

THE ROLE OF FOOT REGIONS AND THE CONTRIBUTION OF
SOMATOSENSORY INFORMATION TO QUIET STANCE

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SOMATOSENSORY INFORMATION TO QUIET STANCE**

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I hereby declare that all information in this document has been obtained and presented in accordance with academic rules and ethical conduct. I also declare that, as required by these rules and conduct, I have fully cited and referenced all material and results that are not original to this work.

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ABSTRACT

THE ROLE OF FOOT REGIONS AND THE CONTRIBUTION OF SOMATOSENSORY INFORMATION TO QUIET STANCE

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This thesis investigates the effect of foot somatosensory during quiet stance at eyes open and closed conditions. Ten healthy right handed and footed male subjects were recruited and tested before and after foot somatosensory loss due to the injection of an anesthetic around the nervus tibialis. Force and moment signals were collected by using a force plate, while a pressure pad was used to estimate overall and regional center-of-pressure (CoP) signals at the anteroposterior (AP) direction. 90 seconds long data in quiet stance with three repetitions were collected .

The overall and regional mean pressure and percent weight due to the foot regions, the Root Mean Square (RMS), variance, and RMS velocity were calculated through overall and regional CoP_x signals. Power Spectral Density Function was used to estimate centroidal frequency (CFREQ), 50% and 90% power, and frequency dispersion (FD).

The hind region had the highest pressure of around 80, while the front and mid region presented around 40 and 25 kPa.

Overall CoP_x magnitude and velocity increased significantly after the eyes closed. RMS regional CoP_x increased in the front foot while decreased at the mid foot at somatosensory loss. The velocity RMS was significantly larger on the left foot and smaller on the hind. RMS velocity was higher at front foot compared to midfoot on the left, while it was smaller on the right. 90% power frequency and FD decreased significantly at eyes closed. CFREQ of regional CoP_x was significantly higher on the left foot and mid foot. After the injection, the 50% Power frequency increased at the fore and midfoot regions in the eyes closed condition. In this thesis it has been shown that foot somatosensory has a considerable impact on the control of human erect posture.

Keywords: Postural Control, Quiet Stance, Somatosensory Loss, Foot Regions

ÖZ

AYAK BÖLGELERİNİN ROLÜ VE SOMATOSENSÖRİYEL BİLGİNİN SAKİN DURUŞA KATKISI

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Bu çalışma, gözler açık ve kapalı durumda sessiz duruş sırasında ayak somatosensöri duyusunun etkisini araştırmaktadır. On adet sağlıklı, sağ elini ve ayağını kullanan erkek denekin, tibialis sinirinin çevresine anestetik madde enjekte edilerek ayak somatosensöriyel duyusu bloke edilmiştir. Deneklerden sakin duruş sırasında ayak somatosensöriyel duyusunun kaybindan önce ve sonra bir kuvvet plakası (Force Plate) ve basınç kilimi (Pressure Pad) kullanarak 90 saniye boyunca, üç kere tekrarlanarak kuvvet ve moment sinyalleri toplanmıştır. Ayak üç anatomik parçaya bölünmüştür: ön ayak, orta ayak, arka ayak. Bu veriden ortalama ve bölgesel basınç değerleri ve ağırlık yüzdeleri, genel ve bölgesel ön-arka (AP) moment kolları (CoP_x) hesaplanmıştır. CoP_x sinyalinin zaman türevinden (hız) Ortalama Karekök (RMS) ve varyans hesaplanmıştır. Genel ve bölgesel CoP_x sinyallerinin güç spektral yoğunluğu fonksiyonundan merkez frekansı (CFREQ), 50% ve 90% kesim frekansı, güç spektrumu saçılması (FD) kestirilmiştir. Arka ayak bölgesi 80 kPa civarında en yüksek basınca sahipken, ön ve orta bölge 40 ve 25 kPa civarında bulunmuştur. Genel CoP_x gen-

liđi ve hızı, gözler kapatıldıktan sonra önemli ölçüde artmıştır. RMS bölgesel CoP_x 'i somatosensöriyel kayıpta ön ayakta artarken orta ayakta azalmıştır. Hız RMS, sol ayakta anlamlı derecede daha büyük ve arka ayakta daha küçük bulunmuştur. Hız RMS, sol ayağın orta kısmına kıyasla ön ayakta daha yüksek, sağda ise daha küçük bulunmuştur. Gözler kapalıyken 90% güç kesim frekansı ve FD önemli ölçüde azalmıştır. Bölgesel CoP_x 'in CFREQ sol ayak ve ayağın orta kısmında anlamlı derecede yüksek bulunmuştur. Enjeksiyon sonrası gözler kapalı durumda ön ve orta ayak bölgelerinde 50% güç kesim frekansı artmıştır. Buna mukabil, enjeksiyon öncesinde gözler açıkken 50% güç kesim frekansı ön ve orta ayak bölgelerinde anlamlı şekilde yüksek bulunmuştur. Bu tezde ayak somatosensörü duyusunun insanın dik duruşunun kontrolü üzerinde önemli bir etkisi olduğu ve ayak bölgelerinin dik duruşun kontrolünde farklı görevleri bulunduğu gösterilmiştir.

Anahtar Kelimeler: Postur Kontrolü, Sakin Duruş, Dik duruş kontrolü, Ayağın bedenduyusu (somatosensory), Ayak Bölgeleri



To my dear parents

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LIST OF ABBREVIATIONS

AP	Antero-Posterior
CNS	Central Nervous System
CoP	Center-of-Pressure
CFREQ	Centroidal Frequency
$CoPx$	Component of CoP on x-direction
$CoPy$	Component of CoP on y-direction
$CoPx_{FP}$	Component of CoP on x-direction obtained from Force Plate
$CoPx_{PP}$	Component of CoP on x-direction obtained from Pressure Pad
R_{xx}	Correlation function
τ	Delay Operator
DV	Dependent Variable
EC	Eyes Closed
EO	Eyes Open
FFT	Fast Fourier Transform
FDB	Flexor Digitorum Brevis
FP	Force Plate
FR	Fore Right
FL	Fore(Front) Left
FD	Frequency Dispersion
HL	Hind Left
HR	Hind Right
IV	Independent Variable
IP	Interphalangeal
kPa	kilo Pascals

L	Left
MLA	Medial Longitudinal Arch
ML	Medio–Lateral
MTP	Metatarsophalangeal
ML	Mid Left
MR	Mid Right
CFREQM	Modified CFREQ
PNS	Peripheral Nervous System
PF	Power Frequency
PSD	Power Spectral Density
$G_{xx}(f)$	Power Spectral Density Function
PA	Plantar Aponeurosis
PP	Pressure Pad
W_{FL}	Regional Weight
P_{FL}	Regional Pressure
$W_{FL}\%$	Regional Weight in percent
RM	Repeated Measure
R	Right
SWV	Shear Wave Velocity
S+	Somatosensory intact
S-	Somatosensory loss
μ_k	Spectral Moments
W_{Total}	Total Weight
σ_X^2	Variance



CHAPTER 1

INTRODUCTION

1.1 Motivation of the Study

The human foot comprises 26 bones, which are about 25% of the total 206 bones in the body when both feet are concerned. The human body has around 350 joints, with an impressive 33 in a single foot. In addition, the human foot is connected to 29 muscles, 10 of which originate externally but cross the ankle joint to act on the foot, while the other 19 are intrinsic. It is worth noting that the human body has 600 muscles. The foot is divided into three parts: the Hindfoot, Midfoot, and Forefoot, each made up of different bones and ligaments [16].

“The 26 bones (seven tarsals, five metatarsals, and fourteen phalanges) and six joints (ankle, subtalar, midtarsal, tarsometatarsal, metatarsophalangeal (MTP) and interphalangeal (IP) joints: for a particular ray of the foot make up its four segments: the hindfoot, the midfoot, the forefoot, and the phalanges” [1, 17] (See Figure 1.1).

“The human foot consists of three arched structures: medial longitudinal, lateral longitudinal, and transverse arch [18].” These arches are shown in Figures 1.2 and 1.3. “The greatest amount of motion during stance, occurs in the sagittal plane around the talonavicular joint, which can also be described as the deformation of the medial longitudinal arch” [19].

The human foot has one of the most complex bone structures in the body. It has three arched structures that store and release energy and various sensors located underneath. These sensors, together with three curved ligamentous structures and the plantar aponeurosis send multiple signals to the Central Nervous System (CNS). Hu-

man posture is governed by three main sensors: somatosensory, vestibular and vision (proprioception is implicit) which exhibit everlasting complex oscillations called postural sway [5, 10, 13, 20].

The study of the role of sensory information from various receptors in maintaining posture has been achieved experimentally by measuring body sway in situations where sensory information has been changed or restricted, either by the researcher or pathology.

“Somatosensory loss due to pressure cuffs placed around the ankles of normal subjects standing freely had little effect on spontaneous body movements with eyes open, but resulted in increased hip sway and larger excursions of the center of foot pressure with the eyes closed” [21]. “Pressure cuffs applied above the knees, which is likely to affect ankle muscle proprioceptors, resulted in large, continuous, 1 to 3 Hz body oscillations which could not be eliminated by vision” [22].

This thesis aims to understand the importance of foot somatosensory information in coordinating postural responses. The researchers compared the postural reactions of healthy individuals before and after they experienced somatosensory loss due to an anesthetic injected around the peripheral sensory nerve (N. Tibialis).

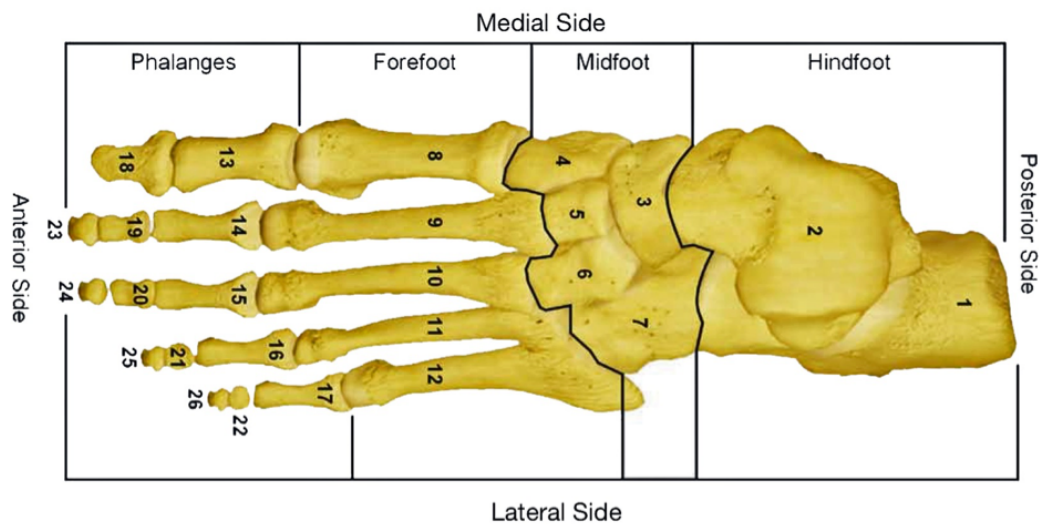


Figure 1.1: The segmentation of foot structure: four segments: hindfoot (1, 2), mid-foot (3-7), forefoot (8-12), phalanges (13-26), (1) calcaneous, (2) talus, (3) navicular, (4) medial cuneiform, (5) intermediate cuneiform, (6) lateral cuneiform, (7) cuboid, (8) First metatarsal, (9) second metatarsal, (10) third metatarsal, (11) fourth metatarsal, (12) Fifth metatarsal, (13-17) proximal phalanges, (18) distal phalange, (19-22) middle phalanges and (23-26) distal phalanges [1]

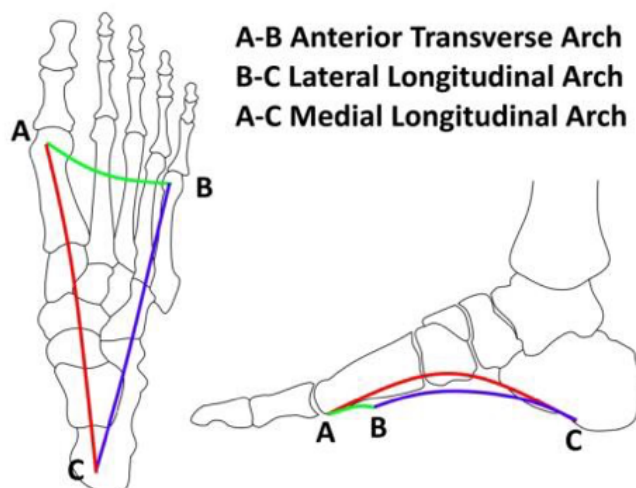


Figure 1.2: Schematic demonstration of the arches of the foot [2]

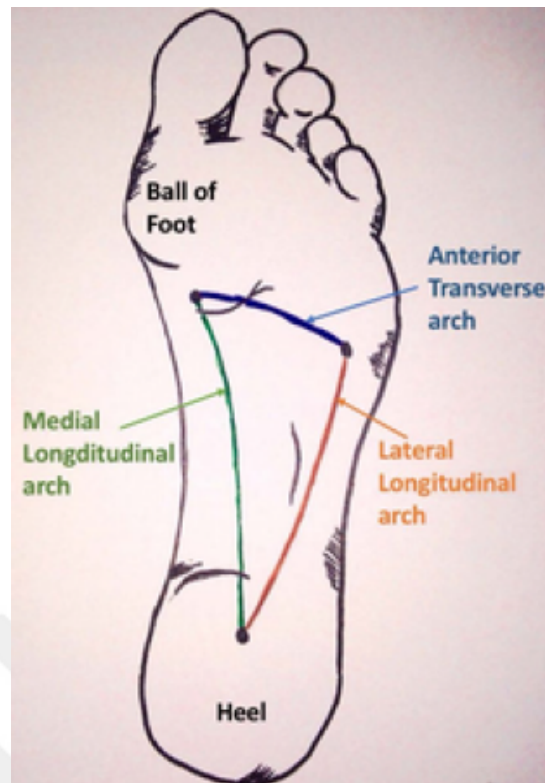


Figure 1.3: Arches of the foot [3]

1.2 Human Posture

Maintaining an upright posture is a daily task that healthy humans perform automatically. However, they are usually unaware of the complex processes that take place in their CNS and peripheral nervous system (PNS) to achieve this. The task of maintaining an upright posture involves aligning the body segments vertically with respect to gravity, which requires integrating multisensory information from various sources such as vision, vestibular, proprioceptive, and cutaneous receptors. This is a developmental process that humans undergo to perform this task efficiently [23, 24].

Human beings are able to stand almost straight due to constant minor muscular corrections. These corrections can only be made if the individual has a reasonable estimate of their position. To determine the correct position, information from various sensors is fed into the central nervous system. This includes visual information from the eyes, vestibular information from the inner ears, proprioceptive information from the muscles and tendons, and somatosensory information from skin receptors.

It is widely accepted that various systems help to maintain a person's posture. However, there still needs to be more understanding of how sensory information from multiple sources is integrated to create a unified and consistent perception of the body's position [25]. Body posture refers to the position of body parts with respect to gravity. On the other hand, balance refers to the capability of keeping the body's center of mass within its base of support. Swaying is the horizontal movement of the center of gravity while standing still and is influenced by the body's inertial forces and segmental characteristics. The postural dynamics can now be seen as an interplay between postural facilitation and information availability [26].

The center of mass (CoM) is a point in space where an object's mass is equally distributed. This means that the weighted position vectors relative to this point add up to zero. The CoM is the weighted average position of all body segments in 3D space that is controlled by the balance system. The CoM's vertical projection onto the ground is often called the Center of Gravity (CoG) [27, 28]. The Center-of-Pressure (CoP) is the point on a body where the total force field is acting, resulting in a force that moves through it. This point represents a weighted average of pressure distribution over the surface of the area in contact with the ground. It's important to note that the CoP is independent of the CoM. In a one-leg stance, the net CoP is located within the foot in contact with the ground, even if the CoM may be positioned outside the body. However, in a two-legs stance where both feet are in contact with the ground, the net CoP is located between the feet, with separate CoPs under each foot [29, 30, 31].

1.3 Human Balance and its Sensory Motor Coordination

A healthy and properly functioning balance system enables clear vision while moving, distinguishes direction concerning gravity, specifies the orientation and speed of movement, and makes automatic and involuntary postural adjustments to maintain posture and stability in various situations and activities. The human body's balance is regulated by a variety of complex control systems, which depend on sensory input from different sources. These include vision or sight, proprioceptive sensors that are present in the muscles, somatosensory senses that come from skin receptors such as touch, and the vestibular system that helps maintain motion equilibrium and spatial

orientation. Apart from these sensory inputs, some psychological factors can also affect our sense of balance.

Human beings have complex sensorimotor systems that maintain balance, which can be disrupted by injury, illness, or aging, leading to symptoms like dizziness, vision problems, nausea, fatigue, and difficulty in concentrating. "The complexity of the human balance system presents difficulties in diagnosing and treating the underlying cause of imbalance"[4]. The vestibular, visual, proprioceptive, and somatosensory systems work together to maintain balance, and disruptions to any of these systems can cause imbalance. Vestibular dysfunction is particularly challenging due to its impact on cognitive functioning, eye movements, and posture control [32].

1.3.1 Sensory Information

The human brain receives information through sensory receptors from four primary sources: the eyes, muscles and joints, skin, and vestibular organs. This information is transmitted to the brain through nerve impulses (See Figure 1.4).

“Sensory receptors in the retina are called rods and cones. Rods are believed to be tuned better for vision in low light situations (e.g. at night time). Cones help with color vision, and the finer details of our world. Rods and cones transmit signals to the brain to provide visual cues that identify a person’s orientation relative to other objects. **Vision** plays a noteworthy role in balance. Pretty nearly twenty percent of the nerve fibers from the eyes interact with the vestibular system” [4].

Sensory receptors are responsible for detecting stretch or pressure in the environment. They send impulses to the brain with every body movement. The brain uses these cues to determine the body’s position in space. This is primarily done through **proprioception**, which is provided by muscle spindles. Muscle spindles measure the changing length of a muscle as it contracts and stretches back to its original size. They are found throughout the skeletal muscle.

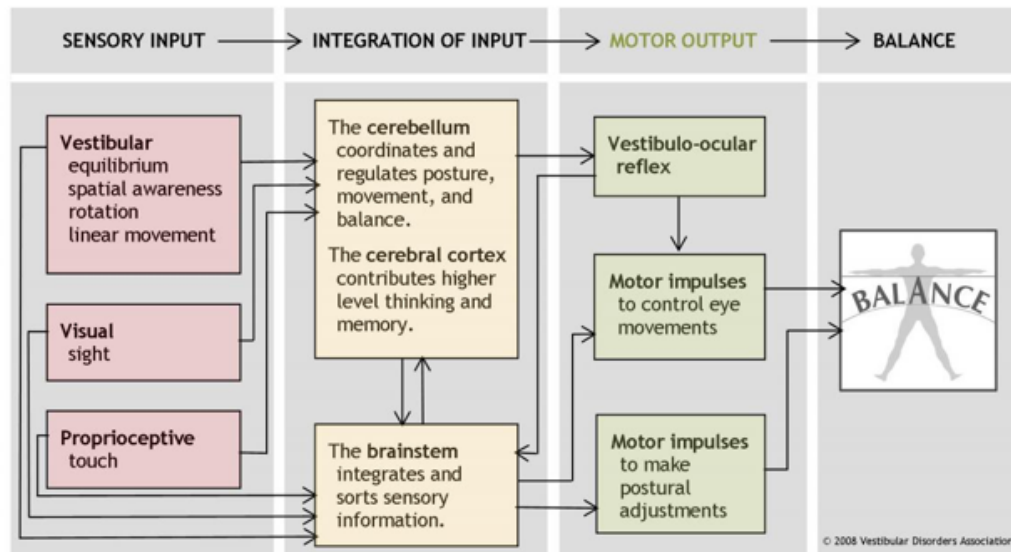


Figure 1.4: Balance motor output (balance is achieved and maintained by a complex set of sensorimotor control system) [4, 5]

The **vestibular** system plays a crucial role in maintaining balance and spatial orientation. It includes the utricle, saccule, and three semicircular canals in each ear, which provide sensory information about motion, equilibrium, and spatial orientation. The utricle and saccule detect gravity and linear movement, while the semicircular canals detect rotational movement. This system transmits symmetrical impulses to the brain when it is working correctly.

The **somatosensory** nervous system is a highly complicated network of sensory receptors distributed throughout the body. These receptors detect changes in the external or internal environment of the body. They are responsible for perceiving sensations such as touch, pressure, pain, temperature, body position, movement, and vibration [33, 34].

1.3.2 The Somatosensory System of the Foot

The posterior tibial nerve originates from the sciatic nerve and runs down the back of the thigh and the inner side of the lower leg as shown in Figure 1.5 [6]. These nerves

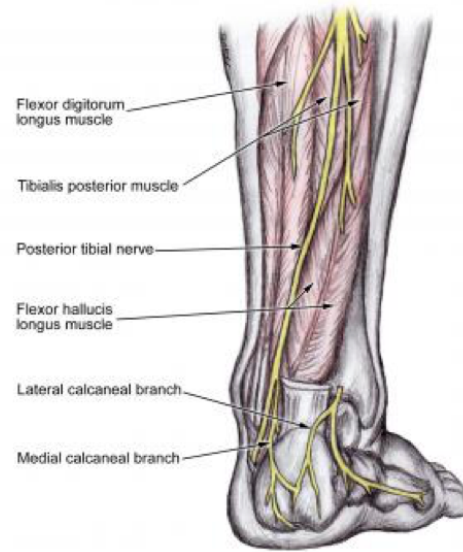


Figure 1.5: Posterior tibial nerve courses down the posterior calf [5, 6]

provide the foot's intrinsic muscles and sensory innervation to the plantar of the foot, except for the extensor digitorum brevis, as shown in Figure 1.6. The posterior tibial nerve can be found halfway between the medial malleolus and the heel at the ankle level [6]. The sensation received from the soles of our feet, known as plantar cutaneous sensation, plays a vital role in helping us maintain balance and stability while standing still. This sensory information provides feedback to our central nervous system, which helps us control our posture and make adjustments to keep our balance and stability intact [35].

1.4 Foot Biomechanics

The human foot is a complex structure that plays a vital role in the biomechanics of the lower extremities. Researchers in this field aim to understand how the foot supports the body and how sensory feedback from the soles of the feet, particularly the plantar cutaneous sensation, affects balance and stability while standing without external disturbances. The plantar sole provides real-time sensory information to the central nervous system, continually monitoring the pressure distribution and orientation of the body, which is crucial for maintaining a balanced posture. Biomechanics

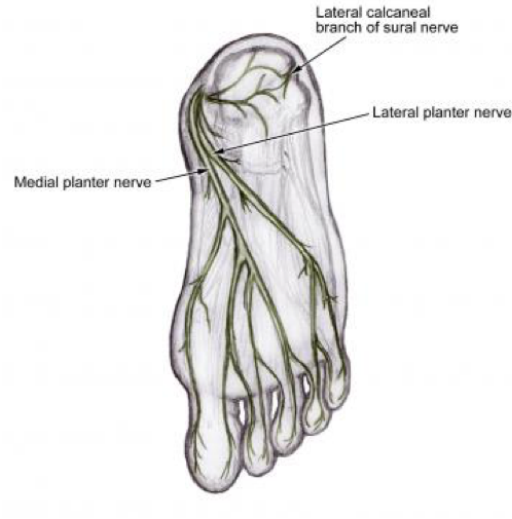


Figure 1.6: Medial and lateral plantar nerves after branching from the posterior tibial nerve [5, 6]

of the foot studies also cover the windlass mechanism, which is essential for maintaining foot arches and improving biomechanical efficiency during walking, running, and jumping [1, 35-42]. From the point of postural control, studying how different muscles with somatosensory feedback from plantar surface of the foot in the lower extremities work together to maintain balance and stability is important [36, 37].

The literature survey in the second chapter provides a detailed report on foot biomechanics.

1.4.1 Truss and Beam Function of the Foot

The truss and beam function of the foot describes how bones, ligaments, and tendons collaborate to provide structural support that distributes weight and maintains stability during movement. The foot's arch acts as natural trusses, absorbing and distributing forces, while the bones and connective tissues act as beams to support the body's weight and facilitate movement (See Figure 1.7) [7, 38, 39].

Truss Function : The human foot consists of three arches, namely the medial longitudinal, lateral longitudinal, and transverse arches. These arches act as natural trusses that are similar to the trusses found in structures. They are formed by a combination

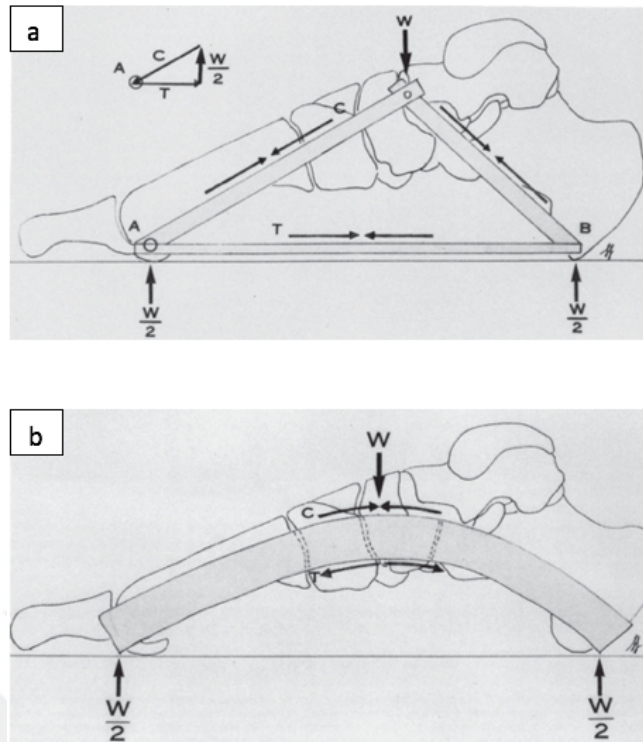


Figure 1.7: a) Truss and b) Beam mechanism of the foot [7]

of bones, tendons, and ligaments, which work together to evenly distribute the body's weight across the foot. This lattice-like structure of the foot helps to absorb the forces generated during activities such as walking, running, and standing, thereby reducing the impact on joints and soft tissues.

Beam Function : The bones of the foot, which include the metatarsals, tarsals, and phalanges, function as beams, providing stability and supporting the weight of our body. These bones are arranged to create longitudinal and transverse arches that assist in distributing the forces exerted on the foot when bearing weight. The ligaments and tendons surrounding these bones act as reinforcements, increasing the structural integrity of the foot and preventing excessive movement or collapse of the foot.

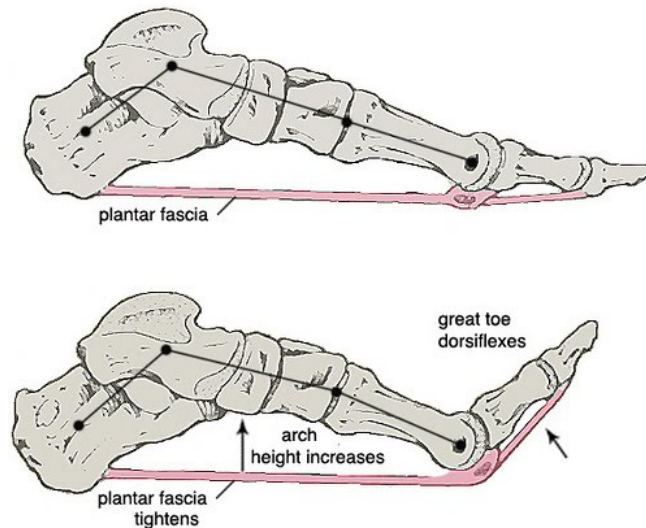


Figure 1.8: The plantar aponeurosis originates from the base of the calcaneus and extends distally to the phalanges [8]

The truss and beam functions of the foot work in tandem to preserve stability, absorb shock, and enable smooth movement. Any imbalance or dysfunction in these structures can result in foot issues such as flat feet, high arches, or overpronation.

1.4.2 Windlass Mechanism

The windlass mechanism refers to the anatomical function of the base of the foot, which involves the plantar aponeurosis, sesamoid bones, plantar pads, and their attachment beneath the metatarsalphalangeal (MTP) joint. In research conducted by Hicks, the term "windlass" was used to explain how the bow-string structure of the plantar aponeurosis pulls the calcaneus and first MTP closer together. This happens when the first ray moves into extension. As a result, the metatarsals rise and leads to an increase in the medial longitudinal arch (MLA) height by increasing the tension on the plantar fascia (See Figure 1.8).

"The plantar fascia is a connective tissue that works like a cable connecting the calcaneus and the MTP joint. During the propulsive phase of gait, dorsiflexion winds the plantar fascia around the head of the metatarsal. This winding of the plantar fascia shortens the distance between the calcaneus and metatarsals and raises the medial

longitudinal arch. The shortening of the plantar fascia due to hallux dorsiflexion is the core principle of the windlass mechanism" [40]. Plantar fascia prevents foot collapse by maintaining the medial longitudinal arch and preventing the spreading of the calcaneus and metatarsals due to its anatomical orientation and tensile strength [38, 41].

1.5 Content and Thesis Statement

This study aims to explore the impact of somatosensory information on the coordination of postural responses during quiet stance under various vision and somatosensory conditions. The role of somatosensory information was examined by comparing the postural responses of healthy participants before and after they experienced somatosensory loss caused by the injection of an anesthetic around the peripheral sensory nerve (N. Tibialis).

In the first step of walking, humans use their leading foot to maintain balance by detecting slight surface irregularities and providing information for motor control. The study examined human behavior with and without somatosensory and visual senses. Models of inverted pendulum systems with sensory feedback are commonly used to understand the physiological mechanisms that control and stabilize posture during quiet stance [42, 43]. In order to better understand how people use their feet, the foot was divided into three regions (Forefoot, Midfoot, and Hindfoot) using an image processing method. This method reveals the motile and supporting areas of the foot and gain insight into how subjects transmit weight and maintain balance [5, 10, 13, 38, 39, 44, 45].

In addition, in this thesis, statistical analysis will address three important factors:

The first factor is foot regions cooperation: The different parts of the foot, namely the forefoot, midfoot, and hindfoot, work together to maintain optimal balance and stability. Each part plays a distinct role in the body's balance. The forefoot, which includes the toes, acts as the first point of contact with the ground in some of the human jumping activities. It helps in the precise distribution of weight and facilitates pushing off during walking movement. The midfoot, which includes the arch of the

foot, acts as a shock absorber and a bridge. It adapts to uneven surfaces and provides flexibility. Finally, the hind foot, including the heel and ankle, stabilizes the foot and absorbs impact forces. Together, these zones form a dynamic and coordinated system that adapts to various surfaces and movements. This ensures that our balance is maintained with efficiency and precision [1, 46, 47, 48].

The second factor is sensory modulation, which is the process of regulating input from different sensory modalities. The interaction between visual factors (related to the eyes) and somatosensory factors (related to touch and body position) is complex and plays a vital role in our overall perception and awareness. For example, seeing and touching an object at the same time improves our understanding of its properties. Information from the eyes and somatosensory system is processed and modulated to create a coherent sensory experience [35, 49].

The third factor is foot lateralization: Foot lateralization plays an important role in maintaining balance and stability. The left foot and right foot are integral components of our body's complex balancing mechanism. Each foot contributes uniquely to our stability. The right foot usually provides a base for weight distribution, while the left foot assists with precise movements [50]. Together, they form a harmonious interaction that adapts to changes in body weight and changes in location. The right foot can take on a more solid role, absorbing impact and supporting basic balance; the left foot, on the other hand, can make fine adjustments, facilitating agile movements and ensuring adaptation. This lateralization helps effectively distribute the body's load, prevent falls, and increase overall stability in various activities [13, 28, 45, 51].

1.6 Aim of the Thesis

1) Foot regions Cooperation: The front part of the foot, with the help of the toe, acting as the driving force, while the middle part of our foot bearing the load and creating a bridge-like effect will be demonstrated. On the other hand, the hind foot is proposed to be comparatively more stable and supportive. To measure how these different areas work together, time and frequency domain measures utilized to explore the collected data.

2) Sensory Modulation: This thesis aims to investigate the question regarding the influence of somatosensory information on postural responses during quiet stance under various visual conditions. To support this aim pressure pad data collected over different somatosensory and visual conditions were used.

1.7 Thesis Outline

Chapter 1 introduces postural control and dynamics, sensory-motor performance and balance, the somatosensory system, and foot somatosensory.

Chapter 2 presents a literature review on previous studies about somatosensory information in postural sensory-motor control and foot biomechanics.

Chapter 3 explains the experimental protocols and data collection procedures, including a brief description of the participants and equipment used for data collection and analyses. Data collection conditions and the technique used for somatosensory blockage of nervus tibialis were introduced.

Chapter 4 provides the experimental results in time and frequency domain. It is structured to present first the pressure and CoP_x results for the overall posture. On the other hand, the individual left and right foot and corresponding front, mid, and hind foot regions pressure and regional CoP_x (moment arms) were analyzed next.

Chapter 5 discusses the relations and coherence between foot and posture under different sensory conditions.

Chapter 6 presents the conclusions of the thesis and the future plans.

The references used in this study are listed at the end of the thesis. In the Appendices, additional information about the measurement and data analysis of the pressure are presented. The second Appendix involves handedness inventory questionnaire run before the data collection.

CHAPTER 2

LITERATURE SURVEY

This chapter provides a summary of essential observations and findings from earlier research and publications, based on a broad literature survey.

2.1 Postural Research Background

2.1.1 Postural Control Strategies

The investigation of the upright equilibrium of free-standing humans was initiated by Lewis M. Nashner in 1976 [52]. In 1990, F.B. Horak conducted a study on postural control during a quiet stance in somatosensory loss conditions [53]. In this study, it was discovered that the loss of foot somatosensory feeling leads to a higher reliance on the hip strategy. However, when there is vestibular loss, the ankle strategy remains normal while the hip strategy is lacking. This indicates that when the vestibular apparatus is functioning properly, individuals use their body, particularly their trunk, as if it were an inverted pendulum to maintain balance and/or orientation. On the other hand, when foot somatosensory is lost, there is a significant discrepancy between the trunk's position and CoP due to the inability to sense the surface [54]. Furthermore, it has been shown in the previous literature that the postural control strategy has shifted from ankle to hip in the case of somatosensory loss [53].

Schieppati and P.G. Morasso studied how muscle stiffness affects balance while standing. They hypothesized that muscle stiffness alone can stabilize the human body and eliminate the need for complex brain feedback systems. The two main shortcomings of the stiffness control model are illustrated by their examination. First, they suggest

that the observed in-phase relationship between COP and COM trajectories is a result of underlying physics rather than active control strategies. Second, they argue that the stability demonstrated in standing human posture cannot be explained by physiological values of muscle stiffness. Ultimately, the authors concluded that although muscle stiffness helps maintain postural stability, it cannot be used alone to maintain standing. This means that the brain's feedback systems play an important role in the ongoing changes necessary to maintain the body's balance [55].

A.V. Alexandrov et al. 2001 study looks into how people coordinate their movements and keep their balance when bending their trunk forward. The research was split into two sections: experimental analysis and modelling. The first stage in the research was to build a biomechanical model to comprehend how posture and trunk motions coordinate. The model investigated possible postural control strategies the body employs to bend while keeping its equilibrium. The first step in the analysis procedure was dissecting the movement into dynamically separate components that were tied to the knee, ankle, and hip degrees of freedom. Understanding how these joints impact balance and general movement was the study's main objective. The study's second section included experimental data gathered from sixteen participants who quickly bent their trunks forward. Both a broad platform and a narrow beam were used for testing the subjects. The investigation showed that the knee played a very minor role in the movement, mainly using the hip and ankle joints. The study made clear how important it is to anticipate changes at the ankle joint in order to start a movement and keep one's balance. The results indicated that in order to accomplish the dual objectives of bending and balancing, the central nervous system progressively regulates these joint motions (hip and ankle). This is consistent with previous researches on postural control strategies and supports the idea of distinct control strategies for different joints. This postural mechanism was first deciphered by Horak and Nashner [56].

According to Jeka and Peterka [57, 58], the human body functions as an inverted pendulum that pivots around the ankle when standing still. However, F.B. Horak [59] found that in challenging situations, such as being on a soft surface or when tilted, multiple coordination patterns are observed.

2.1.2 Postural Dynamics and Sensory Motor Performance

T. Mergner et al.'s 2003 study provides a thorough description of how the human body balances itself by utilizing a variety of sensory information. In order to describe how these senses interact to control posture and adapt for perturbations in balance, the model integrates visual, vestibular (inner ear), and somatosensory (proprioception and touch) data. The study highlights the significance of sensory integration in preserving an upright posture and illustrates how deficiencies in any sensory system might impact the overall regulation of balance [60].

Senih Gürses et al. (2006) use a phase-plane representation of CoP data to study the complex oscillatory behaviour of human postural sway. The study uses time recordings from a healthy person standing still. The researchers suggest that postural dynamics are controlled by a sensory threshold causing nonlinearity in the dynamic. The nonlinear source manifested threshold type dynamics manipulated by ethyl alcohol intake. The nonlinear dynamics were revealed by analyzing complex sway patterns in the phase space [61].

In a study performed by Prieto et al. [62], the sensitivity of CoP based measures was evaluated with respect to postural steadiness under various eye conditions. The study found differences in postural steadiness between the eyes-closed (EC) and eyes-open (EO) conditions in both young and elderly adult groups. However, the differences were more significant and statistically substantial in the elderly adult group than in the young adult group.

Peter F. Meyer's 2004 paper explores the importance of sensory information from the soles of the feet, precisely the plantar cutaneous sensation, in maintaining balance and stability while standing without external disturbances. This sensory input provides feedback to the central nervous system, aiding postural control and adjustments to maintain equilibrium. The paper likely delves into how the nervous system integrates and utilizes this sensory information to regulate muscle activity and coordinate movements, contributing to our understanding of human balance and motor control. Meyer concluded that decreasing sensitivity in the forefoot through local anesthetic injections did not alter the magnitude of CoP displacement, but it did increase CoP

velocity [35].

According to a study conducted by Robert Creath et al. [63], the quiet stance posture of the body cannot be described by a single degree of freedom. Instead, the body behaves like a multi-link pendulum with two distinct modes. Our research in this thesis corroborates this conclusion. When the input from sensors such as vision and somatosensory systems is restricted, changes on the anteroposterior (x-axis) and mediolateral (y-axis) axes become more evident. However, the shifts are more significant on the x-axis since the y-axis is moderately stable. Research indicates that the foot sensation is critical in maintaining our balance, especially when combined with other sensory information. However, the studies conducted so far have yet to explore the specific nature of the touch and the characteristics of foot regions required for this purpose.

Frédéric Viseux et al. (2019) study the role of the foot's sensory input in maintaining balance and stability. The authors describe how engaging plantar cutaneous receptors might enhance postural stability. They emphasize that such stimulation improves sensory perception and processing, allowing the central nervous system to better adapt posture and improve balance [64].

2.2 Foot and Posture Research Background

Research in the field of foot biomechanics covers a wide range of topics, including quiet stance, gait analysis, foot shape, body kinetics and kinematics and muscle function. It helps us understand the mechanics of walking and running, the impact of different surfaces and shoes on foot function, as well as the role of muscles and tendons in foot movement and stability, and how various pathologies and injuries affect foot biomechanics.

R. J. Abboud, in his paper (2002), states that the foot is a complex structure of bones, joints, and soft tissues that plays a crucial role in the lower extremity biomechanics. Both intrinsic and extrinsic muscles control it. The foot's essential mechanical structures consist of three main components: Firstly, the bony skeleton, along with the ligaments and arches, provides the necessary rigidity and lever arm mechanism

to maintain balance while standing and facilitate propulsion. Secondly, the joints provide the foot with flexibility. Thirdly, the muscles and tendons control the foot's movement.

Moreover, he mentioned that the foot region comprises four segments: the hindfoot, the midfoot, the forefoot, and the phalanges [1].

Sarrafian in his book, explains that the foot is a complex structure that bears weight during activities like standing, walking, running, and jumping. To distribute weight effectively and reduce stress on the foot, various biomechanical mechanisms come into play. These include foot arch support, foot joints and bones, muscle activation and control, proprioception and feedback, load transfer mechanisms, and shoes and orthoses [38, 65].

Emily Keshner's paper focuses on the patterns of muscle activation and coordination related to postural stability from head to foot. The study explores how various muscles in the body, from those controlling the head and neck to those in the lower extremities, work together to maintain balance and stability. It examines how changes in muscle activation patterns impact postural control during different situations, such as standing, movements, or in response to external disruptions. These coordination patterns can provide insights into how the neuromuscular system integrates sensory information and produces muscle activity to achieve and sustain stability in different postural tasks [66, 54].

Kavounoudias, 1998 paper suggests that the soles of our feet act as a map for controlling our balance. This means that the pressure distribution and orientation of the body are constantly monitored by the sensory information gathered from the plantar sole, which provides real-time feedback to the central nervous system. This information is crucial for maintaining balance and stability during activities like standing and walking. The paper likely discusses how changes in plantar sensation can impact this map, affecting our ability to control balance. In her paper published in 2001, Kavounoudias suggests that both the foot sole and ankle inputs together regulate human posture while standing upright. This implies that sensory feedback from both the soles of the feet and the ankles are crucial in maintaining a balanced posture. The study examines the relationship between sensory inputs from these areas and how the

central nervous system integrates them to adjust posture and balance in response to different environmental conditions and challenges. Understanding how foot sole and ankle inputs work together provides insights into the intricate mechanisms involved in regulating human posture [49, 67].

2.2.1 Function of Plantar Fascia and Windlass Mechanism

Hicks (1955) explores the mechanical principles of the human foot as a weight-bearing structure. In his discussion, Hicks highlights the significance of comprehending the mechanical behavior of the foot as a support framework for the human body. He stresses the crucial role played by the foot in providing stability, shock absorption, and driving force during activities such as standing, walking, and running. Additionally, he emphasizes the distinctive structure of these components - bones, joints, ligaments, muscles, and soft tissues - that enable the foot to conform to different surfaces and absorb forces efficiently [17, 39, 15].

J.H. Hicks' 1956 study offers a thorough examination of the foot's anatomical and functional biomechanics, with a particular emphasis on the plantar aponeurosis and medial longitudinal arch. According to Hicks, the foot is shaped like an arch, or truss, with the plantar fascia serving as a tie-rod and the calcaneus, metatarsals, and mid-tarsal joint forming the arch. Hicks provided several important ideas, one of which is the windlass mechanism. During toe dorsiflexion, the plantar fascia (aponeurosis), especially at the first MTP joint, tightens around the metatarsal heads. The plantar fascia wraps around the metatarsal heads as the toes dorsiflex, reducing the gap between the calcaneus and the metatarsals to raise the arch and stabilize the foot. In later stages of gait, this process transforms the foot from a flexible framework needed to absorb impact into a rigid lever needed for effective propulsion. Hicks' research highlights how important the plantar fascia is in maintaining the structural integrity of the foot and how the windlass mechanism is important for dynamic exercises such as running and walking. This knowledge has applications in the fields of orthosis design, diagnosis and treatment of foot diseases, and the mechanical foundations of both healthy and unhealthy feet. Because the windlass mechanism plays a crucial role in maintaining the structure of the foot arches and improving biomechanical efficiency

during activities such as walking, running, and jumping, any issues or abnormalities in the windlass mechanism can lead to problems related to the foot and posture, such as plantar fasciitis, flat feet, or changes in gait patterns [44].

Wangdo Kim and Voloshin, in their research paper, investigated the importance of the plantar fascia in supporting the weight of the human foot. The study explores the role of the plantar fascia, a thick tissue beneath the foot, in distributing and supporting the body's weight during various activities such as standing, walking, and running. The study examines the biomechanical properties of the plantar fascia, such as its elasticity and tension, and how these factors influence the foot's ability to bear and distribute loads [46].

Wright et al. (2011) discuss the unique design of the foot for propulsion and support from anthropological and biomechanical perspectives. During quiet standing, there is a significant correlation between the tilting of the tibia (shin sway) and the flattening of the foot arch. When the tibia tilts forward, the foot arch tends to flatten, and when it tilts backward, the foot arch tends to become more noticeable. This emphasizes a dynamic interaction between the lower leg and foot mechanics during standing, where changes in the orientation of the tibia affect the shape and stability of the foot arch [68, 69, 70].

Shuhei Nozaki et al.'s 2022 research explores the biomechanical characteristics of the plantar aponeurosis (PA) of the human foot during quiet stance. In the study, ultrasonic shear wave elastography was used to determine the stiffness and natural length of the PA in young, healthy people. Notable discoveries include: 1. Hardness Measurement: Researchers measured the hardness of PA by measuring shear wave velocity (SWV). Young's PA modulus of 64.7 ± 9.4 kPa suggests substantial stiffness, which increases exponentially with dorsiflexion of MTP joint. 2. Natural Length: When the MTP joint is plantarflexed by approximately 13.8° , the PA reaches to its natural length. This indicates that when standing quietly, the PA is in an extended position due to compression. 3. Intrinsic Muscle Activation: According to this study, PA stiffens while standing as a result of both extension and activation of intrinsic foot muscles. This enhanced rigidity that is required to keep the arch structure and stability during quiet stance [71].

2.2.2 Energy Characteristics of the Foot Arch

Ker et al.'s 1987 research investigates the mechanical features of the human foot's arch, specifically its function as a spring-like structure during movement. The study focuses on the following major points: 1. Arch Mechanics and Energy Storage: Like a spring, the human foot's arch stores and releases mechanical energy during walking and running. Having this ability is essential for efficient locomotion since it reduces the energy required by muscles while increasing endurance and speed. 2. Arch Height and Stiffness: According to the research, the stiffness and height of the arch have a significant impact on its performance. Higher, stronger arches are better at storing energy. 3. Elastic Properties: This spring mechanism relies heavily on the plantar fascia, a thick band of tissue that runs around the bottom of the foot. Its elastic qualities enhance the arch's ability to store and return energy, acting as a natural shock absorber. To summarize, this work provides information on the biomechanics of the human foot, highlighting the complex design and role of the arch in human mobility [47].

2.2.3 Intrinsic Foot Muscles and Arch Stability

L.A. Kelly et al. 2014 research investigated the involvement of intrinsic foot muscles in preserving the structure and function of the foot's longitudinal arch. The study shows that these muscles are more than just passive structures; they actively contribute to the stabilization and control of the arch throughout various activities such as standing, walking, and running. The researchers employed experimental approaches to determine the activity of the intrinsic foot muscles and how they affect arch deformation. They discovered that these muscles can change their activity to dynamically support the arch, affecting its shape and mechanical qualities, and actively changing the position of CoP in the anteroposterior direction. This ability aids in the absorption and redistribution of forces generated during movement, which is critical for maintaining balance and preventing injuries. Overall, the study emphasizes the importance of intrinsic foot muscles in foot biomechanics, specifically their active role in maintaining the longitudinal arch's integrity and function [72].

Welte et al. (2018) investigated at how the windlass mechanism influences the mechanical behavior of the foot arch during movement. The study discovered that dorsiflexion of the MTP joint activates the windlass mechanism, causing the arch to elongate and the plantar fascia to absorb and release more energy. Shifting from stance to swing phase during walking allows the plantar fascia to function more effectively as an elastic band (spring) during movement [73].

Luke A. Kelly's 2019 research focuses on how the intrinsic muscles of the foot actively contribute in energy dynamics during movement. The study used specialized equipment to cyclically load the foot, simulating the pressures experienced while walking or running. They recorded foot movements, stresses, muscle activity, and ultrasound pictures of the flexor digitorum brevis (FDB) muscle. The key findings include: 1. Active Muscle Role: It was revealed that the FDB muscle contracts isometrically during loading, allowing elastic energy to be stored not just in muscular tendons but also in the plantar aponeurosis. This demonstrates how intrinsic muscles play a significant part in the mechanical function of the foot. 2. Energy Storage and Release: Activating these muscles improves the foot's ability to store and release energy, enhancing its elasticity. This helps to lower the metabolic cost of walking and running. 3. Dynamic Modulation: According to the study, the central nervous system can actively alter the mechanical properties of the foot by stimulating the muscles. To sum up, the study emphasizes the importance of intrinsic foot muscles in altering the elastic characteristics of the foot and improving locomotor efficiency [74].



CHAPTER 3

EXPERIMENTS AND METHODS

The Human Subjects Ethics Committee of the Middle East Technical University approved the experimental protocol for this study under the protocol number 0423-ODTUIAEK-2022. All participants provided informed consent before participating. The experiments took place in the Motion Capture Laboratory, located at the Middle East Technical University, Modeling and Simulation Center (MODSIMMER).

3.1 Overview of Research Procedure and Data Collection

Ten right-handed adult males with no known neuro-musculoskeletal disorders, with a mean age of 29.2 ± 3.88 years, mean weight of 736.92 ± 82.50 N, mean height of 178.7 ± 4.96 cm and mean right foot length of 25.56 ± 1.21 cm, participated in the experiment. Before the experiment, the subjects completed a handedness inventory questionnaire (See Appendix B). Their gender, age, weight, height, preferred foot, and foot length were also recorded. After the experiment, the subjects were asked questions about their experience during the test, as well as the test's difficulty.

The experimental protocol for this study involved two steps. In the first step, participants were asked to stand upright on a custom-made force platform and pressure pad. Pressure pad was placed on top of the force plate. Their arms were relaxed at their sides, and their feet were shoulder-width apart. They completed six quiet stance trials, each lasting 90 seconds, these trials were repeated three times under two vision conditions: Eyes-Open (EO) and Eyes-Closed (EC), the sequence of eyes condition was randomized so that participants could not anticipate the next case. During the trials, the participants were instructed to stand upright with face straight ahead and

wear only socks on their feet (See Figure 3.1).

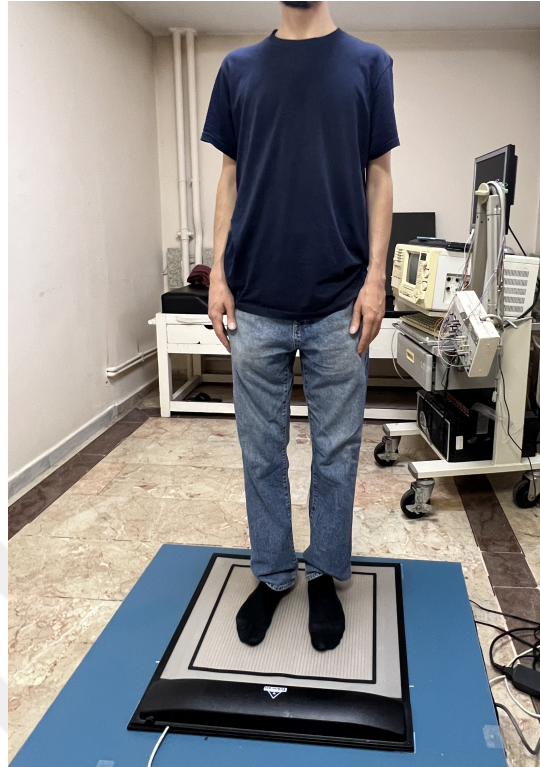


Figure 3.1: An example participant during the data collection on a pressure pad placed on top of a force plate

During the second step, an injection was administered to the feet to numb the soles. The injection period lasted from 15 to 20 minutes. Similar to the first step, the participants underwent six trials for the same duration of 90 seconds, and testing sensory conditions (Vision and Somatosensory). Each scenario was repeated three times, meaning that the participants underwent 12 trials in total. Rests between the trials were allowed based on the subject's needs.

During the experiment, force and moment signals were collected at a frequency of 100 Hz along the x, y, and z axes. A custom-made Bertec® FP1212-25 force plate measuring (1200×1200 mm) was used for this purpose.

The time signals for the COP were calculated in the Antero-posterior (AP) direction. Furthermore, pressure distribution data under each foot were collected at a frequency of 50 Hz using a Tekscan® Pressure Pad made up of over 2288 (44×52) sensing elements also known as "Sensels" (See Figure 3.2).

To ensure synchronized data collection a Matlab code was generated to trigger the data collection by the pressure-pad through its trigger box. Bertec amplifier was used in conjunction with the National Instrument “NI-6221-USB” multifunction USB data acquisition (DAQ) modules. Finally, the data was collected using the Matlab software.

Before scanning the feet of each participant, calibration was performed as per the manufacturer’s guidelines for MatScan, Tekscan®. The participant’s weight was measured using the Bertec® force platform and entered into the Matscan software. The software adjusts the sensing elements’ raw outputs so that the pressure pad’s total force is consistent with the participant’s weight. Next, Pressure distribution data was collected from the plantar surface of each foot. The force signal was obtained by multiplying the pressure with the sensing element’s area. All calculations and analyses were performed using MATLAB® Version 2018b from MathWorks, Inc. The results represent the average of three trials for each case to guarantee reliability.

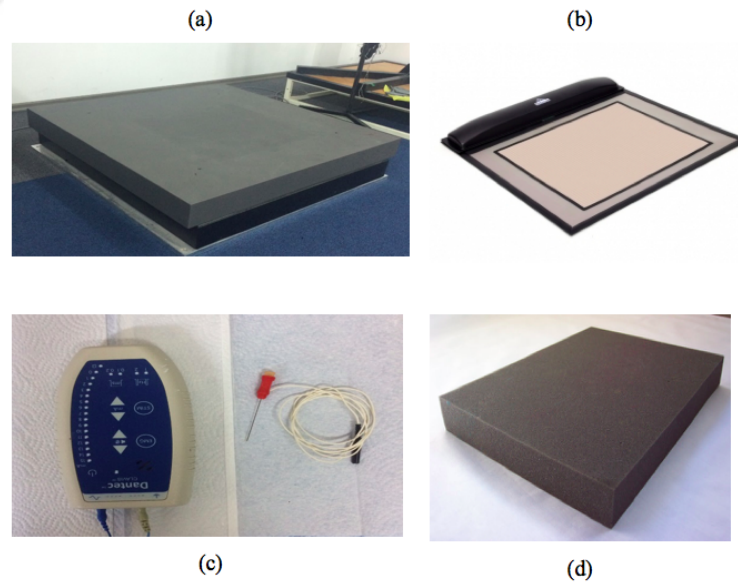


Figure 3.2: (a) Bertec Force plate, (b) Tekscan® Pressure Pad, (c) Peripheral nerve stimulator and peripheral nerve block needle, (d) Foam (compliant surface)

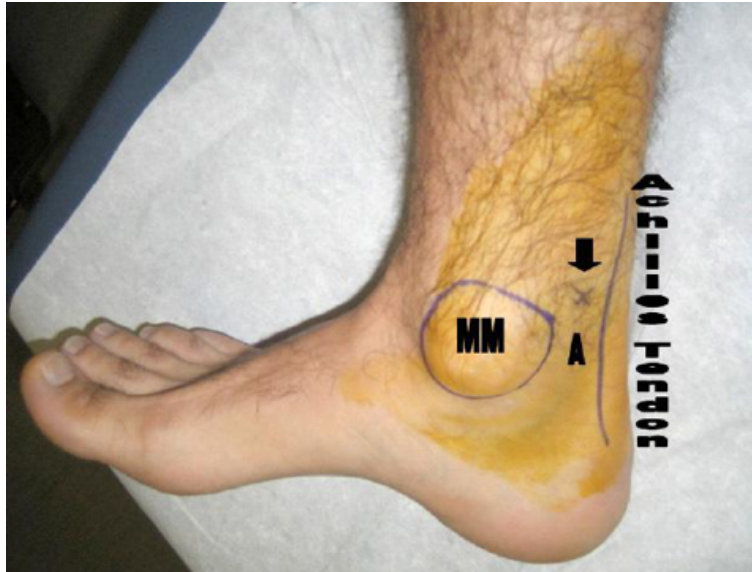


Figure 3.3: Tibial nerve block landmarks: The needle insertion site (arrow) is about 1 cm superior to the posterior tibial artery (A), with the medial malleolus (MM) on the left and the Achilles tendon on the right [6]

3.2 Blocking the Foot Somatosensory System

“Tibial nerve block at the level of ankle allows for rapid anesthesia of the plantar regions of the foot” [6]. The injection site was decided by following the guidance of the Achilles tendon and the medial malleolus. To find the injection point, a monopolar needle was inserted at a spot that lies midway between these landmarks (See Figure 3.3). Baseline recordings were taken prior to performing a tibial nerve block using a stimulator, a 25G monopolar needle, and a surface reference electrode. The block was conducted under sterile conditions with a short-acting local anesthetic agent. A 6 ml volume of 2% lidocaine was used to selectively block sensory fibers. The stimulus was applied with intensities ranging from 1-10 mA, and at a minimum stimulus intensity of 1 mA, the maximum motor response (toe flexion) was observed. Before the perineural injection, aspiration was done to ensure that there were no signs of intraneural injection (such as severe pain or other neuropathic symptoms like dysesthesia). The necessary conditions to manage complications related to the intervention were provided before the manipulation was performed. After that, a physical examination was conducted to assess sensory function at the end of the test period. The

expected duration of the nerve block was approximately between 1.5 and 2 hours. In Figure 3.4, three blocked regions associated with the tibial nerve are shown. These regions were colored pink, green, and dark turquoise, corresponding to the medial, lateral, and calcaneal branches, respectively. The foot sole was tested for sensitivity during the medical examination using a light touch test. The duration of the block and its impact were monitored by intermittently touching a piece of cotton every 10 minutes. Muscle testing was performed every 10 minutes to check for any motor troubles.

The risk in peripheral nerve blocks was reported as permanent neurological damage, which is 1.5 in 10000 [75]. A peripheral nerve stimulator, held at 0.5 mA, is used during the injection to minimize the risk of intraneural damage. Sterile conditions are also maintained to reduce the risk of local infection. Furthermore, precautions are taken to prevent allergic reactions and systemic toxicity, and the hospital is always prepared for such situations.

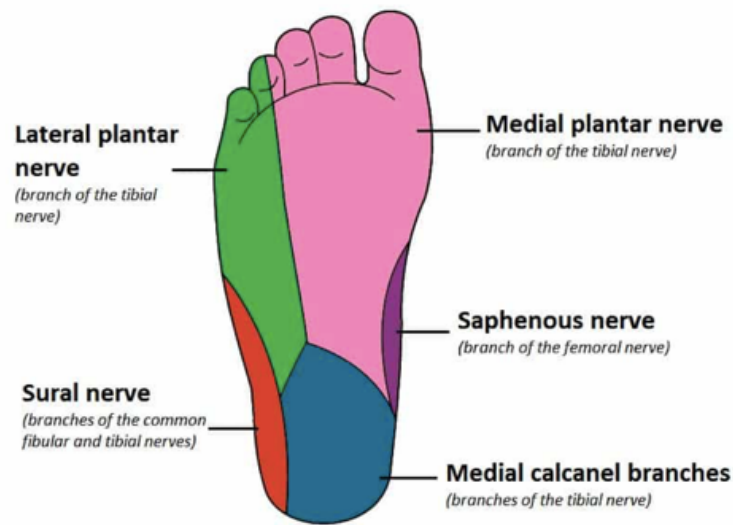


Figure 3.4: Sensory innervation of the foot sole [9]

3.3 Pre-Validation

Before analyzing our dataset, it is essential to perform data pre-validation to ensure its quality and reliability. It involves checking the data integrity, identifying inconsis-

tencies, and addressing missing values. By doing this, the accuracy of our following analyses will enhance and possible errors will be avoided. Ultimately, performing data pre-validation lays a strong foundation for obtaining meaningful insights. In this research, pre-validation is a preliminary assessment conducted on force plate and pressure pad devices to confirm the accuracy and consistency of data collected by these instruments. It involves verifying calibration, comparing readings, and identifying potential differences between the force plate and pressure pad measurements. This proactive validation step is essential for maintaining data integrity and reliability in biomechanical or related studies that rely on precise force and pressure measurements.

3.3.1 Similarity Between CoP_x Signals Obtained from Force Plate and Pressure Pad

Pre-validation process involves several steps:

1. **Calibration Verification:** Ensure that both force plate and pressure pad are properly calibrated to provide accurate measurements.
2. **Spatial Alignment:** Confirm that the force plate and pressure pad are spatially aligned, ensuring that the coordinate system for CoP_x is consistent between devices.
3. **Signal Synchronization:** Align time series data on both devices to synchronize CoP_x signals correctly. This avoids temporal inconsistencies in the records.
4. **Cross-Validation:** Collect simultaneous measurements from both force plate and pressure pad and compare CoP_x signals to identify any systematic differences or discrepancies.
5. **Data Cleaning:** Before analysis, removing outliers and addressing any noise or artifacts in the signals is essential to ensure high-quality data for comparison.
6. **Statistical Analysis:** Use statistical methods to measure the agreement between CoP_x signals from force plate and pressure pad. This may include correlation analysis or other relevant statistical techniques.

By implementing these pre-validation measures, the similarity between CoP_x signals obtained from different devices can be controlled, thereby promoting reliability and consistency in biomechanical analyses.

Figure 3.5 presents the CoP_x signals got from force plate ($COP_{x_{FP}}$) (blue) and pressure pad ($COP_{x_{PP}}$) (red).

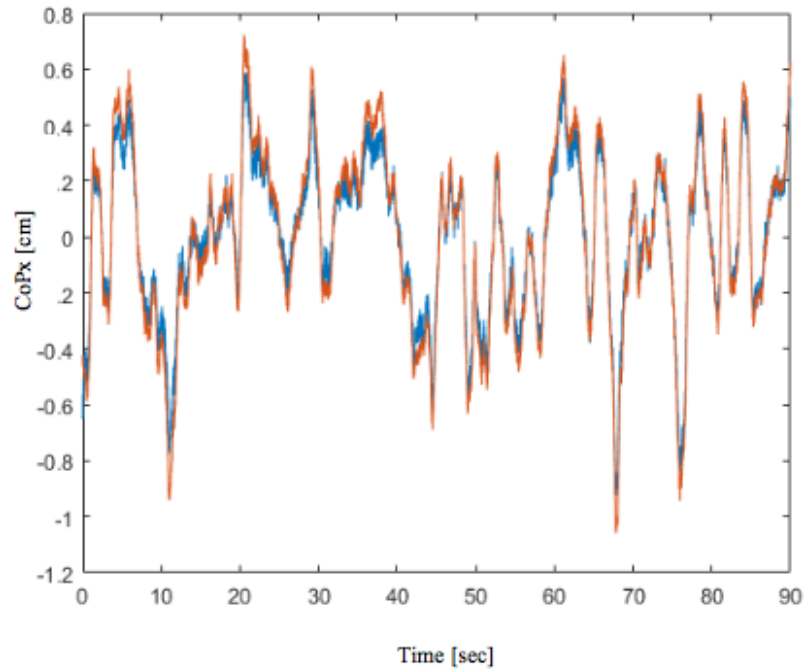


Figure 3.5: $CoPx$ signals got from force plate ($COP_{x_{FP}}$) (blue) and pressure pad ($COP_{x_{PP}}$) (red)

3.3.2 Similarity Between Weight Signals (F_z) Obtained from Force Plate and Pressure Pad

Both pressure pad and force plate are used to measure the vertical ground reaction force exerted by a subject, with the objective of quantifying the force exerted by the body's weight on the supporting surface. In order to make reliable comparisons in biomechanical analyses, it is important to consider the load distribution characteristics of each device and how they distribute the measured force across their sensing elements. By doing so, the similarity between the obtained weight estimations from pressure pad and force plate will enhance, making it easier to compare results (See

Appendix A).

3.4 Data Analysis

3.4.1 Time Domain Analysis

I used image processing analysis frame by frame to analyze the pressure pad data matrices. The pressure pad matrix has 44 rows and 52 columns; the dimensions of each pixel on the pressure pad are equal to 0.702579 cm^2 .

Three segments of the left foot (FL, ML, HL) and the exemplar frame of the left foot pressure data measured in kPa in matrix form are shown in Figure 3.6 and 3.7. This method is automated to monitor all the frames of a matrix in every trial. MATLAB is used to perform comprehensive data processing and analysis, as well as to track changes more quickly, easily and accurately.

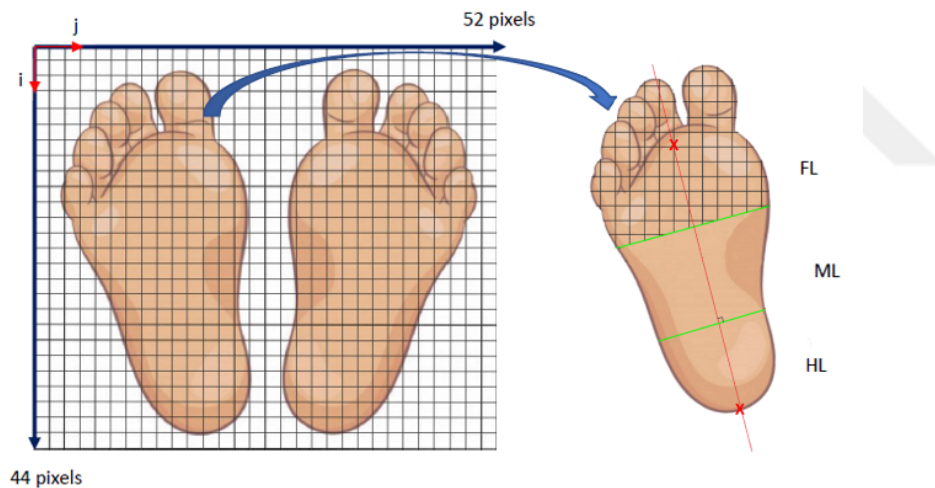


Figure 3.6: Three parts of the left foot (FL, ML, HL) and pressure pad matrix (44 rows and 52 columns) [10]

The pressure pad data is inputted into MATLAB software. Each set of data defines an experimental trial of the subject conducted under different conditions (S+, S-, EO, EC). To determine the position of the longitudinal axis, two points are selected for each foot. The first point is the metatarsal bone under the second finger, and the second point is the end of the calcaneus [65, 76, 77, 78]. The procedure is initiated

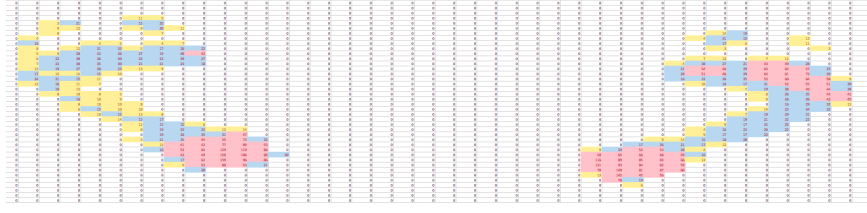


Figure 3.7: Pressure distribution under the foot based on the amount of force. (yellow color represent forces between 1 and 15 kPa, the blue color show forces between 15 and 40 kPa, and red ones represent forces larger than 40 kPa)

with the left foot and then progresses to the right (See Figure 3.8). This process can be repeated multiple times to enhance the reliability of the partitioning. Ultimately, the mean of all the defined longitudinal lines can be considered as the final outcome for the left foot. The procedure will be repeated for both feet, identifying three regions each.

(Fore part of left foot = FL, Mid part of left foot = ML, Hind part of left foot = HL and Fore part of right foot = FR, Mid part of right foot = MR, Hind part of right foot = HR) . The medial longitudinal axis of the foot was determined using a MATLAB code, which facilitated the division of each foot into three regions: Fore, Mid, and Hind. MATLAB's "dissection" code and the "create-mask" function were used to cut the toes (See Figure 3.9) [79].

Following the division of the feet into three regions, critical evaluation during image processing analysis, such as the determination of anteroposterior (AP) and mediolateral (ML) moment arms (cm), needed to be conducted. The following equations were utilized to estimate the weighted average moment arms, where AP (Anteroposterior) and ML (Mediolateral) represent axes, and f_{ij} denotes a random force variable with i rows and j columns (See Equations 3.1 and 3.2).

$$d_{FL}(AP) = \frac{\sum_i d_i \sum_j f_{ij}}{\sum_i \sum_j f_{ij}} \quad (3.1)$$

$$d_{FL}(ML) = \frac{\sum_j d_j \sum_i f_{ij}}{\sum_i \sum_j f_{ij}} \quad (3.2)$$

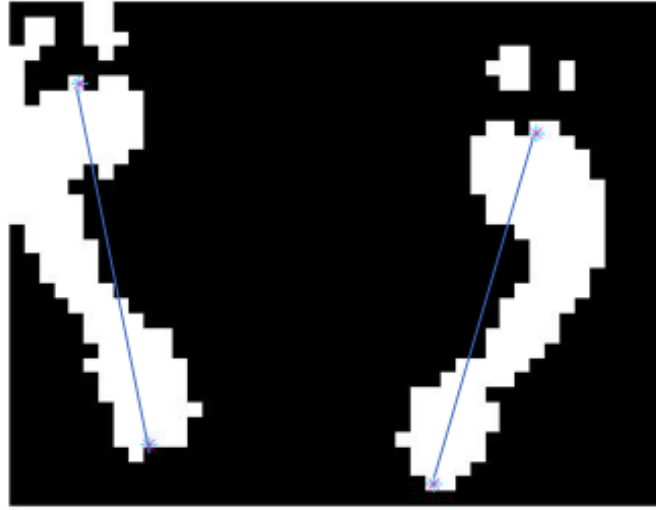


Figure 3.8: Frames of a pressure pad matrix for the left and right foot to define the foot

3.4.1.1 Time Domain Analysis for Pressure and Weight

Plantar pressure measurements are useful in evaluating the forces that the human body experiences while standing still or walking. By measuring the force distribution over each foot, detailed information can be obtained about the contact in each region, which is helpful for analysis.

This thesis aims to investigate the pressure distribution under the feet of ten healthy subjects in different vision and somatosensory conditions during quiet stance.

Statistical estimations like mean and variance of the pressure distribution of the foot regions in the time domain were computed using standard Matlab® software routines. The central tendency of the signals, which reflects the average time of the signals (mean value), was also calculated. Central tendency is a statistical measurement used to define a single score that represents the center of a distribution.

The weight carried by each region (Equation 3.3) is calculated by multiplying the pressure of that region by a sensor area of 0.0702579 cm^2 , representing the force resulting from a pressure change from kPa to N. Equation 3.3 is provided as an illus-

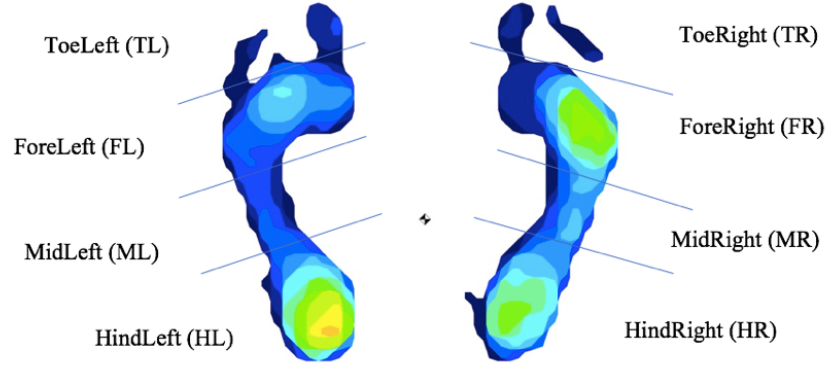


Figure 3.9: Three different regions of the right and left foot (Forefoot, Midfoot, Hind-foot)

tration for estimating the total weight carried by the front left region of the foot. The total weight (See Equation 3.4) is then obtained by summing the weight values of all six regions.

Each region's weight in percentage is also obtained by dividing the weight of the region by the total weight. This is known as the normalized weight (See Equation 3.5).

$$W_{FL} = \sum_i \sum_j W_{ij} \quad (3.3)$$

$$W_{Total} = W_{FL} + W_{ML} + W_{HL} + W_{FR} + W_{MR} + W_{HR} \quad (3.4)$$

$$W_{FL}\% = \frac{W_{FL}}{W_{Total}} \times 100 \quad (3.5)$$

To calculate the pressure on the each region of the foot, the Equation 3.6, which is written for the front left foot as an example was used (See Appendix A). Where i and j are the borders of front left foots in AP and ML directions, respectively and $\sum_i \sum_j 1_{ij} \times 0.702579$ is equal to regional area (cm^2).

$$P_{FL} = \frac{\sum_i \sum_j W_{ij}}{\sum_i \sum_j 1_{ij} \times 0.702579} \quad (3.6)$$

3.4.1.2 Time Domain Analysis for *CoPx*

There are several methods to evaluate postural control, but a widely used technique to assess postural control during movement is to measure the displacement of the CoP over a short time frame, usually ranging from seconds to minutes. CoP is the point on the surface where the vertical reaction vector is applied. It is calculated by taking a weighted average of all the forces applied on the surface [31]. The CoP moves due to the body's neuromuscular response to the center of mass position [31]. "The movement of the Center of Pressure (CoP) over time reflects the adjustments in posture and generates a time series that can be analyzed to determine cyclical patterns" [62]. CoP signal, measured from the horizontal plane on the force plate, consists of two components: CoP_x and CoP_y . CoP_x defines the location of CoP in the anteroposterior direction, while CoP_y represents its location in the mediolateral direction. The F_z , M_x , and M_y signals generated two time series data, namely, CoP_x and CoP_y . These series represent the movement of CoP in the x and y directions, respectively. The signals were recorded for 90 seconds at a sampling frequency of 100 Hz, resulting in 9000 samples for each signal. By analyzing the CoP_x time series, a clear understanding of the dynamics of postural sway in the sagittal plane can be obtained.

$$\text{CoP}_x(i) = -\frac{M_y(i)}{F_z(i)}, i = 1, \dots, 9000 \quad (3.7)$$

The mean value of the CoP_x signal was calculated by finding the arithmetic average of all sample values in the trial. The mean value of the CoP_x signal is a static measure that demonstrates the arbitrary position of the subject while standing on the platform during the test (see Figure 3.10 c), Therefore, it has been subtracted from the related signal. Signals utilized in further analyses have been detrended. The standard deviation and variance (Equation 3.8) of the CoP_x signal were also calculated. The standard error/mean ratio is a measure of the dispersion of the distribution, and in our case, it is a measure of variability or amount of postural sway [80].

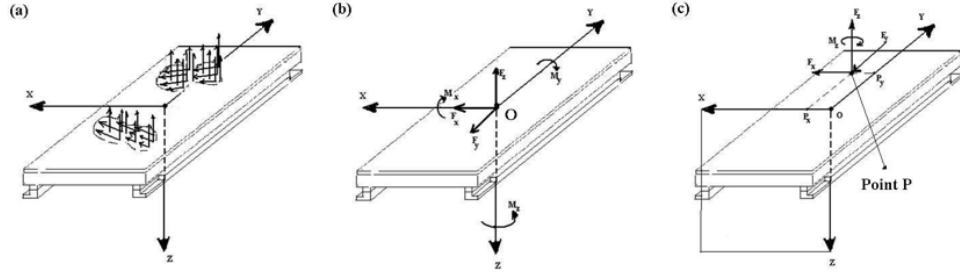


Figure 3.10: (a) Distributed forces at the foot-base, (b) The resultant force-couple system of the distributed forces at the mid-point (origin) of the platform (Point O), (c) The resultant force-couple system has been moved to Point P (CoP), where M_x and M_y are zero [11, 12]

$$\sigma_X^2 = \frac{1}{n-1} \sum_{i=1}^n (X_i - \bar{X})^2, i = 1, \dots, 9000 \quad (3.8)$$

$$R_{xx}(\tau) = \frac{1}{T} \int_0^T X(t)X(t+\tau)dt \quad (3.9)$$

3.4.2 Frequency Domain Analysis

To understand balance maintenance in individuals, it is essential to analyze power across various frequencies. This study employed frequency domain analysis to examine the frequency characteristics of signals. Frequency domain measures were computed using Matlab® Fast Fourier Transform (FFT) routines. Prior to executing the FFT, the time series data were linearly detrended. The frequency domain measures were then adjusted based on methodologies proposed by Prieto et al., Maurer et al., and as outlined in the book by Bendat and Piersol [42, 81, 82].

The Power Spectral Density (PSD) function ($G_{xx}(f)$) was applied to analyze the power distribution of signals within the frequency spectrum (Equation 3.10). PSD estimates were calculated using Matlab's built-in function (pwelch). Furthermore, the autocorrelation function was utilized to derive $G_{xx}(f)$.

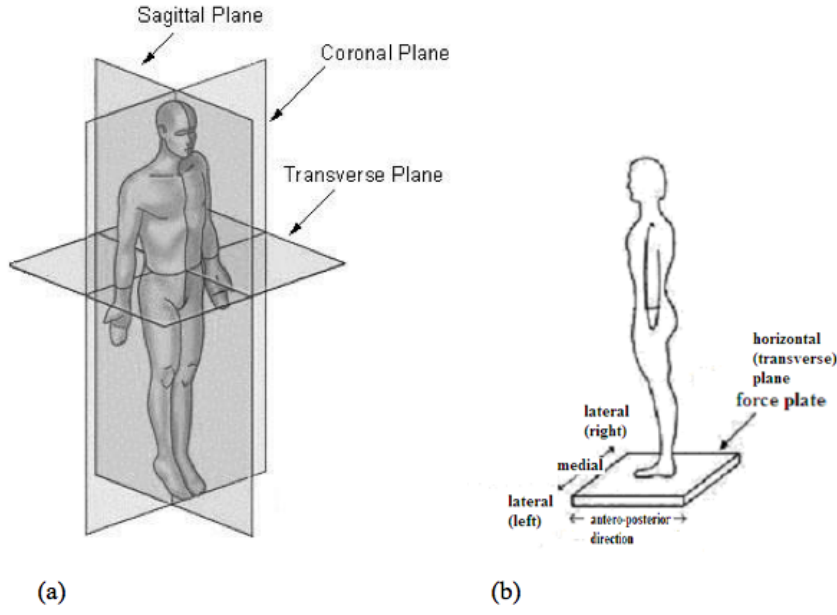


Figure 3.11: (a) Anatomical planes described for the human body, (b) Sagittal plane on force plate [13]

$$G_{xx}(f) = 2 \int_{-\infty}^{+\infty} R_{xx}(\tau) e^{-j2\pi f\tau} d\tau \quad (3.10)$$

The spectral moments, represented by μ_k , were computed for values of k ranging from 0 to 3 using Equation 3.11.

$$\mu_k = \sum_{i=1}^m (i \times \Delta f)^k G_{xx}(i \times \Delta f) \quad (3.11)$$

The formula for calculating frequency resolution (Δf) is dependent on the duration of data collection (T). The formula for Nyquist frequency (m) is half of the sampling rate divided by Δf . The total power can be calculated as the integrated area of the PSD when $k = 0$ in Equation 3.11, and is equal to the variance of the time series $X(t)$ [82].

$$\text{POWER} = \mu_0 = \sum_{i=1}^m G_{xx}(i \times \Delta f) \quad (3.12)$$

The 50% and 90% power frequency represent the frequencies at which 50% and 90% of the total power is observed, respectively.

$$\sum_{i=1}^v G_{xx}(i \times \Delta f) \geq 0.50 \times \mu_0 \quad (3.13)$$

$$\sum_{i=1}^v G_{xx}(i \times \Delta f) \geq 0.90 \times \mu_0 \quad (3.14)$$

Where v is the smallest integer for which the Equation 3.13 and 3.14 are valid.

The centroidal frequency (**CFREQ**) is the frequency where the spectral mass is concentrated, calculated by taking the square root of the second to the zeroth spectral moment ratio as shown in Equation 3.15.

$$CFREQ = \sqrt{\frac{\mu_2}{\mu_0}} \quad (3.15)$$

The frequency dispersion (**FD**) is a unitless metric that quantifies the variability in the frequency content of the Power Spectral Density (PSD). which is calculated using Equation 3.16. For a pure sinusoid, the frequency dispersion is zero, whereas it increases to a maximum of one with spectral bandwidth [20].

$$FD = \sqrt{1 - \frac{\mu_1^2}{\mu_0 \times \mu_2}} \quad (3.16)$$

3.5 Statistical Analysis

An analysis called repeated measures ANOVA can be conducted with a significance level of 0.05 to study the effects of vision and the somatosensory system. There are four random variables within subjects, including Eyes (Open, Closed), Somatosensory (Somatosensory intact, Somatosensory loss), Foot (Left and Right), and Regions (Frontfoot, Midfoot, hindfoot). To analyze the statistical data, Matlab® offers a post-hoc statistical analysis method called "Multi-comparisons" that can be used to compare each dimension of ANOVA. The metrics that are included in the statistical

analysis are Root Mean Square (RMS), Variance, 50% and 90% power, CFREQ, FD, and the mean of overall and regional pressure.

In my thesis, repeated measure ANOVA with a significance level of 0.05 were used to compare independent variables. This statistical approach is used to examine within-subject variations, which increased the sensitivity of this thesis analysis. Repeated measurements of the same individuals were conducted to identify any trends, changes, or effects over time. This approach facilitated the acquisition of a robust understanding of the impact of the investigated variables. Furthermore, it enabled a comprehensive examination of changes in CoP_x and pressure distribution across repeated measurements, thereby providing insights into temporal and frequency-related patterns.



CHAPTER 4

EXPERIMENTAL RESULTS

This chapter presents the results of the experiments. It includes time series and frequency domain analyses, followed by the statistical analysis results gathered from the pressure pad data.

4.1 Time Series Analysis Results

The statistical analysis in time domain includes various metrics such as RMS, Variance, RMS and Variance of Velocity. It also involves mean of overall pressure, regional pressure, and regional weight in percent.

4.1.1 Time Domain Analysis Results for Overall Pressure

The overall pressure of the subjects varies around 32 kilo pascals (kPa); this is significantly different in the eyes condition (Eyes open : 32.324 ± 1.281 kPa, Eyes closed: 32.090 ± 1.288 kPa, $p=0.004$); overall pressure was higher in eyes open condition.

4.1.2 Time Domain Analysis Results for Overall CoP_x

Figure 4.1 displays the pressure pad data of the same subject in the same trial, demonstrating the subject's movements in the anteroposterior axis by tracking the CoP_x over time under various eyes and somatosensory conditions. The magnitude of fluctuations increases as the subject closes his eyes and when foot sole is anesthetized.

Repeated measure ANOVA shows that RMS varies by an average of 0.5 cm around

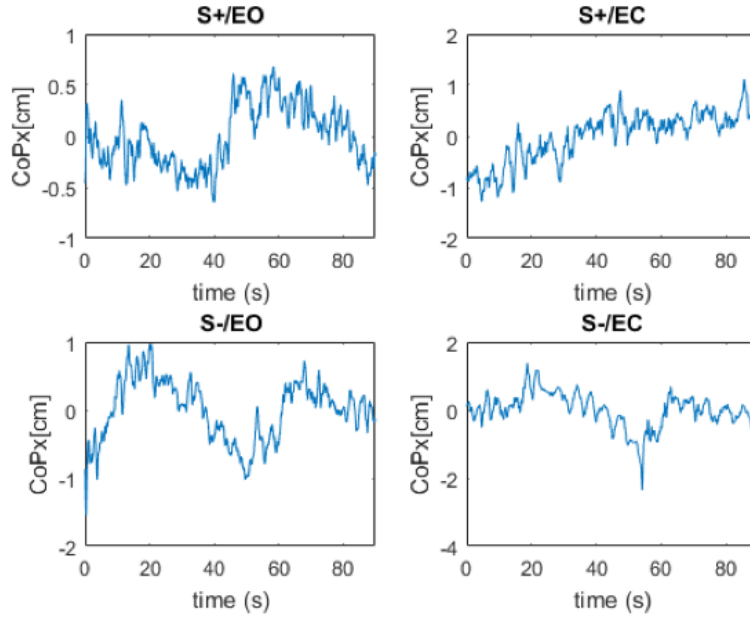


Figure 4.1: Anteroposterior fluctuations (overall CoP_x , detrended) versus Time, before (S+) and after injection(S-), in Eyes Open (EO) and Eyes Closed (EC) conditions

zero according to the Table 4.1. Significant differences were found in RMS, due to eyes condition, at eyes closed condition sway magnitude was significantly higher. Somatosensory shows no significant differences, indicating that it may be a more robust sensor in comparison to vision. In terms of velocity, the eye effects are still significant. Velocity of CoP_x movements increased significantly after the eyes got closed.

4.1.3 Time Domain Analysis Results for Percent Weight and Regional Pressure

In this study, three distinct regions were analyzed on each foot, namely FL, FR, ML, MR, HL, and HR, providing detailed insights into pressure changes within each region individually. This information can be used to compare the data for each foot and both feet. Significant differences in percent weight were observed both across feet and regions, with higher percentages recorded in the right foot, whereas it was highest at the hint, lowest at the mid foot (See Table 4.2).

As expected, in regional pressure, there are significant differences in mean pressure of

Table 4.1: Mean and Standard Error of RMS and Velocity RMS for overall CoP_x , (DV stands for Dependent Variable and IV stands for Independent Variable)

	DV	RMS (cm)	Velocity RMS (cm/sec)
IV			
Eyes	EO	0.331±0.022	0.020±0.002
	EC	0.371±0.013	0.023±0.002
p value		0.037*	0.009*

foot regions, further multi comparisons (post hoc analysis) revealed differential differences in somatosensory versus regions relation such that mid regions pressure was the lowest as expected (See Table 4.3 and Figure 4.2). Fore foot pressure decreased by intervening somatosensory loss through local anesthesia while hind foot pressure increased.

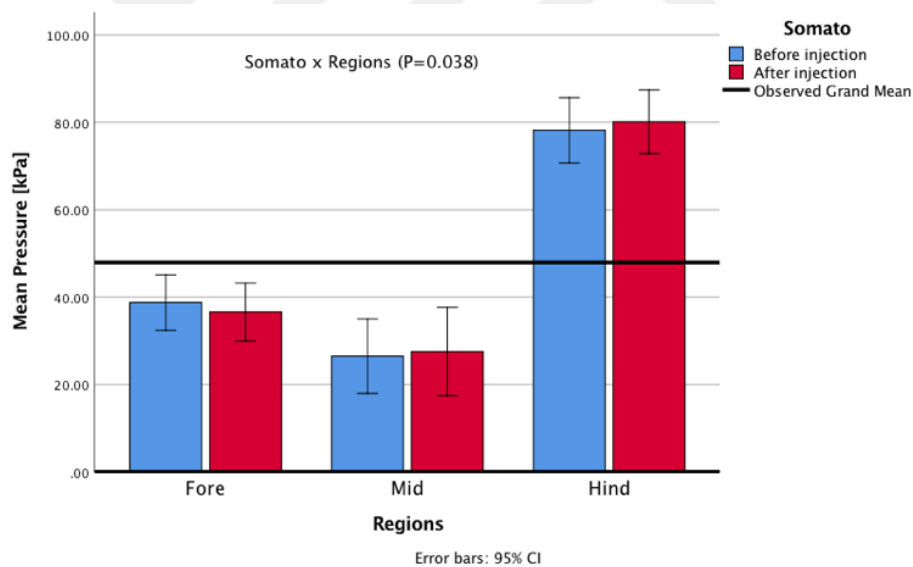


Figure 4.2: Regional pressure multi comparisons results (Somato × Regions)

Figures 4.3 and 4.4 show that subjects put more weight and pressure on their right foot than their left foot (See Table 4.2).

Note that foot percent weight were divided by six (three regions under left foot, three regions under right foot) so the mean percent weight of two feet and three regions are

Table 4.2: Mean and Standard Error for regional pressure and percent weight, (DV stands for Dependent Variable and IV stands for Independent Variable)

	DV	Regional Pressure (kPa)	Weight (%)
IV			
Foot	Left	44.687±1.482	0.157±0.002
	Right	51.193±3.069	0.176±0.002
p value		0.017*	0.005*
Regions	Fore	37.659±2.673	0.200±0.10
	Mid	27.004±3.919	0.055±0.008
	Hind	79.158±2.999	0.246±0.012
p value		0.000*	0.000*

around 0.166 (See Table 4.2).

4.1.4 Time Domain Analysis Results for Regional CoP_x

From a kinetic perspective, the statistical analysis of various foot areas enables the observation of foot lateralization and inter-regional cooperation. Kinematically speaking, analyses of CoP_x need to be conducted over regions, which cannot be achieved through an overall CoP_x analysis.

In this section, a statistical analysis was conducted to determine the significant influence of various factors on various numeric measurements, including time and frequency metrics. These factors encompass vision conditions (EO, EC), somatosensory conditions (S+, S-), cooperation among foot regions (front foot, mid foot, hind foot), and lateralization of the left and right foot.

The results for RMS and RMS Velocity for regional CoP_x are presented in Table 4.4. The RMS of regional CoP_x measured approximately 0.05 cm, whereas the overall RMS was 0.5 cm, representing nearly one-tenth of the latter. The RMS of regional CoP_x was determined to be 0.066±0.008 for the forefoot, 0.040±0.005 for the mid-foot, and 0.043±0.004 for the hindfoot, with a corresponding p-value of 0.08.

Table 4.3: Mean and Standard Error of regional pressure for Somato $\times$ Region, (DV stands for Dependent Variable and IV stands for Independent Variable)

Somato	Region	Regional Pressure (kPa)
Before injection (S+)	Fore	38.731 $\pm$ 2.685
	Mid	26.475 $\pm$ 3.605
	Hind	78.174 $\pm$ 3.166
After injection (S-)	Fore	36.586 $\pm$ 2.813
	Mid	27.534 $\pm$ 4.277
	Hind	80.142 $\pm$ 3.086
p value		0.038*

Like overall CoP_x , RMS values for eyes condition differed significantly in regional CoP_x ($p = 0.032$), eyes closed condition was found to be larger than eyes open. Further, significant differences in somato $\times$ regions ($p = 0.01$) were observed; somatosensory effects on the regions were found to be as an increase in the fore foot while as a decrease in the mid foot when local anesthetic was applied. (See Figure 4.5).

As shown in Table 4.4, the velocity RMS was found to be significantly larger at left foot ($p = 0.003$) and was significantly smaller at hind region ($p \leq 0.001$).

4.2 Frequency Domain Analysis Results

Fourier analysis is a mathematical technique used to transform a signal from its original domain, which could be time or space, into a representation in the frequency domain and vice versa. In this section, the frequency behavior of signals before and after injection will be investigated in both EO and EC conditions.

The statistical analysis in frequency domain includes various metrics such as 50% and 90% POWER, CFREQ, and FREQD for the CoP_x . PSD functions given in Figure 4.7 demonstrate that peak magnitudes increase under the effect of somatosensory loss while high frequency band of power is observed at eyes closed condition.

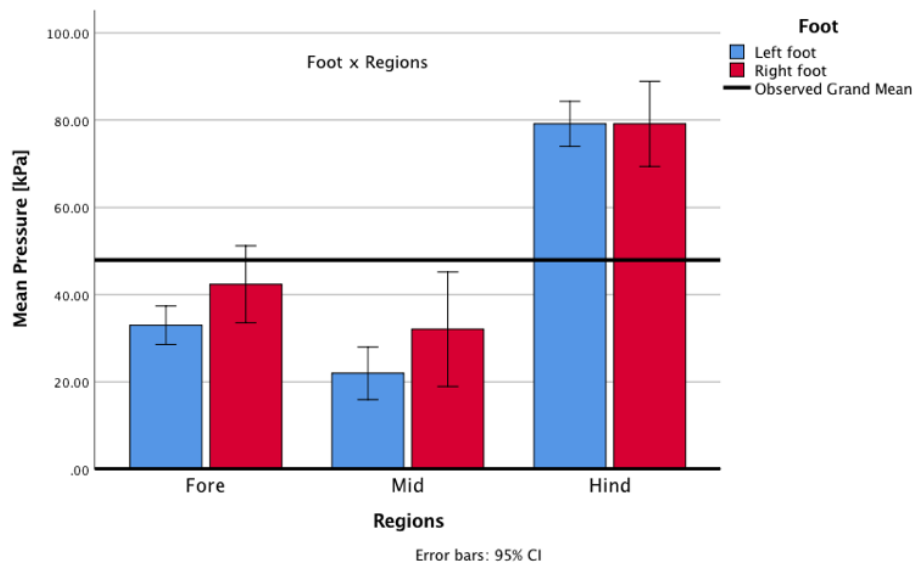


Figure 4.3: Regional pressure multi comparisons results (Foot $\times$ Regions)

4.2.1 Frequency Domain Analysis Results for Overall CoP_x

When eyes are closed, there are significant differences in comparison to eyes open (Table 4.5). 90% power cut-off frequency and frequency dispersion measure decreased significantly at eyes closed condition

4.2.2 Frequency Domain Analysis Results for Regional CoP_x

The centroidal frequency of overall CoP_x was approximately 0.5 Hz, and it did not vary significantly in different sensory conditions. However, there was a significant difference in the regional CFREQ between the left and right foot, as described in Table 4.6. It was found to be higher in the left foot. On the other hand, CFREQ at mid foot was the highest, whereas, in hind foot it was the lowest.

In the overall CoP_x , 50% power was around 0.25 Hz, and no significant results were shown for different sensory factors. In the regional CoP_x , since the regions got smaller, the frequency would be higher, and at 50% power, it was around 2 Hz. During the analysis, it was observed that there were significant differences in the foot ($p=0.000$), eyes ($p=0.020$), region ($p=0.000$), and foot $\times$ region ($p=0.001$). 50% Power was significantly higher at the left foot.

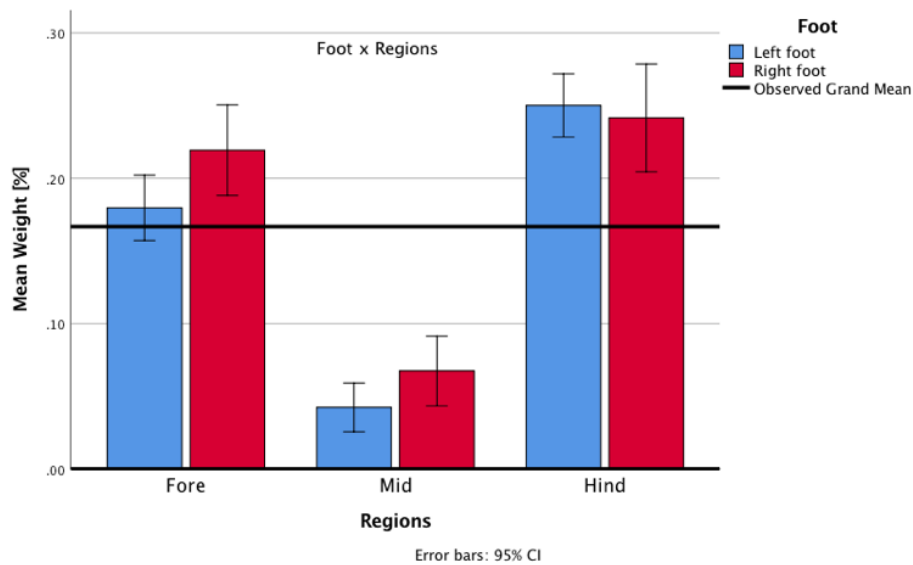


Figure 4.4: Regional weight in percent multi comparisons results (Foot $\times$ Regions)

50% and 90% Power was found to be significantly smaller when eyes are closed, and were found to be the highest at mid foot while were the smallest at the hind foot. Although, mid foot carried the highest 50% Power cut off frequency (around $3.39Hz$) the front foot cut off frequency was considerably distant than hind foot ($2.06Hz$ compared to $0.42Hz$, respectively).

FD was significantly larger in eyes open condition. Multi comparisons revealed higher order effects at Eyes $\times$ Somato $\times$ Region ($p=0.026$). found higher 50% Power frequency was found at fore and mid foot in eyes open condition when somatosensory was intact, however this difference in eyes condition at 50% Power frequency disappeared after injection (See Figures 4.8 and 4.9). After injection 50% Power cut off frequency increased at fore and mid foot regions in eyes closed condition.

Table 4.4: Mean and Standard Error of RMS and RMS Velocity for regional *CoPx*,
(DV stands for Dependent Variable and IV stands for Independent Variable)

	DV	RMS (cm)	RMS Velocity (cm/sec)
IV			
Eyes	EO	0.047±0.004	-
	EC	0.052±0.004	-
p value		0.032*	-
Foot	Left	-	0.017±0.002
	Right	-	0.012±0.001
p value		-	0.003*
Regions	Fore	-	0.019±0.002
	Mid	-	0.020±0.003
	Hind	-	0.005±0.000
p value		-	0.000*

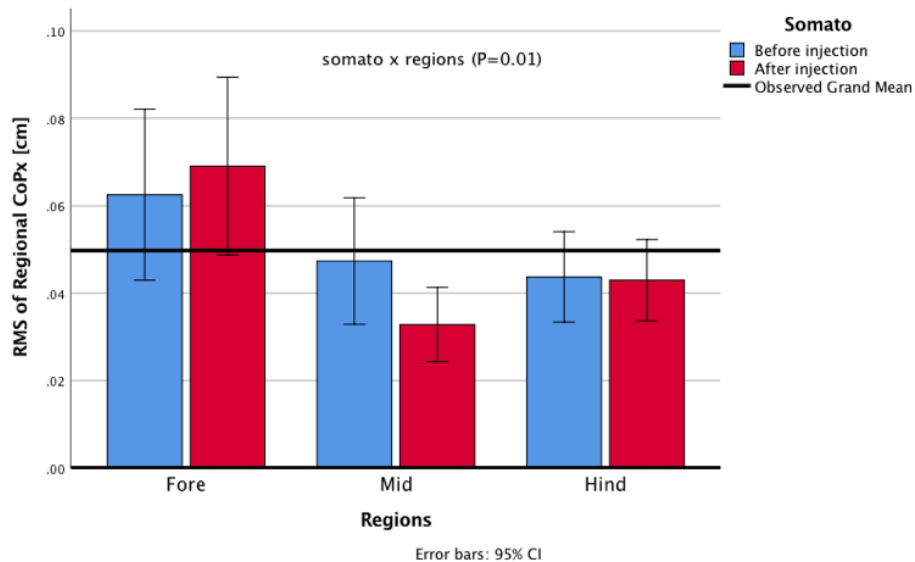


Figure 4.5: RMS of regional CoPx multi comparisons results (Somato × Regions)

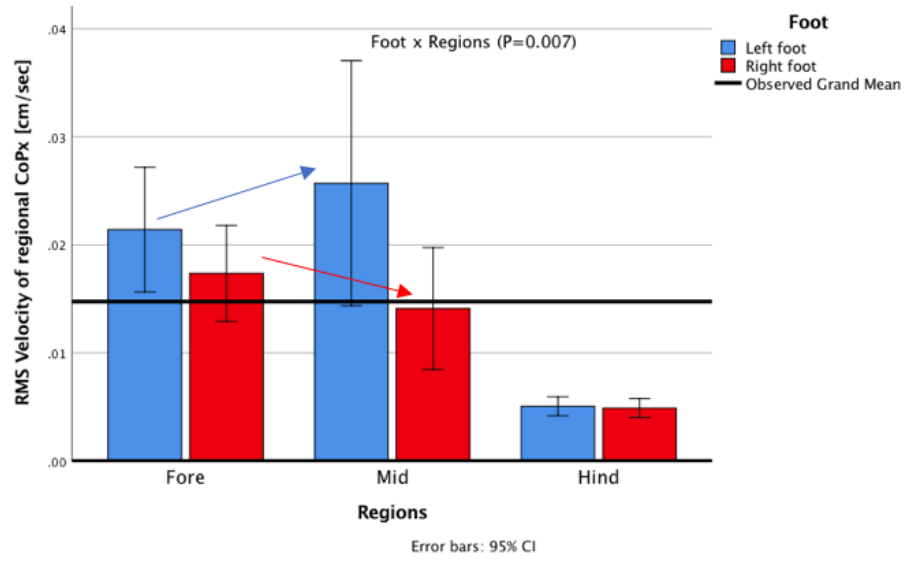


Figure 4.6: Multi comparisons results of regional velocity RMS (Foot $\times$ Regions). RMS velocity at the front region was lower when compared to mid foot at the left foot while was higher at the right foot (see the arrows)

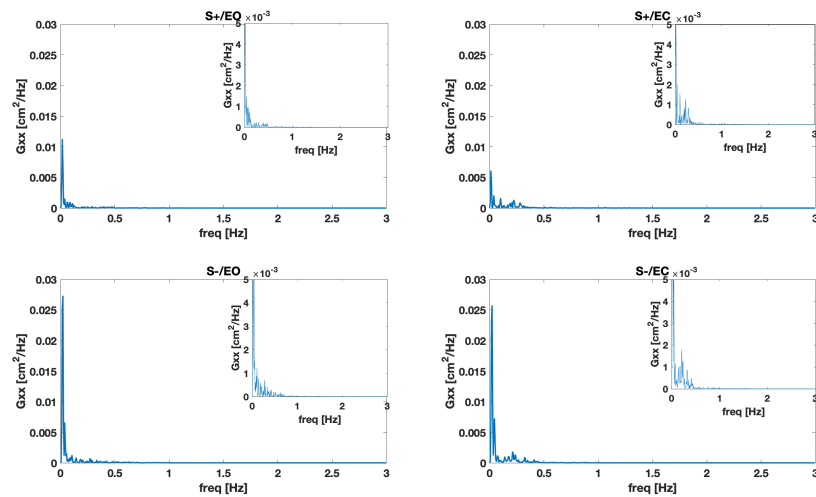


Figure 4.7: PSD functions, Gxx(f), before and after injection in EO/EC conditions

Table 4.5: Mean and Standard Error of 90% power and FD for overall *CoPx*, (DV stands for Dependent Variable and IV stands for Independent Variable)

	DV	90% Power (Hz)	FD
IV			
Eyes	EO	5.006±0.468	0.717±0.010
	EC	3.858±0.272	0.702±0.014
p value		0.011*	0.020*

Table 4.6: Mean and Standard Error of CFREQ , 50% and 90% Power and FD for regional *CoPx*, (DV stands for Dependent Variable and IV stands for Independent Variable)

	DV	CFREQ (Hz)	50% Power (Hz)	90% Power (Hz)	FD
IV					
Eyes	EO	-	2.100±0.137	9.539±0.192	0.708±0.008
	EC	-	1.812±0.156	9.276±0.203	0.659±0.012
p value		-	0.020*	0.017*	0.028*
Foot	Left	0.624±0.015	2.327±0.138	-	-
	Right	0.597±0.017	1.585±0.157	-	-
p value		0.041*	0.000*	-	-
Regions	Fore	0.594±0.012	2.063±0.167	10.017±0.131	-
	Mid	0.645±0.013	3.385±0.373	10.484±0.126	-
	Hind	0.592±0.021	0.420±0.045	7.721±0.381	-
p value		0.021*	0.000*	0.000*	-

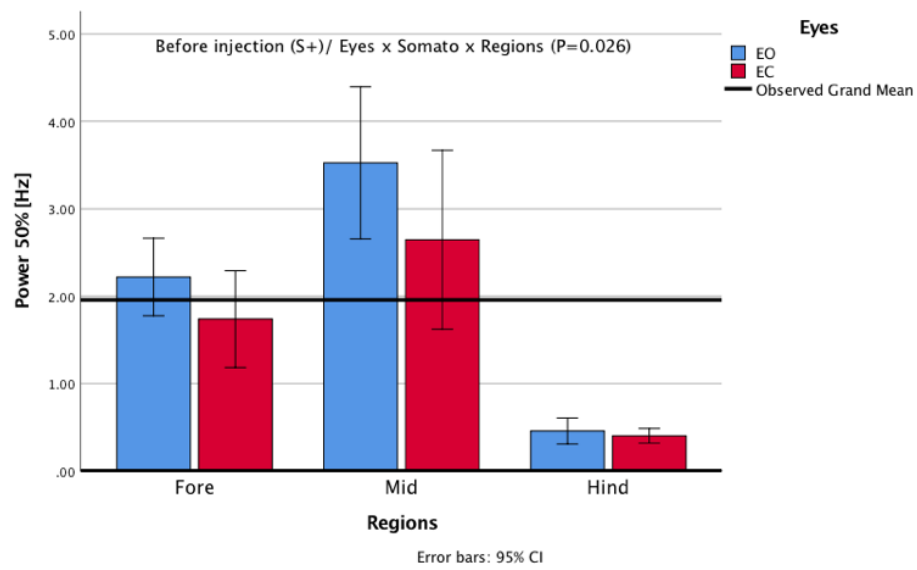


Figure 4.8: Before injection, Regional PWR50 (Eyes $\times$ Somato $\times$ Regions)

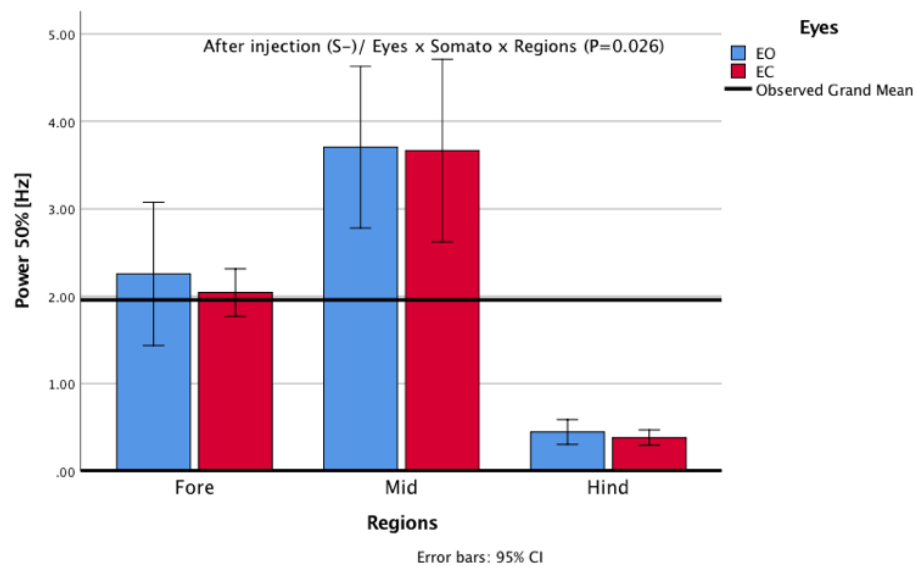


Figure 4.9: After injection, Regional PWR50 (Eyes $\times$ Somato $\times$ Regions)



CHAPTER 5

DISCUSSION

In the classical postural literature, postural research depends on the function of orderly sequenced body limbs starting from the foot and ending at the head. In this sense, these orderly structured body limbs, one on top of the other, represent an open kinematic chain, where the dynamics and control of the open kinematic chain mainly depend on the control of the joints starting from the ankle [66, 42, 55, 56, 60, 61, 83, 84, 85]. The inadequacy of the postural control approach starting by the ankle joint has been shown in this thesis.

5.1 Role of the Foot in the Overall Postural Dynamics

Adjusting the Weight Distribution:

The foot acts as a dual-functioning organ to control the human erect posture. The first function of the foot in the control of the posture is supporting the body (Load bearing), while the second function is the role of actuating during standing and locomotion [72, 44, 68].

The foot plays a crucial role in providing the necessary support, stability, and mobility for human standing and locomotion. The plantar aponeurosis plays a crucial role in supporting the arch of the foot and transmitting forces during weight-bearing activities. The foot has three arches, namely two longitudinal (medial and lateral) and one anterior transverse arch. These arches are formed by the tarsal and metatarsal bones, which are supported by the ligaments and tendons in the foot. The arched shape of the foot functions like a spring, supporting the body's weight and absorbing

the shock of the movement. The flexibility of the foot's arches facilitates everyday movements, such as walking and running [1, 18, 44]. The foot accomplishes these tasks in two ways. Firstly, the bones, ligaments, tendons, and plantar fascia tension in the foot work together like a truss structure, reinforcing the arch of the foot and providing the necessary strength and stability to bear the body's weight and maintain balance. Secondly, the arch of the foot functions like a beam, evenly distributing the body's weight across the foot and cushioning the impact of movement [39]. Donatelli (1985) mentioned that muscles don't need to be active to support a fully loaded foot at rest [48]. However, a new research conducted by Nozaki et al. revealed that plantar aponeurosis gets stiffen through the activation of intrinsic muscles [71]. Further, Kelly et al. showed standing is actively controlled through activation of the intrinsic foot muscles counteracting against the arch deformation. When the foot was loaded by an external load, the intrinsic muscles contracted to reduce the lengthening of the longitudinal arch and to increase the height of it, where both serve for the stabilization of the arch by restoring elastic energy [72, 73, 74]. Thus, the arch in a static foot is maintained through the passive support of ligaments, bones and the tension in the plantar fascia building the buttress effect (truss action). When the plantar aponeurosis becomes taut with toe extension, the arch rises which is known as "toe-extending arch-raising effect". This mechanism was named as the "windlass effect" by Hicks [17, 15]. Hicks in his researches also emphasized the importance of the beam action of the foot observed by measuring the bending strain occurred in the metatarsal and tarsal bones. He noted that in a static stance position when the plantar fascia is intact, the truss model of the foot takes up about 60% of the load bearing function, while the metatarsal beam action takes up approximately 25%. On the contrary, when the plantar fascia is dissected, the contribution of the truss model of the foot in load bearing function reduces to 13%, while the beam action takes up approximately 55%. Thus, he concluded that plantar fascia tension is critical in the truss action of the foot in load bearing at quiet stance. The absence of the plantar fascia in supporting the arch during load bearing, role of the foot in quiet stance is dominated by its beam function [39].

Indeed, the highest pressure and the smallest RMS CoP_x velocity, CFREQ, 50% and 90% power cutoff frequencies were found at the hind foot while the smallest pressure

and the highest CFREQ, 50% and 90% power cutoff frequencies were observed at the mid foot, whereas the front foot presented mean pressure and CFREQ, 50% and 90% power cutoff frequencies in between the hind and mid foot. In this thesis, the participants were asked to perform quiet stance trials. As the plantar fascia was intact, participants were expected to utilise truss and beam functioning of the foot in tandem during postural sway such that. During the truss functioning of the foot, the hind region is likely to be responsible from the support action of the load bearing while the mid and front regions are supposed to present the motile characteristics of the foot via the windlass mechanism [13, 39, 46, 44, 47, 65, 68]. RMS CoP_x velocity, CFREQ, 50% and 90% power cutoff frequencies were found to be higher at mid and front regions pointing to the motile characteristic of the foot in standing, while the support action of the foot was manifested by the higher mean pressure observed at the hind region.

The duality proposed by Hicks in functioning foot during standing viz. truss versus beam action constitute behaviour of the foot during quiet stance. The truss function of the foot is composed of two mechanisms, arch spring and windlass mechanism. Arch spring mechanism supplies flexibility to the foot by restoring energy, while the foot gains rigidity through toe extension by rising the arch. Although, foot establishes functional analogies with mechanics of structures it is difficult to differentiate between these functions in quiet stance, rather it becomes easier to observe these different mechanisms of the foot during large amplitude human movements; e.g., walking, running, jumping, etc. [46, 47, 73, 74].

Besides, truss function of the foot during quiet standing can be thought as two separate beams connected by a hinge joint where the internal reaction is a zero moment. Then the foot presents a truss like function through the hinge joint depending on where the joint is placed (See Appendix C).

Generating the Regional Moment Arms:

It has been observed that the magnitudes of moment arms in the foot regions decreased by approximately one-tenth compared to the overall CoP_x measured during postural sway. This finding inherently suggests that the overall postural sway is constructed from its smaller components, namely, the supportive and functional regions

of both the right and left foot [31, 45].

In this thesis, the beam action of the foot was provoked by creating somatosensory loss at the plantar face of the foot [35, 64, 86]. Somatosensory loss created an increase in the mean pressure of the hind foot while the mean pressure of the front foot decreased. Further, RMS of CoP_x in the front region increased but decreased in the mid region. Hicks in his 1955 paper presented the experimental results behalf of the beam function of the foot when the plantar fascia has been dissected. Local anesthetics were employed to induce somatosensory loss, aiming to deteriorate plantar fascia function by intervening somatosensory feedback [35, 36, 37, 64]. As a result, a significant decrease in RMS CoP_x was observed in the mid region. This result was interpreted in terms of reduced activity observed at the medial longitudinal functional arch of the foot due to the disruption of the buttress effect [72]. Missing buttress effect might have caused an increase at RMS CoP_x in the front region while a decrease in the mean pressure at the front foot. Thus, creating somatosensory loss dissociated the coherence observed at the mid and front regions of the foot functioning together as observed in its truss action [46, 36].

5.2 Role of the Foot Functions in Sensory Motor Performance of the Human Posture

It has been shown that when the foot contact area is decreased, the weight of the ankle strategy in the postural control gets smaller and the body control switches to more proximal joint modes; e.g. from the ankle strategy to the hip control [52, 53, 56, 60, 87]. Indeed, Winter et al. (1996) studied contribution of the ankle and hip strategies to postural control during quiet stance [31]. They decomposed the overall CoP signal into two parts. The first part represents the contribution of right foot and left foot resultant forces normalized by the total weight then multiplied by the corresponding moment arms measured through two different force plates. They revealed the hip strategy dependence (load/unload mechanism) of the partitioning the total weight to right and left reaction forces, which has been found to be sensitive to the configuration of the standing. Parallel to the method proposed by Winter et al., the different roles of the foot regions in load bearing capacity of the foot could have been deciphered by

using a decomposition model [10, 13, 46]. The resultant forces at the front and hind foot regions normalized to the total weight with their corresponding moment arms (regional CoP_x) could constitute the main hypothesis in the control of the posture during quiet stance at anterior-posterior direction.

Significant differences were observed at the generated regional moment arms due to the cross-talk between somatosensory and visual systems. Eyes condition had different effects on mid foot due to the somatosensory condition. Higher 50% Power frequency was found at the fore and midfoot in the eyes-open condition when somatosensory was intact. However, this difference in the eyes condition at 50% Power frequency disappeared after injection (See Figures 4.8 and 4.9). These results were interpreted in terms of the visual effects on the truss function of the foot influenced by somatosensory feedback. When somatosensory feedback was compromised due to the local anesthetic injection, the truss function of the foot reverted to beam action, no longer being modulated by visual effects. Following the injection, the 50% Power cutoff frequency increased at the fore and midfoot regions in the eyes-closed condition.



CHAPTER 6

CONCLUSION AND FUTURE WORK

This thesis demonstrates the inadequacy in postural control mechanisms, excluding the sensory motor role of the foot. During quiet stance, the average pressures in the hindfoot, forefoot, and midfoot regions were approximately 80, 40, and 25 kPa, respectively. Somatosensory loss decreased the mean pressure in the forefoot region while increasing RMS CoP_x . On the other hand, the velocity RMS was significantly lower in the hindfoot region. CFREQ, along with the 50% and 90% power cut-off frequencies, were significantly different, with the forefoot and midfoot regions showing the highest values and the hindfoot the lowest. This finding in the frequency domain posits to the quasi dynamical differences about the role of the foot regions during the control of quiet stance: the forefoot plays an active role, while the hindfoot provides passive support.

Eyes condition had different effects on mid foot due to the somatosensory condition. When somatosensory function was intact, the forefoot and midfoot showed higher 50% power frequencies in the eyes-open condition. However, this difference in 50% power frequency due to eye condition disappeared after the injection.

RMS velocity was significantly larger in the left foot, particularly in the forefoot compared to the midfoot, while it was lower in the right foot. Frequency domain analysis showed that the CFREQ of Regional CoP_x was significantly higher in the left foot.

The original findings of this thesis are warranted to be verified by using a mathematical model in the frequency domain. The active role of the foot in adjusting ankle control torque will be modeled based on regional moment arms multiplied by the

resultant regional ground reaction forces. Additionally, the regional characteristics during postural sway, observed in both time and frequency domains, will be studied during provoked stance by incorporating electrophysiological measurements.



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APPENDICES





Appendix A

SIMILARITY BETWEEN OBTAINED WEIGHT SIGNALS (F_z) FROM PRESSURE PAD VERSUS FORCE PLATE

There was a consistent decrease in weight signal measured in Newton using the Pressure Pad over time as compared to the Force Plate. This issue was observed in half of the subjects (as shown in Figure A.1), while the other half had a similar trend to the force plate (as shown in Figure A.2). The downward trend in the weight signal led us to find a new approach to solve this issue. Therefore, research was conducted using the raw form of the data from the pressure pad. Instead of using the weight data (F_z) measured in Newtons, the raw form of the data was processed. In the information file of each trial (Excel file), raw data is referred to as a raw sum. The raw sum is the measurement performed by the instrument at a given time.

The raw data was analyzed and the results were satisfactory. As shown in Figure A.3, the downward trend issue was successfully solved. However, there is still a constant difference between the weight calculated from the raw data and the weight obtained from the force plate. This discrepancy is mainly due to three factors. Firstly, the force plate signal is measured in Newtons while the pressure pad signal has no units of measurement. Secondly, the pressure pad has 44×52 sensors while the force plate only has four sensors. Lastly, the frequency response bands of both devices differ. To address the issue of different signal physical dimensions, a constant number was determined by following the procedure given below, for the pressure and weight of each subject by using the saturation point of each trial. By multiplying the raw data of each subject with this constant value, we obtained results that were comparable and closer to the weight data obtained from the force plate, as shown in Figure A.4.

To match the signals from the two sensors, we assumed the force plate as our reliable

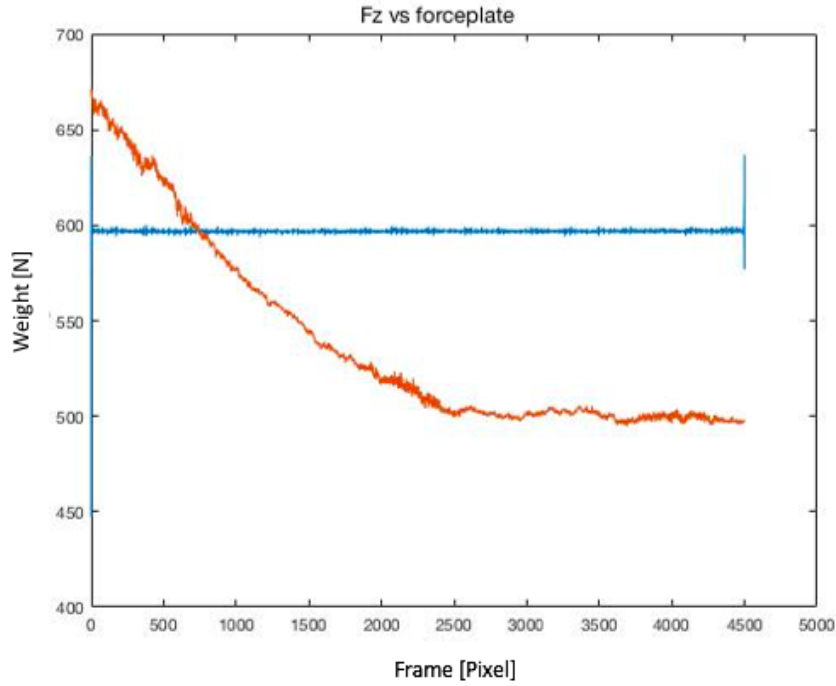


Figure A.1: Weight signal from pressure pad F_z (red) versus weight signal from Force plate (blue), Collected from subject 1, trial 1 (EO, S+)

reference and followed the below steps to synchronize the signals:

1. The saturation point (Pound per Square Inch) of each subject was converted to a value between 0 and 255 by dividing it into 256 increments (PSI/RS). This was done using the information from an Excel sheet saved from a pressure pad application (See Figure A.5).
2. The initial result is multiplied by 0.45 (pound to kilograms) and 9.81 to convert the weight to Newton , respectively ($N/in^2 \times RS$).
3. Divide the result from step two by 6.45 ($in^2 = 6.45 cm^2$) to get the pressure in N/cm^2 . (To convert to Kilo Pascal, multiply the result by 10).
4. To calculate the weight on the sensel after the step two, the weight needs to be multiplied by the area of the pixel, where 1 pixel equals to $0.702579 cm^2$. To view an example and the corresponding conversion process, please refer to Table A.1.

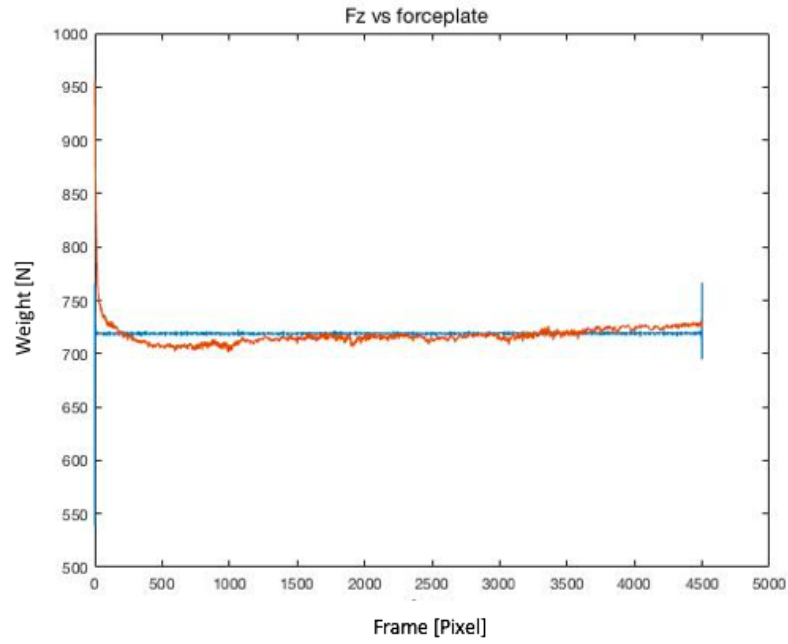


Figure A.2: Weight signal from pressure pad F_z (red) versus weight signal from Force plate (blue), Collected from subject 2, trial 1 (EO, S+)

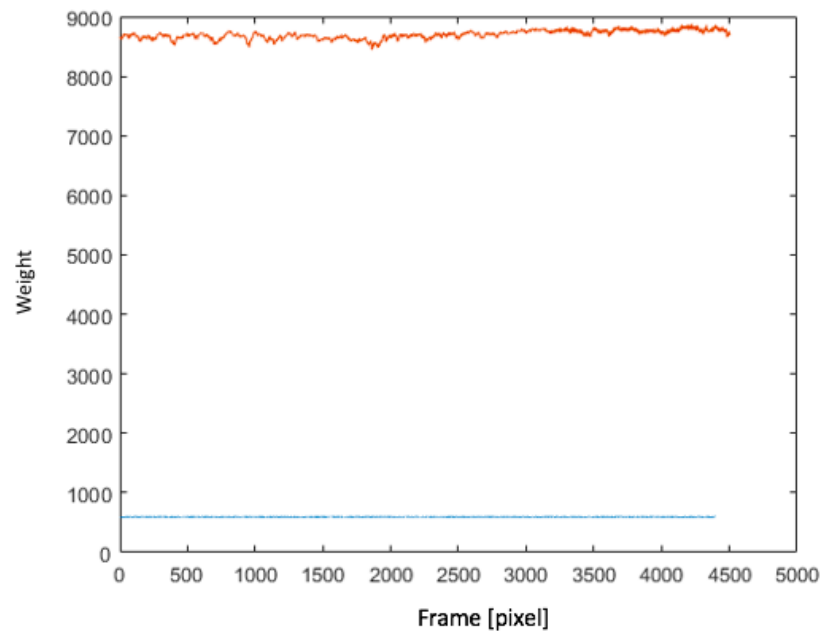


Figure A.3: Weight signal from raw data of pressure pad F_z (red) versus weight signal from Force plate [N] (blue), collected from subject 1. The downward trend which appears at Figure A.1 no more exists

Signal from Force plate [N] (blue), weight signal from pressure pad (red), weight signal from pressure pad (F_z) after getting the process corrected from subject 1, trial 1 (EQ. S1)

TIME 11-03-16 2:47:06.16PM
SATURATION_PRESSURE 58.0154 PSI
CALIBRATION_POINT_1 158.715 (Pounds) 6406 (Raw Sum) 160 (Number of Loaded Cells)
CALIBRATION_MODE_1 Step
CALIBRATION_INFO C:\Users\NaghshAppleStore\Desktop\maryam\thesis flash\my_subjects\2.ehsan zabardast\movie\zabae01.fsx
SENSITIVITY Default
MAP_INDEX 0
SUBMAP 0
Patient ID:

Figure A.5: Calibration information of each trial, subject's weight in Pounds and the number of loaded cells

Table A.1

Steps:	Example:
1. Read the saturation point from excel file.	58.0184 (<i>PSI</i>)
2. Convert the saturation point (Pound per Square Inch) to a value between 0 and 255 by dividing it into 256 increments.	$58.0184 / 255$ $= 0.2275$ (<i>PSI/RS</i>)
3. Multiply the result by 0.45 (pound to kilograms) and 9.81 to convert the weight to Newton.	$0.2275 \times 0.45 \times 9.81$ $= 1.004$ ($(N/in^2 \times RS)$)
4. Divide the result from step three by 6.45 ($Inch^2 = 6.45cm$) to get the pressure in N/cm^2 which is equal to 10 kPa.	$1.004 / 6.45$ $= 0.155$ (N/cm^2)
5. Multiply the result by 10 to get the pressure in kPa.	0.155×10 $= 1.557$ (<i>kPa</i>)



Appendix B

INVENTORY SURVEY

Before the test, participants completed an inventory to determine their dominant foot and hand. Only subjects who are right-handed and have a dominant right foot were included in the test. (See Figure B.1)

Edinburgh Handedness Inventory

Please indicate your preference in the use of your hands in the following activities *by putting a + in the appropriate column*. Where the preference is so strong that you would never try to use the other hand unless you were forced to, *put ++*. If in any case you are really indifferent *put + in both columns*.

Some of the activities require both hands. In these cases the party of the task, or object, for which hand preference is wanted is indicated in brackets. Please try to answer all the questions, and only leave a blank if you have no experience at all of the object or task.

		LEFT	RIGHT
1	Writing		
2	Drawing		
3	Throwing		
4	Scissors		
5	Toothbrush		
6	Knife (without fork)		
7	Spoon		
8	Broom (upper hand)		
9	Striking a Match (match)		
10	Opening box (lid)		
i	What foot do you prefer to kick with?		
ii	Which eye do you use when using only one?		

L.Q. Leave this spaces blank **Decile**

Retyped from Oldfield, R.C. (1971) The assessment and analysis of handedness: The Edinburgh Inventory. *Neuropsychologia*, 9:97-113.

Figure B.1: Inventory Survey



Appendix C

INTERNAL REACTIONS

Let's assume that there exists a beam with uniform load distribution (w denotes the load density), fixed at points A and B by two hinge supports (See Figure C.1). This is known as two hinged arch and is statically indeterminate. The geometry of the two-hinged arch, where internal moments are zero (referred to as zero moment) is determined.

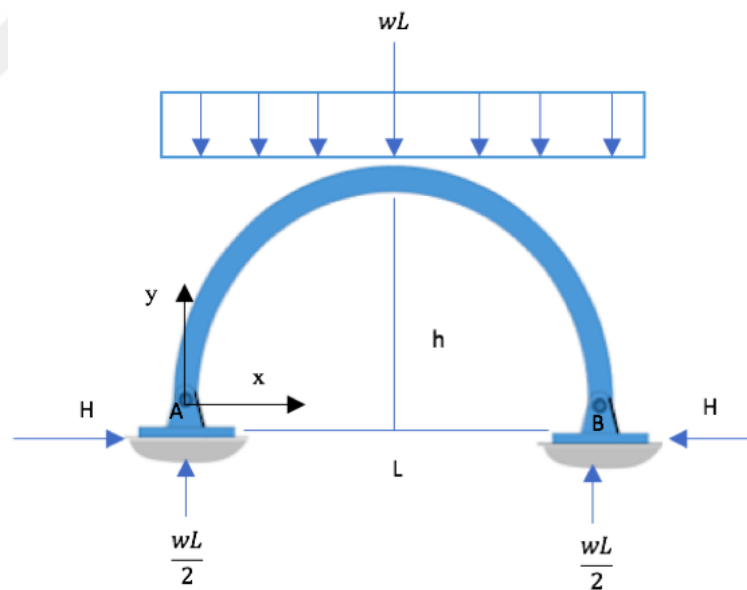


Figure C.1: A uniform load distribution over the beam AB with the length, L and maximum height, h is presented. H represents the horizontal support reaction forces at point A and B, while $\frac{wL}{2}$ stands for the vertical support reaction forces

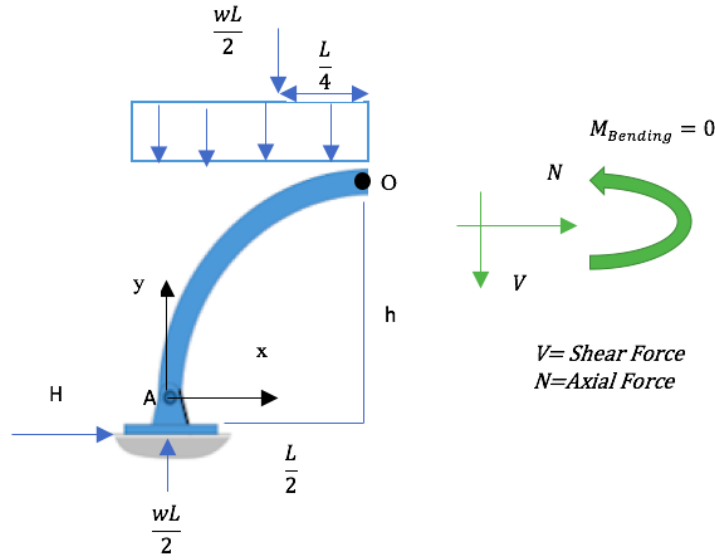


Figure C.2: Presentation of internal reactions of N , V , $M_{bending}$. Point O describes the cross section at $x=L/2$

$$\begin{aligned}
 \sum M_O &= 0 \\
 \frac{wL}{2} \frac{L}{4} - \frac{wL}{2} \frac{L}{2} + Hh &= 0 \\
 Hh &= \left(\frac{wL^2}{4} - \frac{wL^2}{8} \right) = \frac{wL^2}{8} \\
 H &= \frac{wL^2}{8h}
 \end{aligned} \tag{C.1}$$

Horizontal reaction force can also be found by using principle of least work. As a second step, the geometry of the arch will be described by using the moment equilibrium equation, M_O written below for an arbitrary distance, x and height, y (See Equation C.2).

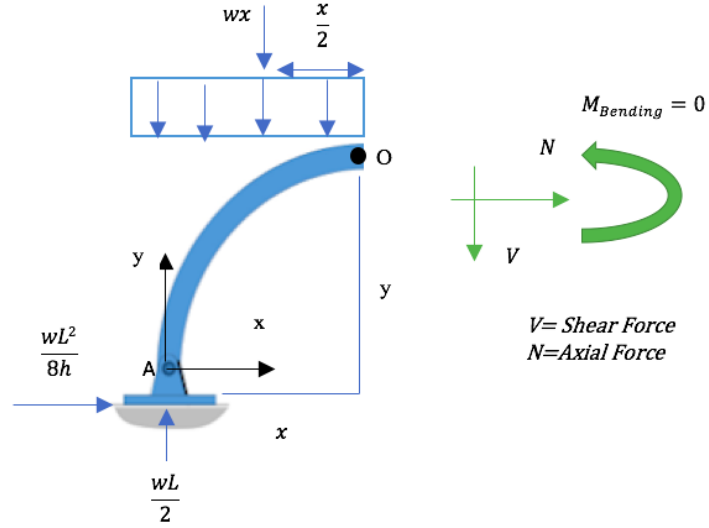


Figure C.3: Presentation of internal reactions of N , V , $M_{bending}$. Point O describes the cross section at an arbitrary x

$$\begin{aligned}
 \sum M_O &= 0 \\
 \frac{wx^2}{2} - \frac{wL}{2}x + \frac{wL^2}{8h}y &= 0 \\
 \frac{wL^2}{8h}y &= \frac{wL}{2}x - \frac{wx^2}{2} \\
 \frac{L^2}{4h}y &= Lx - x^2 \\
 y &= \frac{4h}{L^2}[Lx - x^2]
 \end{aligned} \tag{C.2}$$

The beam and truss function of the foot is presented, where the arch depresses through compression and rises by toe dorsiflexion, respectively (See Figure C.4). The expression found at Equation C.2 as $y(x)$ is the zero moment geometrical description of two hinged arch. Figure C.5 presents the zero moment function scaled and over-plotted to the medial longitudinal arch of the left foot (aspect ratios are identical). Talocalcaneonavicular joint is expected to pass through the zero moment function. However, the curvature of the medial longitudinal arch is not a parabola that is to match with the zero moment function. Further, the foot is not a two hinged arch, rather it can be assumed as a hinged arch at the calcaneus and a roller at the MTP joint connected by the plantar fascia to the calcaneus [13].

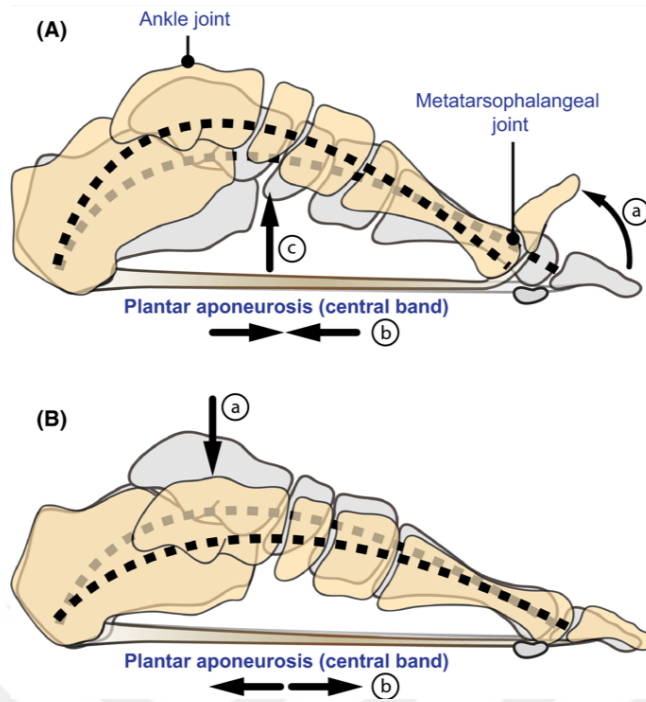


Figure C.4: Medial longitudinal arch of the left foot (medial view) is presented [14]. A: The truss function and the windlass mechanism, a) dorsiflexion of the big toe, b) Contraction of the plantar fascia, c) arch rise (toe extension, arch rise), B: The beam function, a) the body load, b) plantar fascia stretch [15]

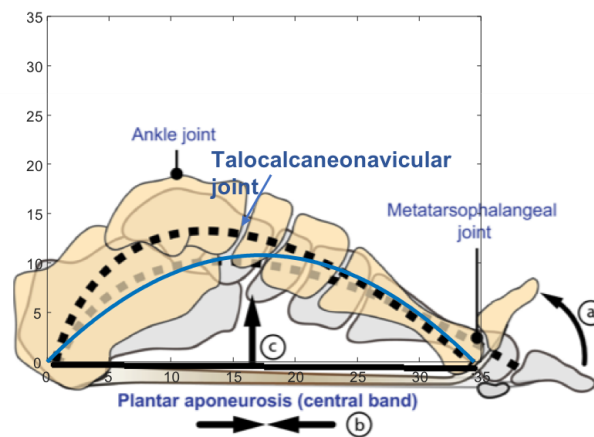


Figure C.5: Medial longitudinal arch of the left foot (medial view, light gray dashed line) is presented with the overplot of the zero moment function derived in Equation C.2 (blue bold curve in the figure). Width and height are presented in *cm*. Foot skeleton diagram is borrowed from [14]

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PUBLICATIONS

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Pourreza E., Hassanpour M., Yaradanakul N. B., Cengiz B. C., Alkan O., Gürses S. "Role of the Foot in Quiet Stance." 20th International Sport Science Congress, Antalya, Turkey, 2022

Usual K. A., Hassanpour M., Pourreza E., Demiroz F. A., Hohenberger A., Cakır M. P. "Eye Movement Analysis in the Simon Task." International Symposium on Brain and Cognitive Science, Istanbul, Turkey, 2018

Hassanpour M., Cengiz B. C., Gürses S., "Foot Somatosensory Information Contributes To Quiet Stance" S. 8th National Biomechanics Congress, Ankara, Turkey, 2016