4.1c-f). For each movement and condition, participants performed three trials, the first and second trials were intended to allow players to become familiar with the boots and surfaces and adapt to an insole pressure sensor. The participants took up to two minutes' rest after each set of four movements. The pressure mapping data was only recorded in the third trial for use in the laboratory test. The purpose of using an insole pressure sensor was to acquire the peak pressure of four specific movements and then obtain the pressure map for those peak pressures. The Opengo insole (Moticon GmbH, Munich Germany) was used in this experiment and it comprised 13 capacitive pressure sensors. Over half of the insole area (52%) was covered by pressure sensors. The measurement range was specified from 0 to 40 $N \cdot cm^{-2}$ and also was sampled at 50 Hz. To differentiate pressure within the plantar area, 13 insole sensors were divided into three sections (forefoot, midfoot, and rearfoot) and nine zones (Figure 4.4.1g). In the cutting movement (135°), participants were asked to perform an 'open' technique, which involves the athlete using the foot on the opposite side to the direction they want to turn. In this movement, the peak pressure on the foot which changed the direction was collected (see Figure 4.4.1c). In sprinting, the peak stance-phase pressure was recorded for a distance of 2 m in the middle of the path (random feet in this area) (see Figure 4.4.1d), and in turn, the data from the main foot which changed the direction was recorded (see Figure 4.4.1e). For the penalty kick, the data from the supporting foot was collected as the ball was kicked towards the goal frame (see Figure 4.4.1f). In all movements, the measured foot was the preferred foot (Wong et al., (2007). A Clegg hammer impactor 2.25 kg (SD Instrumentation Ltd) was used in the field test to evaluate the hardness of the surface. Three trials were performed on four designated strips of the playing surface on a football pitch (for a total of 12 trials). The mean hardness of the experimental surface in the field test aided in subsequently replicating the same playing surface in the laboratory trials.

1a

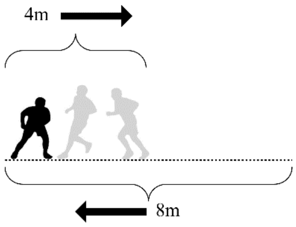


1c

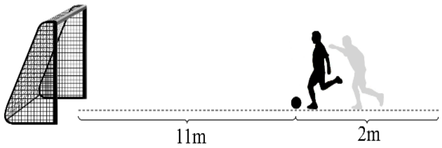


1d

1b

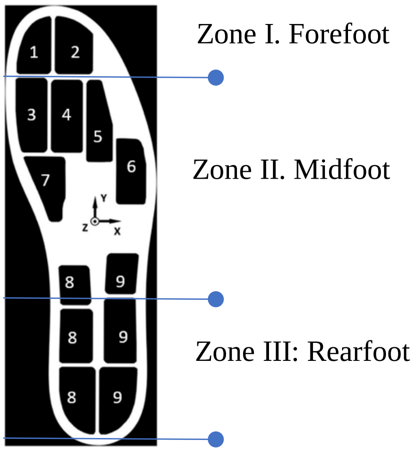


1e



1f

Zone number#	*Zone name*	*Zone number#*	*Zone name*
1	Hallux	5	Central forefoot (ii)
2	Small toes	6	Lateral midfoot
3	Medial forefoot	7	Medial forefoot
4	Central forefoot (i)	8	Medial rearfoot
5	Central forefoot (ii)	9	Lateral rearfoot



1g

Figure 4.4.1.a An original outsole of the football boot. **b** the football boot with flattened outsole used in field test., **c.** cutting movement 135°. **d** sprinting movement. **e** turning movement.**f** penalty kick. **g** different sections and zones of plantar.

Phase 2. Football boots. Five outsole configurations (see Figure 4.4.2) were randomly chosen among 15 designs in our outsole-design data bank. In addition, two outsoles E and H were chosen due to their traditional usage of them in football competition and training:

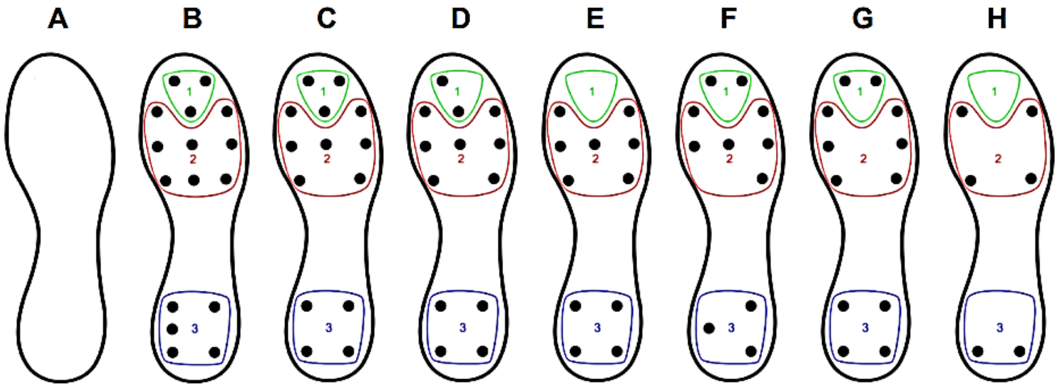


Figure 4.4.2 Outsole configurations of eight outsoles used in the Laboratory. *Outsole 'A' has sandpaper without studs.*

The football boot data bank was classified into three parts: I. variety of outsole configurations (design of stud distribution), II. stud length, and III. surface area per stud. The data bank consisted of the photographed boots and the measurements. For this experiment, we focused only on the outsole configurations and stud lengths (See Figure 4.4.3). The stud lengths were derived from more than forty boots studs over three plantar zones (forefoot, midfoot, and rearfoot) among all football boots in our database. The surface area was adopted from the football boot brand Adidas, model: Copa Mundial FG.

Figure. 4.4.3 Photographs of four outsoles of football boots measured their stud heights used in the base (all dimensions in mm).

To do the laboratory test with the mechanical prosthetic foot (MPF), the studs of a selected football boot, the Adidas ACE 17.3 (size 42.5), were flattened and smoothed with a belt grinder. Then, 16 possible stud positions (see Figure 4.4.4) were created with boreholes on the outsole and equipped with threaded bearings. Three types of conical-shaped studs of varying lengths and areas were created for the three plantar zones (forefoot, midfoot, and rearfoot) on the outsole (see Figure 4.4.4b). These studs were Computerized Numerical Control (CNC) machined (Fräsmaschine Deckel, FP 4A) in Polyoxymethylene (POM). They were used according to the seven outsole designs shown in Figure 4.4.2. The outsole material was Polyurethane.

a.

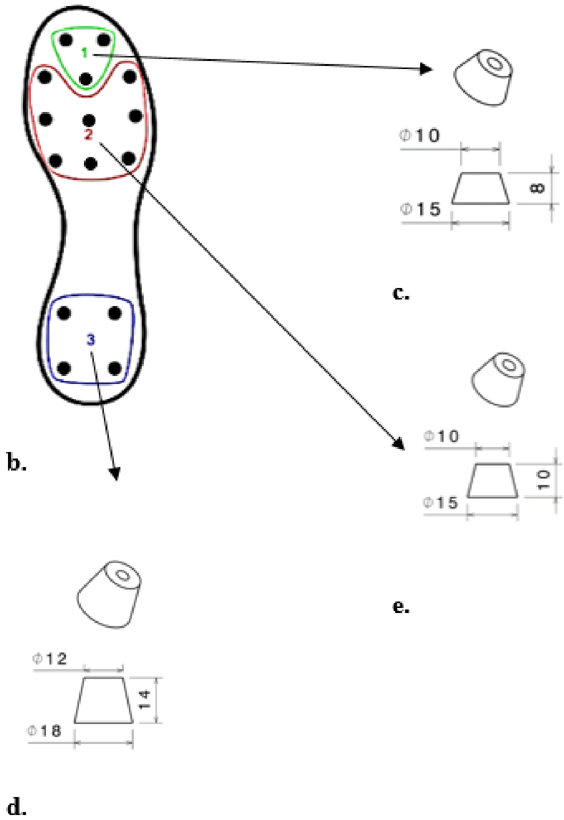


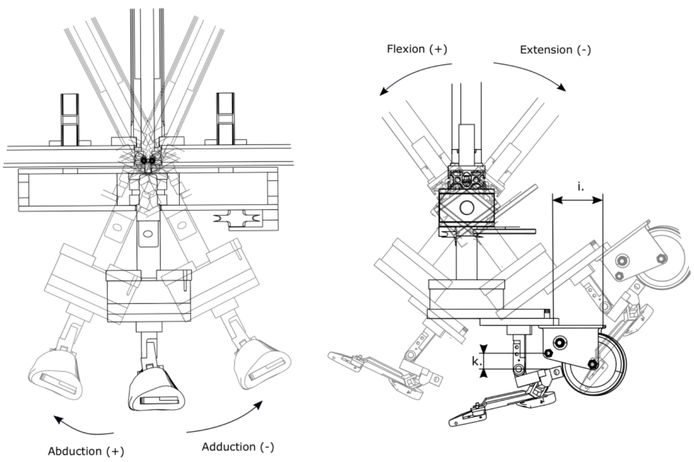
Figure 4.4.4.a Photograph of three types of studs and the boot used in the laboratory test. **b** Studs used in the out sole in three sections 1= forefoot, 2 = midfoot 3 = rearfoot. **c** Stud geometry in forefoot in mm. d. Stud geometry of midfoot in mm. **e** Stud geometry in rearfoot in mm. (all dimensions are in millimeters)

Phase 3. Laboratory test: Mechanical prosthetic foot (MPF). To replicate field conditions in the laboratory setting, the following three steps were undertaken. First, we prepared football boots that could be configured to different and desired stud patterns. Second, we provided an experimental surface that closely matched the impact peak accelerations from the Clegg Hammer test of the surface in our field test. Third, we adjusted the MPF position relative to the surface to replicate the movements measured in the field test pressure mapping.

The MPF was designed as a pneumatic machine, which can pseudo-statically simulate motions leading to non-contact ACL injuries. The main components of the machine were the frame, the artificial leg, and the pneumatic control parts. The frame had two flexion frames mounted on it, which allowed for setting up the flexion/extension and adduction/abduction angles of the artificial leg (Figures 4.4.5 and 4.4.6). The artificial leg comprised a vertical cylinder to statically apply loads along the shaft, a pneumatic muscle (Muscle SIM, Festo AG & Co. KG, Esslingen, Deutschland, F_{Muscle} = Force responsible for rotational deflection) to induce rotation, and a 6-component load cell to record all six occurring force components (ATI-IA Omega160, Apex, NC, USA). The load cell was connected to an ATI Transducer Power Supply box (9105-PS-1). This box was then connected to an NI 9205 card (National instrument) which fits into an NI Crio analog-to-digital converter. The Crio was then connected via an Ethernet cable to the computer.

A custom foot model was used in this study based on (Grund et al., 2007). It consisted of a metal skeleton that was molded in a foot-shaped silicon cast with length and width of 26 cm and 10.5 cm respectively (see Figure 4.4.5). To ensure the results reflected field test conditions as closely as possible, the machine also had a sophisticated foot and ankle model to simulate realistic plantar pressure distribution. To allow for linear traction measurements, a transversal table (Figure 4.4.5g) was added, which could move the ground relative to the mechanical foot. The design was based on the linear table construction of the ski boot testing device concept presented by Knye et al. (2016).

For our laboratory study, we utilized surfaces with similar impact peak accelerations from the Clegg Hammer as those recorded in the field test (155 g-Max). The layered surface sample and the supporting steel block used in the laboratory experiment were 50 × 40 cm and 40 × 40 cm, respectively. The AT top layer was changed after 20 trials of each boot (e.g. a football boot at preload of 400 N in four movements, with each movement composed of 5 trials = 4 × 5 = 20). The experimental surface was fixed to the supporting block with two clamps, one at the upper part and another at the lower part (Figure 4.4.5i). The surface sample could either be moved forward or rearwards relative to the foot. For translational traction testing, the surface was vertically raised until it came into contact with the forefoot of the shoe, and the abovementioned normal loads were reached. After normal load was reached, the platform moved 200 mm anteriorly to the shoe at a speed of 0.2 m. s $^{-1}$ while the horizontal (F_y and F_x) and vertical (F_z) forces were measured by the load cell during movement.



a b

Figure. 4.4.5 Two degrees of freedom for adjusting movements. **a** Abduction /adduction. **b** Flexion Extension- This adjustment facilitated appropriate pressure distribution per each movement (cutting, kicking, sprinting, and turning).

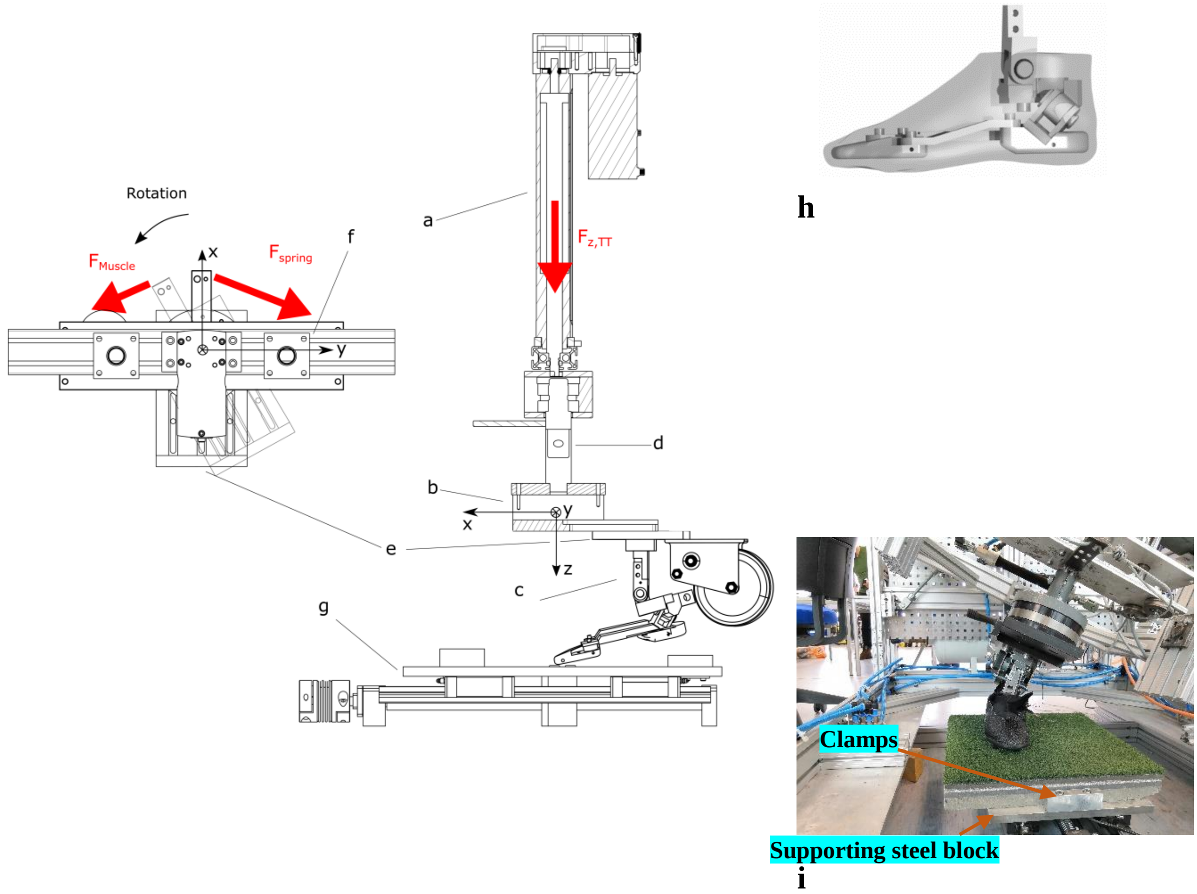


Figure 4.4.6 Diagram showing the MPF in detail. **a** an Electromechanical Cylinder. **b** Load Cell. **c** Traction Foot Model. **d** Rotational axis. **e** Rotational Axis Displacement Mosul. **f** Parallel Guide. **g** transverse table. Forces: F_{Shaft} - Load to recreate Ground Reaction Force, F_{Muscle} - Force responsible for rotational deflection, F_{Spring} - rotational restoring force. **h** foot model with silicon cover. **i** MPF while replicating 135° cutting movement after adjustment with appropriate pressure distribution.

A Pedar Mobile system (Novel GmbH, Munich) was used to collect plantar pressure distribution in the laboratory test. The system consisted of two flexible insoles (each containing 99 sensors in a matrix design) and a portable data logger for data storage. The sampling frequency was 50 Hz. The insole pressure sensor was inserted in the baseline boot to replicate the plantar pressure distribution for the cutting movement with a vertical preload of 400N. To attain this plantar mapping, the MPF was adjusted based on abduction/ adduction and flexion/extension, as shown previously in Figure. 4.4.5. After replicating the pressure mapping of the field test in this laboratory condition, the baseline boot was replaced with other boot outsole configurations. Next, translational and rotational tractions were measured for the other seven boots. Each condition was tested in 5 trials. A similar procedure was followed for preloads of 600 N and 800 N. In all trials, the center of pressure was kept constant per movement while applying preloads of 400 N, 600 N, and 800 N (the location is shown in Figure 4.4.8a-d).

The foot model used in our study was still different from an actual foot. In our attempt to reach a preload of 1500 N, it was observed that the traditional insole sensors (Moticon and Pedar) were partially damaged. The reason for this damage and inability to reach higher loads than 1500 N using the insole sensor is as follows: MPF is a pseudo-static machine requiring a higher contact time to reach peak pressure (than the real movement) and the desired preload. In reality, in a football movement such as cutting, contact time is very short, roughly 200 ms at a running speed of 2.5 m/s. To increase the preload using MPF, an increase of surface interaction contact time is necessary, which risks damaging the insole senor. Some mechanical limitations were observed when the preload exceeded 1500 N. For example, upper ankle movements such as cutting or turning resulted in anomalies in pressure distribution. Therefore, we set the preload below 1000 N (400, 800, and 1000 N) to allow a safety margin for accurate pressure distribution, which helped us confidently predict translational and rotational tractions in the load higher than 1500 N.

Phase 4. Artificial Neural Network (ANN). Neural networks were implemented in Python using the Tensor Flow framework and libraries such as Keras, NumPy, and Pandas. Translational and rotational traction data were separated. Inputs to the ANNs were stud pattern, type of movement (sprinting/kicking for translational traction and cutting/turning for rotational traction), and load (400, 600, and 800 N). To input the stud pattern, all 16 stud positions were numbered from 1 to 16 (see Figure 4.4.7). If one position had a stud attached, it was designated 1, otherwise 0. For both translational and rotational traction, 10 neural networks were implemented. In general, the ANNs consisted of one input layer with 18 inputs, 2 hidden layers, and one output node, only varying in the number of nodes in their hidden layers. The activation functions were ReLU (Rectified Linear Unit) in the hidden layers and linear in the output. The order of data was randomized and split into a training/validation ratio of 80:20. Weights and biases were initialized randomly.

Randomization of data and random initialization of weights and biases were executed only once to establish baseline values and kept constant for all ANNs. This ensured better reproducibility, as different randomized values can lead to different results. The loss function was the mean squared error. The Adam optimizer was used to update weights and biases. In contrast to stochastic gradient descent (SGD), where the learning rate is constant, Adam is an algorithm that adapts the learning rate to different parameters. The training was stopped when the validation loss did not decrease further after 1000 iterations. Preloads were estimated in the ANN as follows: in cutting 135° = 2600 N, in sprinting= 3300 N, in turning = 1700 N, and in penalty kick = 2600 N. The mean of preloads was 2550 N. These data were extracted from the literature (Smith et al., 2004; Dayakidis et al., 2006; Clarke J, Carré, et al., 2008; Udofa et al., 2017) for cutting, sprinting, and penalty kicks. For the turning movement, the load was measured using the Moticon insole sensor pressure (Stöggl and Martiner, 2017) in our study.

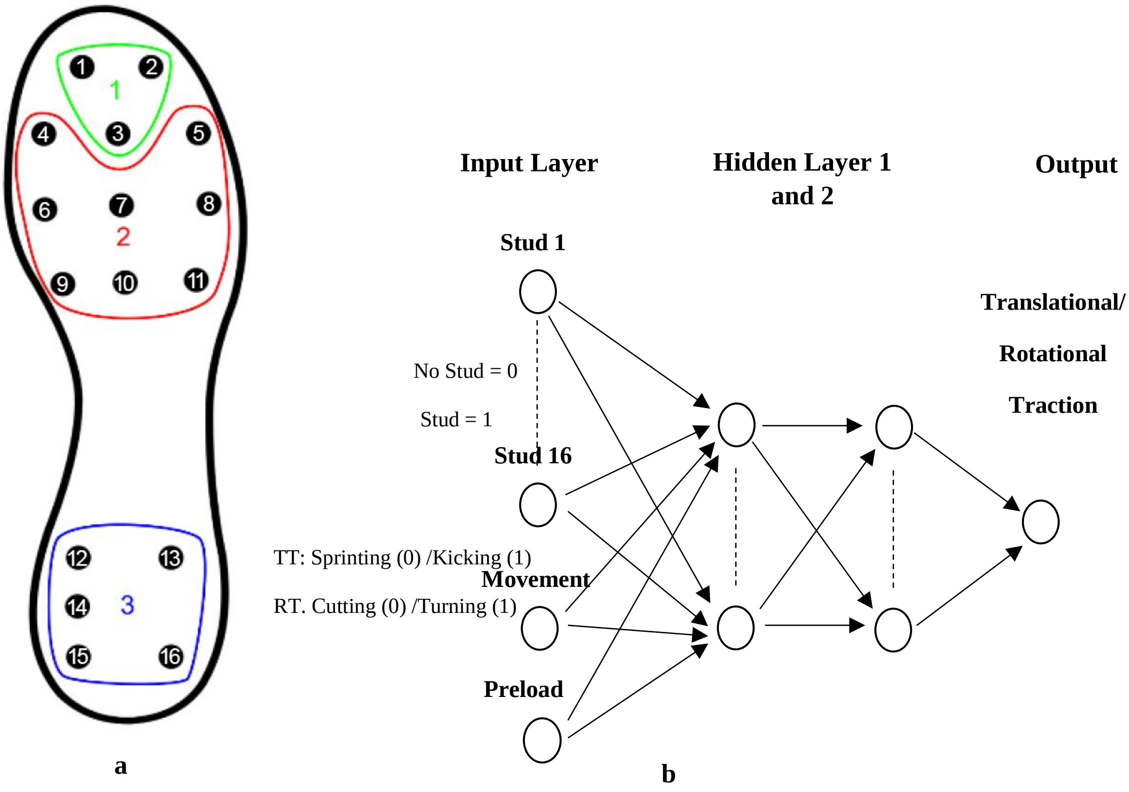
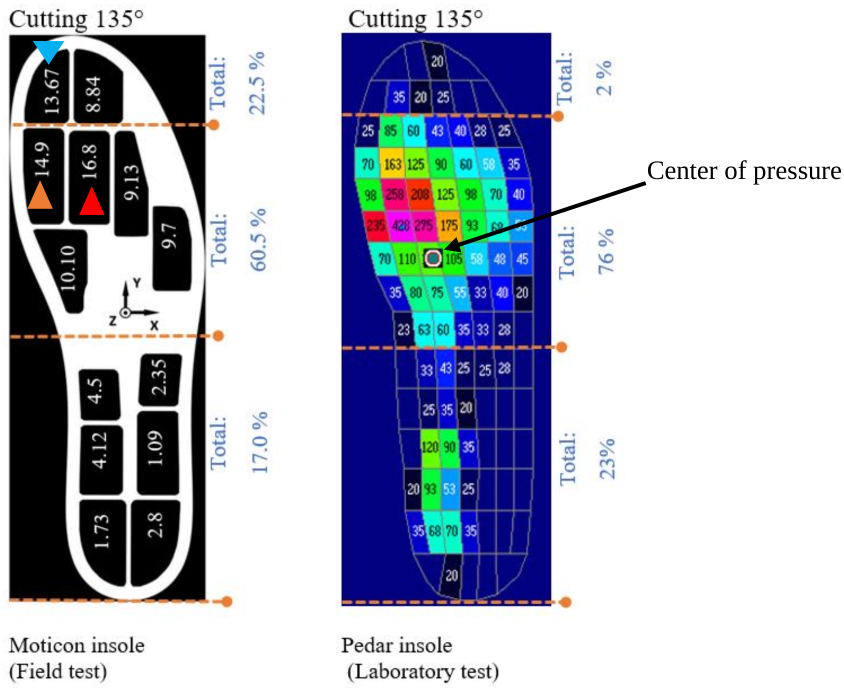


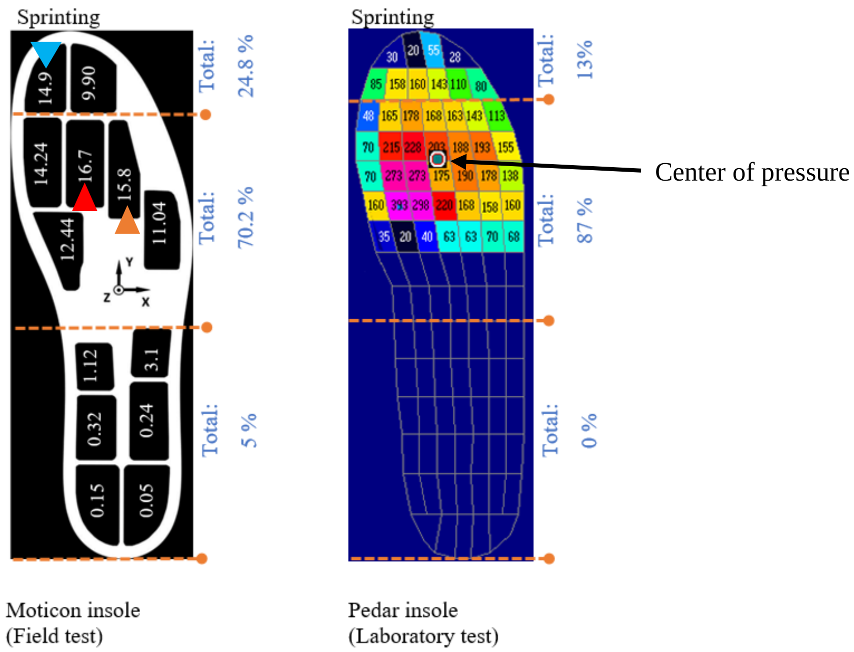
Figure.4.4.7.a Numbering of stud positions. **b** Structure of artificial neural network with two hidden layers. Input layers included: presence of studs (0 or 1), movements (Translational traction and rotational traction) and pre-load.

4.1.4.3 RESULTS

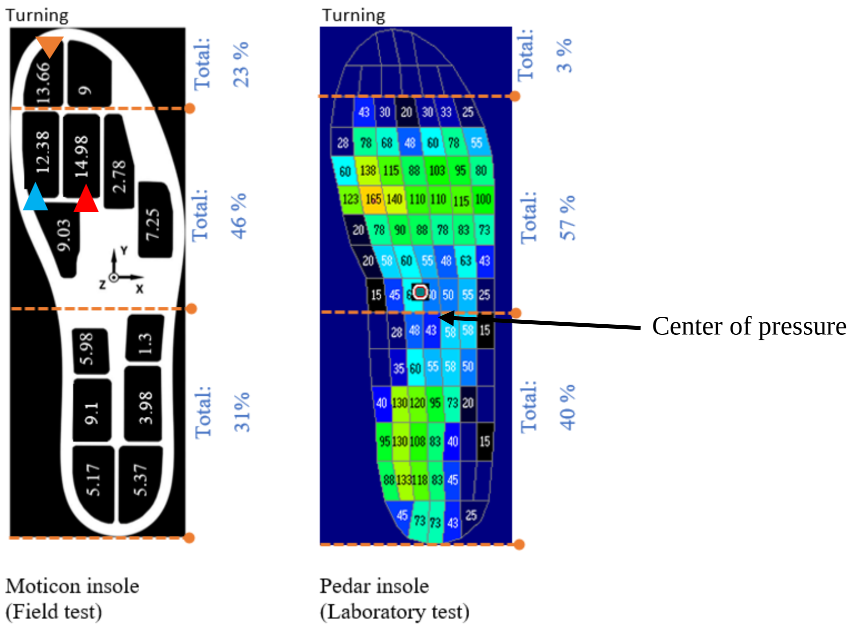
The plantar pressure distributions measured in each movement in the field test were sub-classified into three different sections (see Figure 4.4.8). The peak pressure was normalized (%) for each movement for all 17 participants. Three areas are common to all movements: the medial forefoot, central forefoot, and hallux. Pressure mapping in the laboratory test followed the normalized data (Figure 4.4.8) for preloads of 400, 600, and 800 N.


Cutting 135°
13.67
8.84
14.9
16.8
9.13
10.10
9.7
4.5
2.35
4.12
1.09
1.73
2.8
Total: 22.5 %
Total: 60.5 %
Total: 17.0 %
Cutting 135°
Total: 2 %
Total: 76 %
Total: 23%
Center of pressure
Moticon insole
(Field test)
Pedar insole
(Laboratory test)

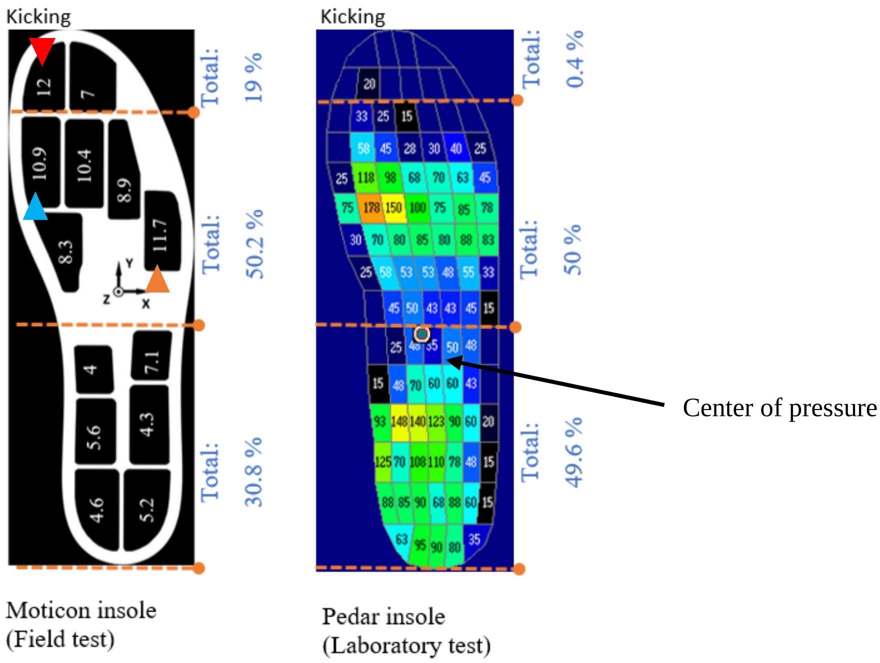
a.


Sprinting
14.9
9.90
14.24
16.7
15.8
12.44
11.04
1.12
3.1
0.32
0.24
0.15
0.05
Total: 24.8 %
Total: 70.2 %
Total: 5 %
Sprinting
Total: 13%
Total: 87 %
Total: 0 %
Center of pressure
Moticon insole
(Field test)
Pedar insole
(Laboratory test)

b.



c.



d.

Figure. 4.4.8. a-d Pressure mapping data of Moticon (in the left shoe) were normalized (%) for each movement and shown in three sections: The three first peak sections of plantar pressure are colored as follows: highest in red, second highest in orange, and third highest in blue. Data of pressure mapping of Pedar insole (on the right side) were normalized for each movement (%) and shown in three sections: The plantar pressure of football boot used in laboratory test by Pedar insole in maximum preload 800 N (Unit is in kPa). The locations of the center of pressure are shown per movement in the laboratory experiment which used the Pedar system.

In the field test, the hardness of the surface was 155g (SD = 4.9) and in the laboratory, it was 155.3g (SD =3.4). The ANNs were run based on overall 480 measurements (8 boots, 3 preloads, 4 movements: each conditions include five trials). 12 measurements were excluded based on

Grubbs' test for outliers. The three best performances based on validation error were chosen for translational and rotational traction (see Table 4.4.1 and 4.4.2).

Table 4.4.1 The three best-performing neural networks in translational traction

hidden layer 1*	hidden layer 2*	training iterations	training loss	validation loss	training error [%]	validation error [%]
20	10	4569	2.1×10^{-3}	1.6×10^{-3}	13.7	9.9
12	6	6522	1.6×10^{-3}	1.8×10^{-3}	11.9	11.1
18	5	5547	8×10^{-3}	1.9×10^{-3}	8.3	11.1

Table 4.4.2 The three best-performing neural networks in rotational traction

hidden layer 1*	hidden layer 2*	training iterations	training loss	validation loss	training error [%]	validation error [%]
20	20	5257	1.9	1.4	9.3	7,0
10	10	4757	2.1	1.9	10.3	8.0
20	18	5574	1.32	2.8	7.8	8.1

*Number of nodes in the hidden layer

Rotational (Nm and translational traction coefficient were predicted for a vertical preload of 2550 N based on the mean results of the three best performances of the ANN (Tables 4.4.1 and 4.4.2). These results are sorted in Table 4.4.3 for TTC from highest to lowest, while the data from rotational traction (cutting and turning) are sorted from lowest to highest. Higher rotational tractions were attributed to 'foot fixation' and knee injuries whereas low translational traction was related to the risk of slippage (Olsen et al., 2003; Wannop et al., 2013; Lambson et al., 1996; Teymouri et al., 2017: Eils E et al., 2004). Therefore, the upper ranges of both RT and TT in Table 4.4.3 are related to the abovementioned injuries and, in addition, such a sorting can facilitate understanding the selection of a 'safe zone, which will be explained in the Discussion below.

Table 4.4.3 Neural networks' mean predicted traction for all stud patterns

Rank	Sprinting		Kicking		Cutting		Turning	
	Pattern	TTC	Pattern	TTC	Pattern	RT(Nm)	Pattern	RT (Nm)
1	C	0.14	C	0.14	G	35.22	G	14.62
2	B	0.13	B	0.13	C	35.39	C	14.80
3	G	0.13	G	0.12	B	35.60	B	15.00
4	F	0.12	F	0.12	E	35.60	E	15.00
5	H	0.10	H	0.10	D	35.90	D	15.25
6	D	0.09	D	0.08	F	37.90	F	17.30
7	E	0.08	E	0.08	H	38.00	H	17.40

TTC= translational traction coefficient, RT=rotational traction (Nm)

The mean translational traction coefficient among all seven boots in sprinting was 0.10 and in kicking 0.11. The mean rotational traction in cutting was 36.23 and in turning 15.62. Based on the ANN predictions, the highest translational traction coefficient in both sprinting and kicking was for stud pattern C while in rotational traction, stud patterns G and C were lowest.

4.1.4.4 DISCUSSION

This study established a methodological framework that can predict traction (rotational and translational) with a mean preload of 2550 N. Such a method should help to identify a safe zone of traction for different outsole configurations in specific football movements. Pressure mappings of four specific football movements were measured in field tests and analyzed by normalized peak pressure. Peak pressure in all four movements was highest in the mid-foot of the plantar surface. The sprinting movement had the highest mid-foot peak pressure (60.8 %) while the turning movement had the lowest, with 46. 40 %.

The lowest pressure in all three plantar sections of all four movements was on the heel in sprinting, with only 4.9%. Turning and kicking movements determined that plantar pressure was distributed more evenly in comparison to sprinting and cutting movements. The difference between the highest and lowest peak pressure values of three plantar sections in turning was (46 % – 23 %) = 23.0 % and in kicking was (50.20 % – 30.80 %) = 19.4. However, such differences were large in sprinting, (70.2 % – 5 %) = 65.2 %, and in cutting, (60.5 % – 17 %) = 43.5 %.

Our findings determined that the peak pressure in all four movements had some common areas, such as the central forefoot, medial forefoot, and hallux. In the cutting (135°) movement, our results showed that for central forefoot I., the medial forefoot and hallux contributed more than other plantar areas in generating peak pressure. This finding is in line with those of Teymouri et al. (2017). Moreover, our results also determined that in cutting (135°), peak pressure in the medial heel was greater than in the lateral heel. This finding is also in line with Teymouri et al. (2017) and Eils et al. (2004).

In sprinting, the medial forefoot and hallux showed the highest peak pressure in comparison to other areas of the plantar surface (this is in line with Eils et al. (2004). In the turning movement, the pressure distribution tended to shift to the medial sections (Hallux, medial mid-foot, and medial heel). In turning movement, the central forefoot I showed higher peak pressure. Finally, in kicking, the peak pressure was highest in the hallux, central, and medial forefoot (this is in line with Teymouri et al. (2017)). In the kicking movement, the lateral heel showed greater peak pressure than the medial heel, which is in line with the findings of Eils et al. (2004). It should be added that plantar pressure measurements cannot take account of all players or their various individual techniques and movements. The Moticon insole system has been used in sports research (Lehner et al., 2013; Björklund et al., 2019) and clinical experiments (Devarajan et al., 2019; Braun et al., 2019) to measure plantar pressure, temporal parameters, and peak vertical force. Some studies show that force impulses were lower using Moticon insole when compared to AMTI (Nagahara et al., 2018), although the insole-sensor (Moticon) has demonstrated moderate-excellent reliability (0.61 – 0.90) (Stöggl et al., 2017; Barratt et al., 2021). Since our study focused on pressure distribution, specifically, peak pressure, we have compared our results with past studies that used Pedar as a gold standard system (Teymouri et al., 2017; Eils et al., 2004; Lehner et al., 2013). As explained in the first paragraph of this discussion, the peak pressure data obtained with the Moticon system in four movements are in line with those of past studies (Teymouri et al., 2017; Eils et al., 2004; Lehner et al., 2013). This finding may be useful for other researchers interested in replicating this methodological framework in that they can be flexible in choosing between Moticon and Pedar insoles.

The MPF was adjusted according to the pressure mapping obtained in the field test. By measuring rotational (turning and cutting 135°) and translational traction (sprinting and penalty kick) in three preloads (400, 600, and 800 N), the three best performances of the ANNs were selected. The result of rotational traction with an estimated preload of 2600 N in cutting (135°) determined that the rotational traction lay in the range of 35-38 Nm. This predicted traction range can be defined in the range of high-risk injuries according to Wannop et al. (2013). However, the result of rotational traction with an estimated preload of 1700 N in the turning movement had values of rotational traction ranging from 14-17 Nm. This entails low risk of injuries, as determined by Wannop et al. (2013). It has been mentioned that higher rotational traction causes foot fixation related to injuries such as ACL (Wannop et al., 2013; Villwock et al., 2009).

According to our results among seven boots, high rotational tractions are specific to outsole stud patterns H and F. However, the risk of slippage due to low RT should be considered. Outsole G and outsole C had the lowest rotational traction. To reduce the risk of injuries and find the desirable range of rotational traction, our findings determined that the *optimal range* is to be found between outsoles B, E, and D for rotational traction. Accordingly, in translational traction, too many studs in the forefoot and mid-foot area cause excessive translational traction. In translational traction, the seven outsoles' translational tractions ranged from 0.08 - 0.14. Outsoles C and B increased traction to nearly 0.14.

The low translational traction coefficient in our study in comparison to other studies, such as Silva et al. (2017) and Valiante et al. (1985) can be seen from different perspectives. The penetration behavior of studded outsoles was influenced by the type of surface. i.e., AT or

Natural grass (NG). Penetration of AT causes instability and lower depth (Abdelhamid et al., 2016; Xiong et al., 2018). The friction coefficient of AT and its interaction with POM studs varies between 0.16 to 0.65 (Potthast et al., 2010). In one of the pin-on-disc tribometer tests, POM had a friction coefficient of about 0.4 at a velocity of 200 mm per second on carbon steel (Hennig et al., 2010). In general, the low friction between the studs and turf could be another reason for the generation of lower traction in comparison to the abovementioned studies. In summary, low penetration of studs on AT, and the low friction coefficient of AT and POM studs could result in low translational traction.

In general terms, a higher translational traction coefficient can affect joint loading, although the perceived grip and perception of running can be increased (Wannop et al., 2012: Wannop et al., 2015). On the other hand, low translational traction will contribute to slippage risk. This indicates that it would be advisable to avoid outsoles with upper ranges (outsoles C and B) and lower ranges (outsoles D and E) of translational traction due to the mentioned risk of injuries. Therefore, outsoles G, F, and H can produce a *safe zone –translational traction*, where the risk/performance tradeoff between joint loading and slippage is optimized. Our study suggests another approach that can also be used to find the optimal outsole configuration. Low translational and rotational traction can cause injuries from slippage. To avoid such a risk, we suggest determining the mean of translational and rotational traction per movement. In sprinting, the mean translational coefficient was 0.10 corresponding to outsole H. In kicking, the mean translational coefficient was 0.11 corresponding to outsole F and H. In cutting and turning, the mean rotational traction was 36.23 Nm and 15.62 Nm, respectively. According to these ranges, the theoretical safe outsole stud pattern with respect to traction (where traction is close to the mean) was outsole D. This outsole might therefore mitigate the risk of injuries since it lies within the identified safe zone of rotational and translational traction. However, this would need to be determined in further research.

Our study provides a methodological framework that determined a traction ‘safe zone’ within our database of loads and conditions (see Table 3). The mechanical prosthetic foot played a significant role in generating rotational and translational tractions in this study. However, there are some limitations regarding the replication of those four movements from field test to laboratory test using this apparatus:

I. FIFA has defined appropriate and acceptable hardness for *NG* in a range between 40 g to 100 g. but not for AT (FIFA Quality Programme, 2022). The AT is generally harder than NG, especially when it comes to outdoor surfaces, and has a duration of four years of usage (Fleming et al., 2020). Logically, some maintenance can be applied to NG, in that its hardness can be maintained or improved over time, which cannot be done to AT. The surface used in this study was an AT with a hardness of 155 g. An advantage of such AT in comparison to NG is that it has fewer deformations due to environmental changes on different days of experiments. Since the football boots used in the field test were stud-free, excessive changes in the environment may influence slippage risk on the outdoor AT. Our priority was to minimize surface variations caused by extrinsic variables in both field and laboratory tests. Another advantage of using AT in our study is that it allowed us to measure traction which may cause a risk of injuries among adolescent players (Aoki et al., 2010). Lastly, using AT in our methodology should help other researchers facilitate the replication of this experiment flexibly.

II. The plantar pressure distribution of four movements replicated with the MPF was not identical to the field test in the forefoot section. The pressure mapping of Hallux and other toes was not replicated similarly to the pressure mapping obtained in the field test. To fulfill the pressure desirable pressure mapping of the field test in the laboratory test, the foot model should have dorsiflexion/plantar flexion and toe extension. These degrees of freedom would help replicate movements more accurately to reach the desired plantar pressure distributions.

Since stud length plays a significant role in stud penetration and consequently in rotational and translational traction (Sterzing et al., 2008), our approach was to use three different heights of studs in the forefoot, mid-foot, and rear-foot sections as per available boots on the market. These mean heights were derived from our data bank. Using a mean of stud lengths from different football boot manufacturers can aid in minimizing bias in producing stud lengths. Such a measurement helps to classify stud length for three different sections. Future studies should investigate the mean of surface areas of all available studs on the outsole to identify the optimal surface area per stud for each section (forefoot, midfoot, and rearfoot).

Using ANN and TUM Tracktester offers a robust method for defining rotational and translation traction for different movements in football, especially under different preloads. Such tools may help footwear developers customize an outsole based on pressure mapping data in different movements, particularly for football. To achieve better results, such customized outsoles should also take account of players' injury histories and anatomical inputs.

The future mechanical prosthetic foot should be able to endure higher preloads (at least 2500 N). This more robust foot would offer the possibility of validating the results of our study. This should be noted that data for tractions (translational and rotational) in higher preloads presented here should be used with caution due to the gap between inputs preload (400 N, 600 N, and 800 N), and estimated output preloads (exceeding 2500 N). In future studies, two modifications will be considered: 1. modifying the foot model to make it behave more like an actual foot while interacting with the surface (changing the design and materials), and 2. Introducing a hydraulic pressure sensor in place of the pressure insole. This modification would aid in increasing reliability and reducing the cost. Validation of our data at a preload higher than 2500 N can be achieved in a future study, where the abovementioned limitations have been minimized. In addition, future research should also consider various outsole configurations to define upper and lower ranges of traction. Since players are equipped with sensory perception and can adapt to some acute traction changes, understanding perceived traction in a field test and comparing this with our mechanical testing machine could aid in defining such a safe zone more accurately and also examine a wider range of conditions.

4.1.4.5 CONCLUSION

We have sought in this study to establish a robust methodological framework within which to identify an optimum zone of translational and rotational traction. Data from the field test was replicated in a laboratory test using a mechanical prosthetic foot with preloads below 1000 N. An artificial neural network was used to predict translational and rotational traction of different studded outsole configurations in realistic preloads of specific football movements: cutting 135°, sprinting, turning, and penalty kicking. Rotational traction related to the cutting 135° ranged from 35 - 37 Nm and rotational traction related to turning movement ranged from 14 - 17 Nm. The translational traction coefficient for both sprinting and penalty kicks was in the range of 0.08 to 0.13 units. We hope that the results of this study may assist other researchers and sports professionals in estimating the risk of injuries for different stud configurations.

4.2 SUMMARY

In the first and second subsections of this chapter, kinetic data on cutting movement, such as pressure mapping, mean and peak pressure, and ground reaction forces, were evaluated. Our findings showed that the medial forefoot, hallux, and medial heel play a significant role in such movements, and the vertical ground reaction force reaches up to 4.12 BW. Section 2 was replicated based on pressure mapping in a quasi-static condition using a TUM Tracktester. Rotational traction of three different outsole configurations was compared on natural grass under three loads: 700 N, 1100 N, and 1400 N. Our findings showed that the full studs pattern defined the highest rotational traction under all three loads (700, 1100, and 1400 N). Next, to understand whether players can recognize traction changes in rotational movement, a subjective assessment (perceived grip) was conducted among participants. Players explained that traction means an interlocking of studs and surface occurrence, defining the grip. This was successfully demonstrated in 78.4 % of all paired tests. In detail, our results determined that players were able to distinguish a difference of 0.9 % to 2.6 % in the rotational traction of three boots: full stud pattern (S14), soft-ground boot (S06), and H-shape (S11). Participants could significantly distinguish changes in traction by changing boots. However, regarding high or preferred traction, players could not perceive any difference between boots. This was shown to be relevant to surface hardness since players experimented on different days (as shown in Section Three by regression model).

The lack of constant surface conditions may prevent the definition of higher or preferred traction among boots. To overcome this barrier in the future, the use of the ‘sensory-trained panel’ method is advisable. This panel characterizes sensory facets of various products. Employing a methodology where 8 to 12 participants collaborate (Lawless & Heymann, 1999), a consensual vocabulary is generated to arrive at an objective description of the products. Despite the ubiquity of the term “shoe traction,” it persists as a technical name originating from the industry, but is susceptible to diverse user interpretations. This introduces the potential for unreliable outcomes owing to disparities in participants’ construals. This is exacerbated by diverse conditions, particularly when alterations occur in the outsole configuration in response to varying surfaces. The main objective of using the panel is the refinement of sensory characterizations. Concurrently, the collective method serves to establish a definition for all participants. A secondary aim of the trained panel is to foster a comprehensive understanding of various traction-related factors among participants, including but not limited to durability, stud material, surface hardness, stud penetration, and pressure mapping. These factors are discussed among participants, and used to modify of the questionnaire using the collectively generated definition. Considering the potential variability in climate and surface conditions, such as frozen or hard surfaces with minimal stud penetration, additional parameters may be needed for the effective evaluation and comparison of traction. Drawing on our empirical observations, Scenario 1 posits that pressure mapping on specific zones of the plantar contributes to the definition of traction, thereby distinguishing between preferred and high traction. In Scenario 2, conversely, if the intensity is high and discomfort accompanies it, the traction may be deemed not preferable. The panel training extends beyond the establishment of a unique traction definition in both scenarios; rather, the aim is to construct a comprehensive framework for defining traction based on various parameters in complex situations, as exemplified by Scenarios 1 and 2, which includes varying surface hardness. Facilitated by the collaborative discourse within the workshop, this definition, along with potential parameters, is shared with all participants. Consequently, participants are better equipped to discern high/low, preferred, and unpreferred traction, even in adverse conditions. The reliability of the method has been tested in many studies on traction; however, it has not been considered in the context of sports footwear.

Similarly, it would also be worth enhancing perception of in torque. The ability to sense and interpret rotational forces or torques acting on the body, typically at joints, involves an awareness and understanding of the magnitude, direction, and rate of change of these rotational forces. This understanding is crucial for executing precise and coordinated movements, especially in cutting maneuvers in football. Torque perception training incorporates exercises that have been utilized in numerous clubs and institutions. These include isometric exercises, resistance training, balance and stability training, functional movements, joint mobility exercises, proprioception drills, visual feedback, varying resistance levels, task-specific training, biofeedback devices, and mind-body awareness practices. The effectiveness of such training is heightened when players can train and improve torque perception with the incorporation of external factors (such as boots and surfaces). To increase torque perception, therefore, training should also involve different components such as rotating between boots, a gradual transition from one surface to another, or using different boots, varying surface, barefoot training, foot strengthening, sensory drills, varying lacing techniques, and wearing socks and insoles with different thicknesses.

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5. APPENDIX

Rotational and translational traction of the laboratory test used in Chapter 3- Section 4

		Rotational Traction (Nm)		*Translation Traction Coefficient*		*Rotational Traction (Nm)*		*Translation Traction Coefficient*	
PRE-LOAD(N)	***Outsole Design***	***Cutting***	***SD***	***Sprinting***	***SD***	***Turning***	***SD***	***Kicking***	***SD***
400	B	20.55*	1.23	0.42	0.03	7.05	1.21	0.21	0.02
600		21.91	0.64	0.37	0.04	6.22	1.71	0.23	0.02
800		21.31	0.65	0.33	0.00	7.18	0.58	0.24	0.02
400	C	18.90	1.07	0.38	0.04	6.26	1.13	0.30	0.02
600		20.49	0.52	0.36	0.01	6.10	0.30	0.27	0.00
800		21.17	1.68	0.30	0.01	8.47	1.26	0.26	0.01
400	D	19.32	0.64	0.34	0.02	6.29	1.21	0.24	0.03
600		22.55	0.44	0.24	0.03	6.38	0.82	0.24	0.02
800		22.87	1.48	0.25	0.02	7.19	0.70	0.24	0.01
400	E	19.87	1.24	0.27	0.00	5.29	1.35	0.26	0.01
600		21.77	0.18	0.28	0.01	7.52	1.29	0.25	0.02
800		20.48	2.45	0.25	0.01	8.05	0.66	0.26	0.02
400	F	21.83	1.49	0.39	0.02	6.73	0.80	0.23	0.02
600		24.31	0.53	0.33	0.04	8.38	0.90	0.26	0.01
800		24.14	1.16	0.27	0.02	10.28	1.20	0.24	0.00
400	G	17.64	1.01	0.34	0.01	5.52	1.19	0.29	0.01
600		19.76	0.33	0.32	0.01	7.32	0.62	0.29	0.01
800		21.47	1.98	0.29	0.01	8.08	0.44	0.28	0.00
400	H	17.82	0.49	0.30	0.02	9.21	1.99	0.26	0.01
600		21.63	1.64	0.30	0.01	9.50	1.33	0.29	0.02
800		24.87	1.04	0.26	0.00	12.18	0.71	0.26	0.01

*MEAN OF FIVE TRIALS

6. LIST of FIGURES

7. LIST of Tables