Exploring the Effect of Ankle Braces on Foot Posture

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Abstract

Foot posture is an important characteristic that can affect kinematics, plantar loading, and injury risk. Arch height is one common aspect of foot posture, and it is estimated that about 60% of the population has normal arches while 40% of the population is either pes planus or pes cavus. It is important to be able to accurately and reliably assess foot posture characteristics in order to propose interventions that could prevent injuries due to abnormal foot alignment. However, despite multiple classification metrics, many of the devices that are commonly used for foot posture measurements are not economically feasible for smaller clinics or research labs. Therefore, the first purpose of this study was to develop an affordable device to measure different foot posture characteristics. The Foot Posture Measurement System was developed and can measure total foot length, truncated foot length, foot width, dorsum height, and navicular height. This system was shown to have good to excellent validity (ICC = 0.908-0.994) and repeatability (ICC = 0.867-0.996) when compared to a 3D scanner. This device was then used in the second portion of this study, which evaluated the effects of ankle braces on plantar loading patterns in individuals with different foot postures. Contact area, peak force, force-time integral, and center of pressure were evaluated during a walk, run, and cut while the participant was unbraced, wearing a lace-up stabilizer brace, and wearing a semi-rigid brace. It was demonstrated that arch height did affect the maximum plantar forces during all tasks ($p=0.001$-0.047), as hypothesized based on previous studies. Additionally, this study found that ankle braces affected contact area ($p=0.001$-0.0014), maximum force ($p<0.001$ – $p=0.043$), and force-time integral ($p<0.001$ – $p=0.015$) during the walk, run, and cut. This is a novel finding and points to the potential for an impact of ankle braces on plantar loading during athletic activities, independent of foot type.

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General Audience Abstract

Foot posture is an important characteristic that can affect daily life and contribute to the risk of injury. Arch height is one common aspect of foot posture, and it is estimated that about 60% of the population has normal arches while 40% of the population is either high arched or low arched/flat footed. It is important to be able to accurately and reliably assess foot posture characteristics in order to propose interventions that could prevent injuries due to abnormal foot alignment. However, despite multiple classification metrics, many of the devices that are commonly used for foot posture measurements are not economically feasible for smaller clinics or research labs. Therefore, the first purpose of this study was to develop an affordable device to measure different foot posture characteristics. The Foot Posture Measurement System was developed and can measure five different length, width, and height characteristics of the foot. This system was shown to be valid when compared to a 3D scanner and repeatable between days. This device was then used in the second portion of this study, which evaluated the effects of ankle braces on individuals with different foot postures. Four different force and pressure variables were examined within the foot during a walk, run, and cut while the participant was unbraced, wearing a lace-up stabilizer brace, and wearing a semi-rigid brace. It was shown that arch height did alter plantar loading measures during all tasks, as hypothesized based on previous studies. Additionally, this study found that ankle braces affected all variables during the walk, run, and cut. This is a novel finding and points to the potential for an impact of ankle braces on plantar loading during athletic activities, independent of foot type.

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List of Abbreviations

AHI	Arch Height Index
AP	Anterior-Posterior
AHIMS	Arch Height Index Measurement System
BMI	Body Mass Index
CMC	Coefficient of Multiple Correlation
COP	Center of Pressure
ERLLP	Exercise Related Lower Leg Pain
FMPS	Foot Posture Measurement System
FPI	Foot Posture Index
FTI	Force-Time Integral
ICC	Interclass Correlation Coefficient
ML	Medial-Lateral
MLA	Medial Longitudinal Arch
MTSS	Medial Tibial Stress Syndrome
NICA	Normalized Insole Contact Area
PTI	Pressure-Time Integral
RM ANOVA	Repeated Measures Analysis of Variance
ROM	Range of Motion
SPM	Statistical Parametric Mapping

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Introduction

Aims

Arch height is an aspect of foot posture which classifies the foot as being pes cavus (high arched), pes rectus (normal), or pes planus (flat footed/low arched). Approximately 60% of the population have normal feet, 20% have high arches, and 20% have low arches [1]. Rearfoot angle is another component of foot posture, where the rearfoot can be categorized as hindfoot valgus, neutral, or hindfoot varus. These two measures are often related, as many individuals with high arches are hindfoot varus while many flat-footed individuals are hindfoot valgus.

While there are a variety of foot posture classification methods, they often require measurements of different dimensions of the foot. This is commonly done with a custom-built foot measurement device or a 3D scanner, however these technologies are not well established for research use and are often too expensive for small scale research projects. The first goal of this study was therefore to develop and validate an affordable foot measurement device that is easily accessible and can reliably be used in future research.

Measuring and classifying foot posture is important as there is a significant relationship between abnormal arch heights and lower extremity injuries [2]. More specifically, individuals with a pes planus foot have increased mobility compared to those with a pes cavus foot. As a result, studies have found that pes planus individuals are more prone to soft tissue injuries due to increased joint rotation [3]. On the other hand, pes cavus individuals are more susceptible to bone injuries, foot injuries, and lateral leg injuries as a result of increased plantar pressures on the lateral side of the foot [3]–[5]. Those with pes cavus feet have a more lateral center of pressure in addition to higher pressures in the heel and lateral forefoot [6]. Conversely, those with pes planus feet have a more medial center of pressure and have higher contact areas, pressures, and forces in the medial midfoot, middle forefoot, and hallux [6], [7].

Understanding the mechanics of the foot and ankle and the causes of injury is crucial for injury prevention. Common preventative measures include in-shoe orthoses, ankle taping, and ankle braces [8]. Braces and taping have been shown to reduce both the occurrence and the severity of ankle sprains, with braces being more effective [8], [9]. Even though bracing has been shown to be effective for preventing ankle sprains, there has been no research examining their effect on various foot postures and the resulting plantar loads. If ankle braces can be shown to effectively correct abnormal plantar loads in the second portion of this study, they can serve as an affordable and easily accessible preventative measure for the injuries that may result due to abnormal arch height.

Therefore, the objectives of this study are: 1) Develop an affordable system to accurately measure foot posture characteristics, and 2) Determine the effects of ankle braces on the plantar loading patterns of individuals with different arch heights.

Specific Aim 1: **Develop an affordable system to accurately measure foot posture characteristics.**

Hypothesis: An inexpensive system can be made to accurately and repeatably measure the foot posture characteristics of foot length, truncated foot length, foot width, dorsum height, and navicular height when compared to a 3D scanner.

Specific Aim 2: **Determine the effects of different ankle braces on the plantar loading patterns of individuals with different arch heights.**

Hypothesis: Ankle braces will help to correct abnormal foot posture by redistributing irregular plantar loads. More specifically, ankle braces will shift plantar loading laterally in pes planus individuals and medially in pes cavus individuals.

Background

Foot posture is an important structural characteristic that can alter plantar loading, lower extremity kinematics, and contribute to risk of injury. The most well-known classification method for foot posture is arch height, where the foot is classified as pes planus (flat footed/low arched), pes rectus (normal), or pes cavus (high arched). Approximately 60% of the population have normal feet, 20% are pes planus, and 20% are pes cavus [1]. Many things can contribute to arch height, including age, gender, ethnicity, and use of footwear [10], [11]. Furthermore, arch structure can be related to midfoot mobility, as a pes planus foot usually has greater mobility than a pes cavus foot [3], [6]. The medial longitudinal arch is one of the most important measures of foot posture because it serves to absorb the majority of the force that is transmitted through the foot during ground contact [12], however there are many other aspects that should also be considered. Rearfoot alignment is another important foot posture measurement and can be closely related to arch height, as many pes planus individuals are in hindfoot valgus while pes cavus individuals are often in hindfoot varus. The angle of the rearfoot will change in a normal stride as the foot pronates upon impact and supinates for propulsion [13], but varus and valgus determinations are made statically. While there are many important factors that relate to the foot's structure, arch height and rearfoot alignment are two commonly known measures.

Due to the abnormal posture of pes planus and pes cavus feet, these individuals are often at an increased risk for lower extremity injuries [2]. Both planus and cavus individuals have been linked to having twice as many stress fractures [5] and an increased occurrence of ankle sprains [14], [15] when compared to individuals with a normal arch height. However, the structure of each foot posture leads to differences in how these injuries occur in addition to causing injuries unique to pes planus or pes cavus individuals. More specifically, the pes planus foot usually has increased mobility while the pes cavus foot is more rigid and results in increased lateral plantar pressures compared to a normal foot [3], [4]. Since these factors usually result in slightly different injuries for both pes planus and pes cavus individuals, it is important to understand how and why these injuries occur in order to prevent them in the future.

Common treatments and injury prevention methods for irregular foot posture and resulting lower extremity injuries are orthotics, ankle taping, and ankle braces. Orthotics are commonly used to correct foot posture, specifically low arches, as they help to control the direction of the ground reaction force, restore geometry of the medial longitudinal arch, and control the geometry and motion of the rearfoot [1], [16]. However, these orthoses are often only used by pes planus individuals and have little effect on the geometry of the foot or the motion of the rearfoot in pes cavus individuals. Taping and braces are more commonly used in sport due to the fact that they provide more ankle support and have been shown to prevent and reduce the severity of ankle sprains [8], [9], [17]. Furthermore, both ankle tape and braces restrict range of motion, specifically ankle inversion and eversion and hindfoot motion [9], [17], [18], reduce the rate of ankle reinjury [17], improve proprioception [9], [17], [18], and have a limited negative effect on other joints in the lower extremity [17]. While it has been reported that taping and braces may result in increased risk of knee injuries, there have been no significant results supporting this statement of increased risk [9]. Due to the many positive attributes of taping and bracing and their success in injury prevention, both treatments are beneficial for use in sport.

More specifically, taping is often preferred because, compared to braces, it is less bulky and can be fitted to the individual while accounting for any anatomic abnormalities [17]. Furthermore, taping has been shown to improve the ankle's functional and mechanical stability [17] in addition to being able to control arch height. The most commonly used taping technique to support the arch is Low-Dye taping, where strips of tape are placed longitudinally and transversely on the plantar surface of the foot [19]. This method has been reported to control vertical navicular height and the arch height ratio both statically and during exercise, suggesting that this method can help control pronation [20], [21]. Other taping techniques such as rigid taping and a navicular sling have also been shown to regulate navicular height [13], [22], and the rigid taping technique has been suggested as a preventative measure for medial tibial stress syndrome (MTSS) [22]. In addition to controlling the navicular height, taping can also support the arch structure through the redistribution of plantar pressure. While some studies have reported no difference in plantar pressure with ankle taping [22], others have found significant differences when using Low-Dye taping. More specifically, plantar pressure was decreased in the medial and lateral forefoot [13], [19] and heel [19], and increased in the lateral midfoot [13], [19] and toes [19]. These results and the lateral shift of pressure led to the conclusion that arch height was increased and pronation was reduced. However, tape has been shown to loosen during exercise [9], [13], [17], limiting its effectiveness for arch support and pronation control. This emphasizes the need to maintain the proper support of the arch and ankle without losing the restrictive properties during exercise.

Ankle braces may be one solution to this problem as they can maintain support for longer periods of time in addition to being easily readjusted during sport [17]. Similar to taping, braces have been shown to provide ankle support in addition to supporting the geometry of the hindfoot and arch in pes planus individuals [16]. Further use of braces on individuals with flatfoot deformity is supported by the finding that these braces may provide more support than the widely used foot orthoses [23]. Despite these promising results, no research has been done on the use of braces in flatfooted individuals, therefore highlighting the needs for further studies on this topic. Nevertheless, there has been significant research into the effectiveness of using braces for ankle injury prevention in sport. It has been reported that, while both bracing and taping can effectively reduce and prevent ankle sprains, bracing is more successful than traditional taping methods [8], [9]. More specifically, braces help to align the ankle joint in a neutral position and restrict the range of motion, therefore preventing inversion of the ankle [9], [18]. For proper sprain prevention and mechanical support, the strength of the brace should exceed the strength of the ligaments it is aiming to protect, which is about 6 - 56 kg in the ankle [24]. This is one reason that semirigid and rigid braces are more effective in ankle sprain prevention when compared with soft braces [18]. Despite the positive results for ankle brace use in injury prevention, there is still some concern over the effect of braces on performance. Previous studies have tried to address this concern and found that ankle braces did not affect physical performance, balance, or lower leg muscle power and activation [9], [17], [18]. Finally, in addition to the ankle support and injury prevention effects of ankle braces, they are also convenient for the user. Compared to taping, braces are more cost and time effective [17], [25], can be applied without an athletic trainer [9], [17], can be tightened as needed [9], [17], and are washable and reusable [17]. All these findings support the use of braces for prevention of ankle sprains in sport, but further research should be done to quantify their effectiveness in arch support based on foot type. This would allow researchers to determine if ankle braces have the potential for prevention of other lower extremity injuries that are associated with an abnormal arch structure or foot posture.

In order to properly analyze foot posture and the effect that a brace may have, it is important to use a well-established classification method. There are currently many different techniques that examine specific elements of foot structure and can therefore contribute to its classification as pes planus, pes cavus, or pes rectus [10]. While radiographs are considered the gold standard for determining measures of foot posture, they are often unnecessary due to the wide variety of other classification methods. One common method is arch height, which is a measurement of the medial longitudinal arch (MLA) height from the ground to the highest point of soft tissue on the dorsum of the foot [10]. Similarly, the arch height index (AHI) defines arch height as the height of the dorsum at half of the total foot length divided by the truncated foot length (posterior heel to first metatarsal head) [26]–[28]. This has been shown to be a reliable and valid method for measuring arch height and normative values have been established [26]–[28]. Arch height measurements are often interchangeable with navicular height, which is measured from the ground to the prominent navicular bone. Measurements of navicular height have been shown to be very accurate and strongly associated with radiographic measures of navicular height [11], and cutoff values for arch height classification based on navicular height have been determined [12]. The navicular bone is also commonly used to measure navicular drop, which is the difference in navicular height between an unweighted condition and an even bipedal stance [10]. While this navicular height measurement is commonly used, it has been shown to be the least consistent foot classification method [29].

Another method that can be used to classify foot posture is the longitudinal arch angle, also known as the feiss line. This is the angle formed by the medial aspect of the first metatarsal head and a line between the navicular tuberosity and the medial malleolus [10]. This has been shown to be a consistent method for classifying foot posture [29] and cutoff values for this classification have been determined in previous literature [12]. Rearfoot angle is the angle between a line in the middle of the calcaneus and a line in the middle of the distal third of the shank [10], which can be measured statically or dynamically [30]. While this is one of the most common ways to measure rearfoot alignment, previous studies have found only a moderate consistency of this measure between testing sessions [29]. Arch index is another method of foot posture classification, and is the ratio of the contact area of the middle third of the toeless footprint to the contact area of the entire toeless footprint [10]. This method has been found to be accurate as it correlates to radiographic measures of arch height [11]. The foot posture index (FPI) is another classification method that has become increasingly common and is based on a series of six to eight static foot posture measures while the participant is in a relaxed stance [11], [29]. Overall, the FPI has been reported to be the most consistent and accurate classification method [2], [29], [31], [32], despite some reports of its limited relationship to radiographic measures [11]. Even though static characteristics are often reported to have a poor relationship to dynamic motion of the foot [10], [32], these are a few of the commonly used and relatively accurate foot posture measurements that can contribute to the classification of the foot as pes planus, pes cavus, or normal.

While there are many established methods for determining the dimensions of the foot, the technologies used to do so are variable across research studies and are often expensive upon initial purchase. 3D scanners are commonly used by shoe retailers to measure the foot; however the measured dimensions are tailored for shoe fit and not for research purposes. Additionally, these scanners are extremely expensive to use in a research setting and many of the commonly used scanners have not been validated. Many researchers have therefore developed devices to measure specific foot dimensions such as foot length, truncated foot length, food width, and dorsum height [28], [33], [34]. However, many of these designs have not been published or validated or are not commercially available to the public. While the Arch Height Index Measurement System (AHIMS) is a commonly used device available for purchase, this system is also expensive and may not be feasible for smaller scaled research projects. This therefore highlights the need for an affordable and validated foot measurement device that can be used in a variety of research and clinical settings.

Differences in foot posture and arch height can also be evaluated using plantar loading measures, which help identify the pressures and forces applied to different regions of the foot. This method allows for the analysis of a variety of variables, including peak pressure, pressure-time integral (PTI), maximum force, force-time integral (FTI), contact area, and center of pressure (COP). These variables can be analyzed throughout the entire gait cycle for the entire foot, which can be divided into 3-12 different regions for a more specific analysis [6]. As previously mentioned, differences in the plantar loading variables have been attributed to different arch heights, which helps provide a more detailed analysis of how different foot postures can affect gait.

More specifically, pes planus feet have been reported to have an increased contact area in the medial midfoot [6], [7], [35], lateral forefoot [7], and 2nd toe [7] regions when compared to cavus and normal feet during walking. The increased contact on both the medial and lateral sides of the foot is likely a result of the increased mobility in the planus foot. Furthermore, planus feet have been shown to have more medially deviated plantar loading values and COPs during walking when compared to the cavus or normal foot [6]. Specifically, the maximum force has been shown to be higher in the hallux [6], [7], 2nd toe [6], [7], and medial midfoot [7], [35] and lower in the medial forefoot [6], [7], lateral forefoot [6], [7], [35], and lateral heel [7]. Likewise, the FTI was found to be higher in the hallux [6], [7], 2nd toe [7], central forefoot [6], and midfoot [6], [7] and lower in the medial and lateral forefoot [6], [7], [36]. The similarity in the maximum force and FTI results is expected as both measures are taking into consideration the forces in these regions during the gait cycle. The increased FTI in the middle forefoot despite normal force and the normal heel FTI despite lower forces can be attributed to increased contact time for these regions, which is likely due to the increased mobility of the planus foot [6]. Furthermore, the maximum force and FTI values support the notion that there is more force on the medial side of the foot in pes planus individuals than in pes cavus or normal arched individuals. The fact that the medial forefoot has a lower maximum force than the cavus foot is likely because the force in the planus foot is more evenly distributed over a greater contact area. The peak pressure in pes planus individuals has been found to be higher in the medial forefoot [6], hallux [6], [7], and 2nd toe [6], [7] and lower in the lateral forefoot [7], [35] and lateral heel [7]. Similarly, the PTI is higher in the medial midfoot [6], lateral forefoot [7], and all toes [7] while it is lower in the lateral forefoot [6] and lateral heel [7]. It is important to note the contrasting results for the PTI, which was reported to be lower in the lateral forefoot in planus feet in a meta-analysis [6] and higher in this region in a single study [7]. These conflicting results highlight the need for more robust studies on this topic in the future. The overall results for peak pressure and PTI are comparable to each other and indicate that the pressures in the planus foot are lower in the hindfoot and higher in the toes and on the medial side of the foot, as expected based on characteristics of the planus foot [6], [7].

The changes in plantar loading of the pes planus foot are likely due to the increased mobility of the foot. This mobility in the planus foot is also commonly associated with injuries due to abnormalities in the joint rotation or coupling, such as soft tissue stress injuries, knee injuries, and injuries on the medial side of the lower extremity [3], [4]. MTSS is one common injury that has been linked to the pronated foot posture and increased navicular drop that are usually found in pes planus feet [31], [37]. However, the relationship between pronation and MTSS in runners was found to be of small effect and not necessarily linked to dynamic foot function. This suggests that other factors such as hip range of motion (ROM), increased body mass index (BMI), gender, limited running experience, and orthotic use may also contribute to the development of MTSS [38]. Studies have also found that pes planus and pronation may be linked to patellofemoral pain [31] and exercise related lower leg pain (ERLLP) [37], but the strength of these relationships is low. In addition to injuries in the lower leg, it has also been reported that planus feet may predispose individuals to flatfoot deformity [23]. This occurs when the posterior tibial tendon and the ligaments supporting the arch of the foot begin to fail, resulting in a flattening of the arch and medial ankle pain [23]. However, despite the moderate evidence linking the pes planus foot to a variety of injuries, other studies have reported that the higher arch index usually associated with planus feet may not be associated with lower extremity injuries [39] and may even protect against [40] injuries of the foot and knee. These varying results highlight the need for more thorough investigations into the injuries associated with pes planus feet and their biomechanical causes in order to draw more accurate conclusions.

Equally important, the pes cavus foot has unique plantar loading and injury patterns. The examination of plantar loading in pes cavus feet has shown that they have a more laterally deviated COP trace and higher plantar loading values on the lateral side of the foot during walking [6]. The maximum force was reported to be higher in the lateral rearfoot and lower in the medial midfoot when compared to normal and planus feet [6]. However, the FTI has been found to be higher in the medial and lateral forefoot [6], [36] and lower in the medial midfoot [6], [36] and hallux [6] of pes cavus individuals. This suggests that the impact force on the lateral rearfoot is greatest, but that this force is quickly dissipated and the individual may spend an increased amount of the gait cycle with forces on the forefoot. Additionally, these results support the fact that pes cavus individuals have increased forces on the lateral side of the foot compared to the medial side, which also has a lower contact area. Furthermore, the peak pressure has been shown to be higher in the heel [3], [6] and forefoot [3], [6], [7] and lower in the medial arch [3]. Similarly, the PTI is higher in the heel [6] and medial and lateral forefoot [6], [7] and lower in the midfoot [6] and hallux [6] of cavus feet. This further supports the claim that individuals with this foot type may exert forces and pressures on the forefoot for a greater period of time, and that these pressures are more laterally deviated [6]. Additionally, the lower force, FTI, pressure, and PTI in the midfoot is likely due to the rigidity of the cavus foot and the fact that the midfoot may have decreased contact relative to the midfoot of a normal or planus foot. Furthermore, the contact area has been reported to be higher in the lateral forefoot of cavus feet [6], once again providing evidence of the lateral gait pattern in these individuals. While plantar loading trends have been determined for pes cavus feet, there is significantly more research into the plantar loading variables of pes planus feet, highlighting the need for a larger study with participants of each foot type.

The increased rigidity of the pes cavus foot contributes to increased plantar loading and results in reduced shock absorption [3]. These individuals are therefore more susceptible to bone, ankle, and lateral leg injuries [4]. Furthermore, the hindfoot varus and lateral center of pressure that is associated with the cavus foot has been linked to an increased risk of ankle sprains [14], [41], [42]. However, there has been limited research into other injuries that may result from a cavus foot. Additionally, most studies examining the association between foot posture and injury have focused on runners [2], [4], [31], [38], [40] or military members [2], [31], [37], as these populations put a lot of stress on the lower extremity. It would be beneficial to expand this research to an average population, who may also be at risk for injuries due to abnormal foot posture. This would provide more generalizable results and allow researchers to determine how to more effectively prevent these injuries.

Motivation, Purpose, and Hypothesis

As discussed, a large portion of the population has either pes planus or pes cavus feet [1], making it important to accurately classify and measure foot posture and foot dimensions. Unfortunately, the commonly used 3D scanners and custom foot measurement devices are often inaccessible or too expensive to be feasible for research projects and in clinical settings. The development and validation of an affordable foot measurement device would therefore allow researchers to easily and accurately classify foot dimensions in all settings.

Accurately classifying foot posture would allow researchers and clinicians to determine which individuals are predisposed to lower extremity injuries [2], [5], [14], [15] as a result of abnormal plantar loading [6], [7], [35], [36] and kinematics [3], [43] during daily activities. While foot orthoses are commonly used in pes planus individuals, a more generalizable solution is needed to stabilize the arch and ankle for individuals with a variety of foot postures. Ankle taping and bracing may be one solution as they can be universally worn and have been shown to reduce the occurrence of injuries such as ankle sprains [8], [9], [17]. While taping has also been reported to support arch structure [20]–[22], limited research has been done on the effect of braces on arch structure and abnormal foot posture. If ankle braces can be shown to support the structure of the foot and ankle for a variety of foot postures, they can potentially be used to help prevent injuries resulting from abnormal foot postures.

Therefore, the first purpose of this research was to develop an affordable system to accurately measure foot posture characteristics. It was hypothesized that an inexpensive system can be made to accurately and repeatably measure the foot posture characteristics of total foot length, truncated foot length, foot width, dorsum height, and navicular height when compared to a 3D scanner. The second purpose was to determine the effects of different ankle braces on individuals classified as normal, pes planus, and pes cavus. It was hypothesized that ankle braces would help correct abnormal foot posture by redistributing irregular plantar loads during dynamic tasks. More specifically, it was hypothesized that ankle braces would shift plantar loading laterally in pes planus individuals and medially in pes cavus individuals.

The Design and Validation of a Low-Cost Foot Measurement System

Abstract

It is estimated that approximately 40% of the population suffers from abnormal foot posture, specifically high arched or low arched feet. While the evaluation of foot posture can involve many aspects, it commonly requires the measurement of basic dimensions of the foot. Clinicians and researchers often rely on the use of specialized devices or 3D scanners to evaluate specific aspects of a patient's foot posture. However, current technologies are extremely expensive, therefore highlighting the need for a cost-effective device to be used in rural and clinical settings. Therefore, the purpose of this study was to develop an inexpensive system to measure total foot length, truncated length, dorsum height, navicular height, and foot width. This system had excellent validity when compared to a 3D scanner (ICC = 0.908-0.994), and good to excellent repeatability when compared between days (ICC = 0.867-0.996). These results demonstrate that it is possible to design an inexpensive, valid, and repeatable system that can be used in clinical, research, and rural settings to successfully evaluate basic dimensions of the foot for determination of foot type.

Introduction

Foot posture is an important characteristic that can affect injury risk, plantar loading, and lower extremity kinematics. The most well-known foot posture classification is by arch height, where 60% of people have normal arches (pes rectus), 20% have high arches (pes cavus), and 20% have low arches (flat footed or pes planus) [1]. Podiatrists and other specialists often rely on measurements of foot posture to help improve patient comfort and the reduce risk of injury associated with abnormal foot posture. More specifically, individuals with both pes planus and pes cavus feet are more susceptible to ankle sprains and stress fractures than individuals with normal foot posture [5], [14], [15]. Furthermore, the pes planus foot has increased mobility and is therefore linked to injuries resulting from abnormal joint motion, including soft tissue injuries, flatfoot deformity, knee injuries, and medial leg injuries [3], [4], [23], [31], [37]. Conversely, the pes cavus foot is more rigid and has been linked to injuries due to reduced shock attenuation, such as bone injuries, ankle injuries, and lateral leg injuries [3], [4], [14], [41], [42]. In order to determine the causes of these injuries and prevent future injuries from occurring, specialists often rely on specific measurements of foot posture. These measurements allow them to classify different aspects of foot posture, which in turn allows them to determine appropriate interventions.

There are many characteristics that are commonly analyzed during the evaluation of foot posture. These can include angular measures of the medial longitudinal arch [10], [12] or the rearfoot [10], or more complex measures based on the footprint [10]. However, the foot can also be characterized using different dimensions of the foot such as length and height. More specifically, medial longitudinal arch (MLA) height is a measure from the ground to the highest point of the dorsum of the foot [10] and the arch height index (AHI) is a measure based on the height of the dorsum at half of the total foot length and the truncated foot length, which is measured from the posterior heel to the first metatarsal head [26]–[28]. The navicular bone is often used to evaluate foot posture, either based on navicular height [11], [12] or navicular drop, which is the difference the in height of the navicular bone between an unweighted and weighted stance [10]. The ability to quantify these measures is important for a successful evaluation of foot posture.

While many clinical specialists rely on the trained human eye to evaluate foot posture, many also want a more accurate and objective analysis tool. Radiographs are the gold standard for these measurements, however this requires access to equipment and personnel that may not always be available to the treating specialist. 3D scanners are also becoming increasingly common for foot measurements, especially for shoe fitting purposes. These scanners can be used in the clinical setting, but are expensive costing around $10,000, often not validated, or not designed for medical measurements. The Arch Height Index Measurement System (AHIMS) is a more common tool that is simple and validated [28], however this system is also more expensive than smaller clinics or many research labs can afford. This therefore highlights the need for an inexpensive and valid foot measurement tool for use in clinical, research, or rural settings.

Therefore, the purpose of this study was to develop and validate a simple and inexpensive foot posture measurement system. This system would allow for the easy measurement of total foot length, truncated foot length, foot width, dorsum height, and navicular height. It was hypothesized that this system would be valid when compared against a 3D scanner and would be repeatable between days.

Materials and Methods

A custom built, inexpensive foot measurement system was built from easily accessible materials in order to meet the design requirements specified in Table 1.1. This system was designed to measure five major foot dimensions (Table 1.2, Figure 1.1), which were chosen as they are used most commonly in the various foot measurement methods. The Foot Posture Measurement System (FPMS) consisted of a gridded cutting mat attached to a plywood base to allow for the measurements of foot length, truncated foot length, and foot width. Heel cups were 3D printed to allow for easier positioning of the foot at the origin. A laser leveling ruler was mounted on a track between the feet to allow for adjustable measurements of navicular height and dorsum height. The finalized system is shown in Figure 1.2 and purchasing information for each component is displayed in Table 1.3.

Table 1.1: Design requirements and customer needs for a new foot measurement system.

Safe and Comfortable • No sharp edges • Smooth surface finish
Low Cost
Valid and Repeatable • Accurate and precise • Tight tolerances
Robust and Durable • Impact resistant • Long life
Easy to Use **Works Consistently**
Transportable • Low weight • Smaller size
Visually Appealing

Table 1.2: Foot dimensions measured in the present study, with their associated descriptions and labels in Figure 1.1.

Measure	Definition	Figure Label
Total Length	length of the foot from the back of the heel to the tip of the toes	1
Truncated Length	length of the foot from the back of the heel to the first metatarsal head	2
Navicular Height	height from the ground to the navicular tuberosity	3
Dorsum Height	height from the ground to the dorsum of the foot at half the total foot length	4
Width	width of the foot at the widest point	5

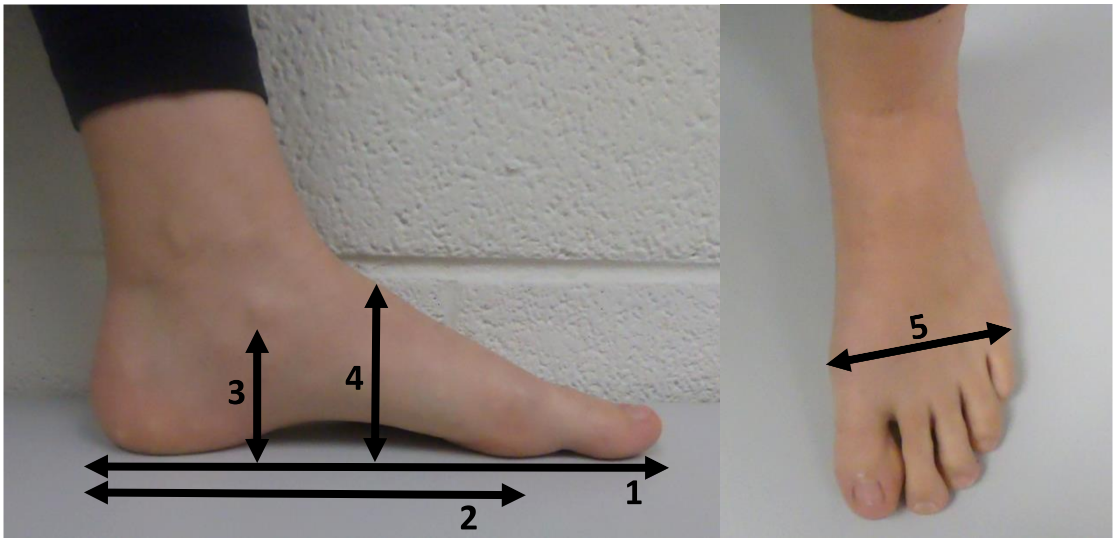

Figure 1.1: Schematic of the foot dimensions measured in the present study, including total length (1), truncated length (2), navicular height (3), dorsum height (4), and width (5).

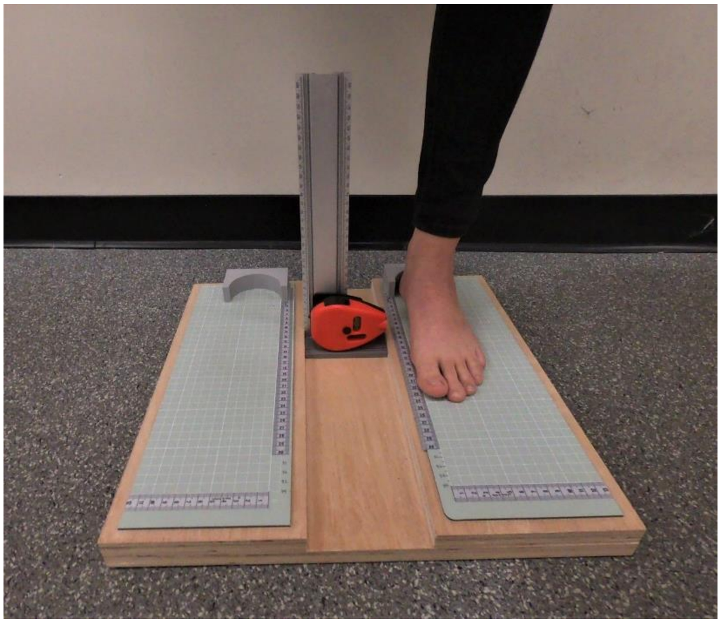

Figure 1.2: Photograph of the finalized Foot Posture Measurement System.

Table 1.3: Purchasing information for each component of the Foot Posture Measurement System.

Product	**Purpose**	**Price**
Plywood: 3/4"x2'x4'	base mount	$27.89
A3 Grey Cutting Mat - Double Sided CM and Inch Grid	lengths and width measurements	$25.41
Adhesive Metric Tape Measure	easier measurement guides	$7.99
3D Printed Heel Cups (2)	heel alignment	$3.00
300mm Aluminum Laser Level Ruler Measuring Tool	dorsum and navicular height measurements	$24.48
3D Printed Ruler Mount	ruler alignment and stabilization	$2.00
Gorilla Glue	attachments	$6.29
TOTAL		**$97.06**

In order to validate the FPMS, a total of 15 healthy young adults ages 18-30 with no current foot or ankle injury were recruited. Participants signed an institutional review board approved informed consent before study initiation. Demographic information including height, weight, gender, and age were collected. Participants began by removing shoes and socks in order to complete the foot measurements. Their feet were then measured bilaterally while standing with the custom-built measurement system and the 3D scanner, in a randomized order. The same procedure was repeated at two different visits, approximately a week apart, in order to determine between day repeatability.

Measurements with the FPMS were taken by having the participant step onto the surface of the device, placing their heels in the heel cups and lining up the medial side of their foot with the edge of the gridded mat. Total foot length, truncated foot length, and foot width measurements were taken from the lines on the mat. Navicular and dorsum heights were measured with the laser mounted onto the ruler by adjusting the ruler anterior/posterior and dorsal/plantar until the laser was lined up with the desired landmark.

Each foot was also scanned with a 3D scanner (Go!SCAN 50, AMTEK Creaform, Levis, Canada; accuracy: 0.500mm, resolution: 0.100mm) while standing, with weight evenly distributed between each foot. Reflective stickers were placed on the heel, navicular bone, first metatarsal, and fifth metatarsal to allow for easier scans and improved measurements. Total foot length, truncated foot length, foot width, navicular height, and dorsum height from the scan were measured using the associated VXelements software (AMTEK Creaform, Levis, Canada).

Interclass correlation coefficients (ICCs) were used to compare the measurements of foot length, truncated foot length, foot width, navicular height, and dorsum height between the FPMS and the 3D scanner. Bland-Altman plots were used to visualize the bias and error of the new measurement system when compared to the 3D scanner, and this was statistically evaluated using paired t-tests and 95% limits of agreement. ICCs and Bland-Altman plots were also used to determine the reliability of the system's measurements between each time point. These were calculated in MATLAB (MathWorks, Natick, MA), and ICCs were categorized as follows: >0.90 is excellent, 0.75-0.90 is good, 0.50-0.75 is moderate, and <0.50 is poor [44].

Results

ICC and Bland-Altman results for validity and repeatability of the FPMS are shown in Table 1.4. ICCs were all excellent for the system's validity (ICC = 0.908-0.994) and good to excellent for the system's repeatability (ICC = 0.867-0.996). Bland-Altman plots assessing the validity and repeatability of the foot measurement system can be found in Figure 1.3, with the majority of the points being within the 95% limits of agreement. The custom system and 3D scanner returned significantly different results for total length, truncated length, and navicular height ($p<0.001$). When evaluating the repeatability of the system, only the measure of dorsum height was significantly different between each visit ($p=0.022$).

Table 1.4: ICC and Bland-Altman results for the validity and repeatability of the Foot Posture Measurement System (CI = confidence interval, LoA = limits of agreement, * indicates significance).

		ICC	95% CI	Bias	95% LoA	p
Validity	*Length*	0.994	(0.990, 0.996)	-0.168	(-0.569, 0.233)	<0.001*
	Truncated Length	0.985	(0.975, 0.991)	0.160	(-0.352, 0.672)	<0.001*
	Dorsum Height	0.908	(0.846, 0.945)	0.050	(-0.452, 0.552)	0.136
	Navicular Height	0.960	(0.933, 0.976)	0.160	(-0.266, 0.586)	<0.001*
	Width	0.929	(0.880, 0.957)	0.077	(-0.609, 0.763)	0.095
Repeatability	*Length*	0.996	(0.992, 0.998)	-0.027	(-0.360, 0.307)	0.397
	Truncated Length	0.981	(0.961, 0.991)	-0.007	(-0.596, 0.582)	0.904
	Dorsum Height	0.895	(0.780, 0.950)	0.113	(-0.390, 0.617)	0.022*
	Navicular Height	0.867	(0.722, 0.937)	-0.010	(-0.722, 0.702)	0.881
	Width	0.930	(0.853, 0.967)	-0.080	(-0.720, 0.560)	0.190

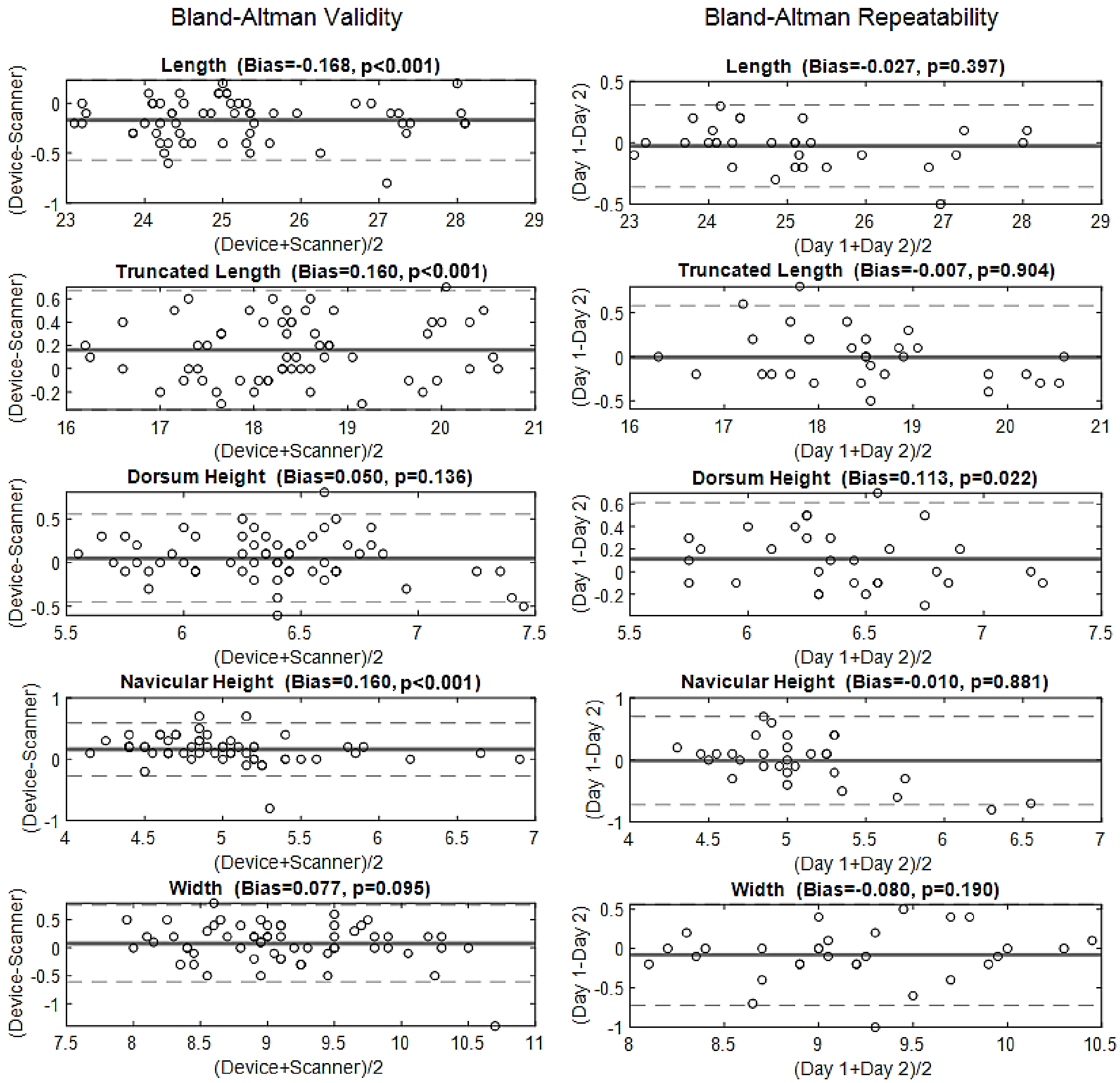

Figure 1.3: Bland-Altman plots of the validity and repeatability of the Foot Posture Measurement System for each foot dimension. The bias and paired t-test significance are reported for each. The thick dark line represents the bias while the thin dashed lines represent the 95% limits of agreement.

Discussion

The results of this study demonstrate that it is possible to design and fabricate an inexpensive and valid system to measure different aspects of foot posture. These included foot length, truncated foot length, foot width, dorsum height, and navicular height, which are commonly used to assess foot posture [10]–[12]. The Foot Posture Measurement System (FPMS), built in the present study, cost less than $100, making it easily affordable for smaller scale clinics, researchers, and for use in rural settings while still allowing for further customization or improvement. A large portion of this system's cost came from getting tools in metric units, such as the cutting mat and laser level ruler, shipped to the United States. This cost could be reduced for individuals in European or Asian countries or if the system is made in English units. Nevertheless, when compared to the cost of 3D scanners or the AHIMS, the current system is extremely cost effective and produces accurate results for simple measures of foot posture.

The FPMS returned accurate and repeatable values for all aspects of foot posture when validated against a 3D scanner. ICCs for the validity of the system were excellent for every measure (ICC = 0.908-0.994). These results are more accurate than those reported in a previous study comparing radiographs to foot measurements made without a specific device [26]. Despite the high correlation in this study, Bland-Altman plots showed some statistically significant differences in the agreement between the custom system and the 3D scanner for measures of total length, truncated length, and navicular height ($p<0.001$). More specifically, the custom system underestimated total foot length by an average 0.168 cm and overestimated truncated foot length and navicular height by 0.160 cm. However, this result is similar to the biases reported in a previous study completed using a foot scan [33]. Furthermore, these biases are small when compared to the total scale of the measurements (average total length: 25.10 cm, average truncated length: 18.50 cm, average navicular height: 5.08 cm).

When looking specifically at the repeatability of the FPMS, the ICCs were good for dorsum height (ICC = 0.895) and navicular height (ICC = 0.867) and excellent for total length, truncated length, and width (ICC = 0.930-0.996). The lower correlation between days for dorsum height and navicular height is likely a result of operator inconsistencies in locating the necessary landmarks on the foot [26], [29]. Repeatability Bland-Altman plots revealed that only dorsum height was statistically different between visits ($p=0.022$). The results from the first visit were an average of 0.113 cm higher than the results from the second visit, however this is unlikely to have a large impact as the average dorsum height was 6.39 cm. Overall, the results for the custom foot measurement system demonstrate that it is valid compared to a 3D scanner and repeatable between days.

There were a few limitations associated with the present study. First, the successful building of the FPMS requires attention to detail. The alignment of the heel cups on the gridded mat and the orientation of the vertical ruler must be done precisely in order to get the most accurate results. Furthermore, the measurement of each foot required the successful identification of landmarks, such as the navicular bone. Failure to do so consistently could negatively impact the validity or repeatability of the system. Finally, there was a significant amount of manual processing involved for each of the 3D scans. Small errors in the calibration, alignment, or measurements of these scans could compile to affect the results of the study. Nevertheless, most of these errors can be avoided with careful manufacturing of the system and measurement of the participant’s feet.

In conclusion, a foot measurement system can be built for less than $100, which is significantly less expensive than similar products currently on the market. The FPMS successfully measured total foot length, truncated foot length, dorsum height, navicular height, and foot width, and was shown to be valid when compared to a 3D scanner and repeatable between days. This methodology can therefore be used by clinicians and researches in smaller facilities and rural settings to successfully evaluate foot posture and improve comfort as well as reduce the risk of injury in individuals with abnormal foot posture.

Effects of Ankle Bracing on Foot Posture and Plantar Loading

Abstract

Arch height is one important aspect of foot posture and it is estimated that approximately 60% of the population has a normal arch, while 20% of the population is flat footed/low arched or pes planus, and 20% of the population is high arched or pes cavus. These different foot postures can alter lower extremity kinematics, plantar loading, and contribute to injury risk. Ankle taping and bracing are commonly used in sport to prevent these injuries, and while studies have examined the effects of ankle taping on foot posture and plantar loading, there has been no research into the effects of ankle bracing on these measures. Therefore, the purpose of this study was to evaluate the effects of ankle braces on plantar loading during athletic tasks. It was hypothesized that ankle braces would shift the plantar loading patterns medially in pes cavus individuals and laterally in pes planus individuals. 36 participants were recruited for this study and completed walking, running, and cutting tasks in an unbraced condition, in a lace-up stabilizing brace, and in a semi-rigid brace. Plantar loading variables of contact area, maximum force, and force-time integral were analyzed for two midfoot and three forefoot regions, in addition to an analysis of the displacement of the center of pressure. A 3x3 mixed model repeated measures ANOVA was used to determine the effect of brace and foot type (α=0.05). It was found that that foot type significantly affected force measures on the medial side of the foot in all tasks (p=0.001-0.047). Brace type affected contact area (p=0.001-0.0014), maximum force ($p<0.001$ – $p=0.043$), and force-time integral ($p<0.001$ – $p=0.015$) in the medial and midfoot regions during all tasks, with the majority of the differences resulting from the rigid brace. Portions of the center of pressure displacement were significantly affected by brace-wear in both the medial-lateral and anterior-posterior directions (p=0.001-0.050). This allows for the conclusion that ankle braces can be worn during athletic tasks to redistribute plantar loading patterns and potentially reduce the risk of some overuse injuries independent of foot type.

Introduction

Foot posture can alter movement and potentially result in lower extremity injuries. One aspect of foot posture or foot type is arch height, which is extremely important as the medial longitudinal arch is responsible for absorbing the majority of the impact on the foot during daily activities [12]. Analyzing arch height and mobility can allow an individual to be classified as pes planus (flat-footed or low arched), pes rectus (normal), or pes cavus (high arched). While approximately 60% of individuals are normal, 20% of the population is pes planus and 20% of the population is pes cavus [1]. Different foot types are often associated with different rearfoot angles, which is the angle between the calcaneus and the shank [10]. More specifically, pes planus feet are often in hindfoot valgus and pes cavus feet are often in hindfoot varus [45]. While there are many aspects of foot posture, arch height and rearfoot angle are two well-known characteristics that are commonly used to determine foot type in clinical settings.

In addition to different foot types leading to different structural characteristics, they can also result in different plantar loading patterns [6], [7], [35]. For example, the planus foot has a more medial center of pressure (COP) and higher pressures, forces, and contact area in the medial midfoot, medial forefoot, and hallux during the walk [6], [7]. This is due to increased midfoot mobility and the hindfoot valgus position of the rearfoot, which allows the foot to collapse medially resulting in increased plantar loading on the medial aspect of the foot [6], [7]. On the other hand, the cavus foot has a more lateral COP and higher pressures in the heel and lateral forefoot during walking [6]. This is the result of a more rigid foot and the hindfoot varus position of the rearfoot, causing increased loading on the lateral aspect of the foot [6]. These plantar loading patterns for different foot types are important to consider when evaluating the function of the foot and potential injury risk.

Injuries have been shown to be related to abnormal arch height [2] as a result of both abnormal plantar loading [6], [7], [35], [36] and kinematics [3], [43]. Both pes planus and pes cavus individuals are two times more likely to suffer from stress fractures [5] and are at an increased risk for ankle sprains [14], [15]. More specifically, pes planus individuals are predisposed to soft tissue injuries, knee injuries, and injuries on the medial side of the legs [3], [4]. This could be due to the increased mobility in the foot as well as the increased medial plantar loading patterns [3]. Conversely, pes cavus individuals are more likely to experience bone injuries as well as foot and lateral leg injures [3]–[5]. These injuries are a result of the increased rigidity of the foot, reduced shock attenuation, and increased plantar pressures in the pes cavus foot [3]. The incidence of injuries due to abnormal foot types highlights the need to identify potential injury prevention options.

Currently, ankle bracing and ankle taping are used commonly in sport to prevent injury, and have both been shown to be effective for reducing the occurrence and severity of ankle sprains [8], [9], [17]. Taping has the advantage of being less bulky with a more individualized fit [17], however tape loosens during exercise potentially reducing injury prevention effectiveness [9], [13], [17]. Furthermore, ankle taping has been shown to control navicular height and reduce pronation [20]–[22]. Bracing, on the other hand, maintains support throughout the duration of exercise, can be easily readjusted by the individual, and is reusable [9], [17]. Additionally, bracing has been shown to be more effective in the prevention of ankle sprains [8], [9], [18], however no research has been completed to assess the ability of ankle braces to control the arch.

Therefore, the purpose of this study was to evaluate the effects of different ankle braces on foot posture through the analysis of plantar loading patterns during walking, running, and cutting. It was hypothesized that ankle braces would reduce medial plantar loading in pes planus individuals by shifting pressures laterally and reduce lateral plantar loading in pes cavus individuals by shifting pressures medially.

Materials and Methods

36 individuals were recruited for this study based off a power analysis on previously reported effect sizes from a comparison between plantar loading variables for different foot posture groups [6], [7]. Healthy young adults ages 18-30 who were recreationally active were recruited from Virginia Tech and the surrounding area. Individuals with a history of lower extremity surgery or injuries within the past six months were excluded. Participants signed IRB approved informed consent documents before completing the study.

Arch height index (AHI) was used to classify participants into the following foot types: cavus, rectus, or planus. This was done by measuring from the floor to the height of the dorsum at half the total foot length and dividing that by the length from the posterior aspect of the calcaneus to the first metatarsal head (Equation 1) using the Foot Posture Measurement System (FPMS). A ratio of 0.315 or less classified participants as pes planus, between 0.315 and 0.365 classified them as normal, and a ratio of 0.65 or more classified participants as pes cavus, based on previously reported normative values [28] and the fact that 60% of the population has normal arches [1].

$$AHI = \frac{Dorsal\ Height_{50\%\ Foot\ Length}}{Truncated\ Foot\ Length} \tag{1}$$

Participants were fitted for a series of braces before completing the remainder of the testing session. Brace conditions included no brace, a combination lace up and figure eight stabilizer brace (DonJoy Performance Anaform Lace-Up Ankle Brace, DonJoy Orthopaedics, Vista, CA), and a brace with semi-rigid lateral stabilizers and increased midfoot support (AirCast AirLift PTTD Brace, DonJoy Orthopaedics, Vista, CA) (Figure 2.1). These braces were worn bilaterally and the order of the brace conditions was randomized.

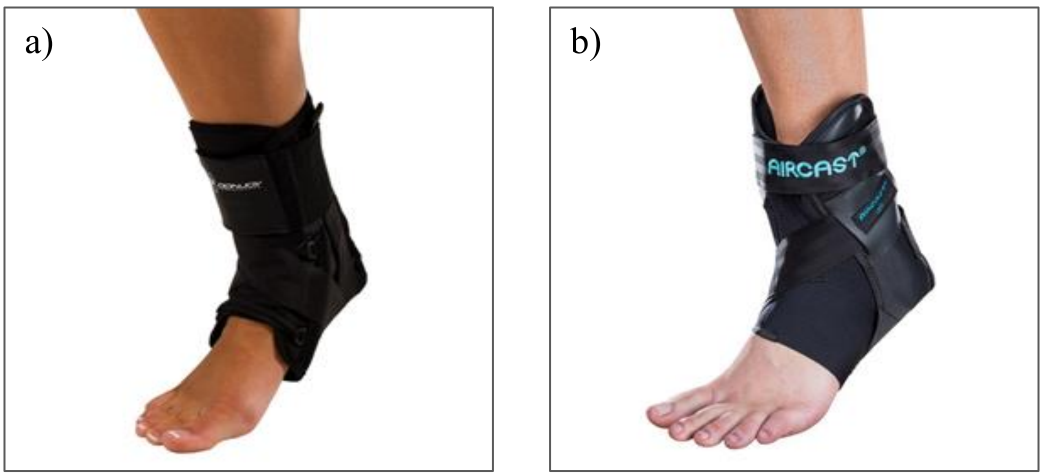

Figure 2.1: The (a) stabilizer brace and the (b) semi-rigid brace worn during the study.

Participants were asked to use a neutral cushioned running shoe (Nike Zoom Pegasus, Beaverton, OR) during all testing to control for footwear effects [46]. All participants completed seven walking, running, and cutting trials in a randomized order. Participants completed three walking and running practice trials at a self-selected pace in order to get an average speed, determined by timing gates (Brower Timing Systems, Draper, Utah) set 6 meters apart. Trial completion speed for all subsequent trials was held to within 5% of the average pace from the practice trials. The side cut was performed by planting on either foot. The planted foot remained the same for each trial and this foot was used for analysis across all conditions. The three tasks and three brace conditions combined for a total of 9 conditions, and volitional breaks were provided to prevent fatigue. This procedure is diagrammed in Figure 2.2.

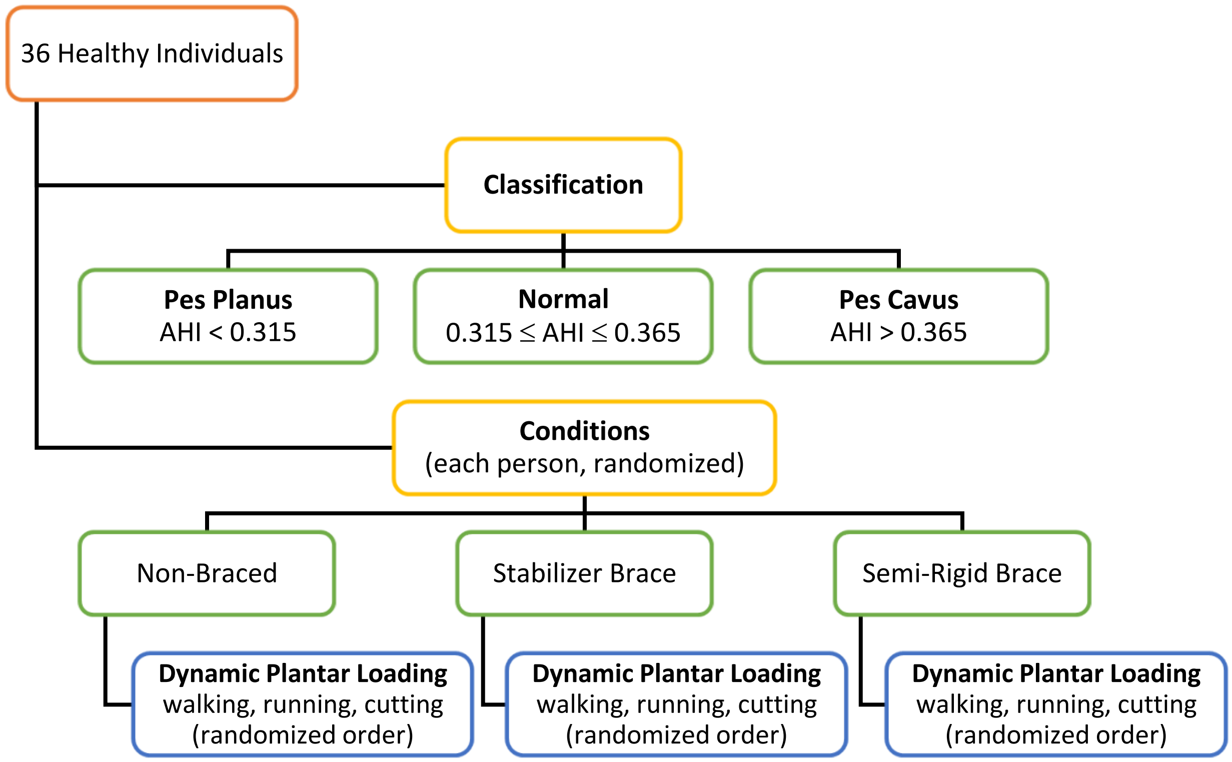

Figure 2.2: Flow chart of testing procedure (AHI = arch height index).

The plantar loading variables of contact area, maximum force, force-time integral, and center of pressure were evaluated during the dynamic trials using the pedar®-X system (Novel, Munich, Germany) at 50 Hz. The Novel Multiproject software was used with an eight-region mask [35], [47], and plantar loading was evaluated in the medial midfoot, lateral midfoot, medial forefoot, middle forefoot, and lateral forefoot. This was evaluated in the walk, run, and cut for whichever foot the participant used to plant during the cut. Additionally, participants rated brace comfort on a Likert scale from 1-10.

The plantar loading variables were examined for each region of the foot and in each condition, with maximum force being normalized to body weight and contact area being normalized to the contact area of the entire insole (NICA: normalized insole contact area) [47]. After verifying that all other ANOVA assumptions were met, data was log-transformed to result in a normal distribution of variables. A 3x3 mixed model repeated measures ANOVA (α=0.05) was run to examine the effects of brace (within-subjects factor) and foot type (between-subject factor) on all variables (SPSS version 26, IBM, Armonk, NY). Bonferroni adjusted pairwise comparisons were run to determine the simple main effects of brace and foot type on any variables for which a main effect was identified. Center of pressure displacement for the walking and running trials were evaluated using the coefficient of multiple correlation (CMC) [48] and statistical parametric mapping (SPM) [49], [50] in MATLAB (MathWorks, Natick, MA) to determine if brace wear had an effect on COP displacement. COP curves were separated into the anterior-posterior (AP) component and the medial-lateral (ML) component for evaluation, as is commonly done for analysis of this type [49], [50]. Finally, since the brace comfort ratings were sets of ordinal data, they were compared with a Wilcoxon signed-rank test.

Results

36 participants (11 male/25 female; age: 23.1 ± 2.5 years; height: 1.72 ± 0.09 m, weight: 66.3 ± 14.7 kg) completed the entire testing protocol. Results from the 3x3 mixed model repeated measures ANOVA revealed that both brace and foot type had a significant main effect on certain variables. During walking (Figure 2.3, Table 2.1), no foot type by brace interaction existed. There was a main effect of brace on contact area in the medial midfoot (p=0.005, η_p^2=0.280), medial forefoot (p=0.001, η_p^2=0.338), and middle forefoot (p=0.007, η_p^2=0.264). Brace type also significantly affected maximum force in the medial midfoot (p<0.001, η_p^2=0.710), lateral midfoot (p<0.001, η_p^2=0.561), and medial forefoot (p=0.011, η_p^2=0.247) and force-time integral in the medial midfoot (p<0.001, η_p^2=0.476). Overall, the rigid brace increased values in the midfoot and decreased them in the forefoot compared to the unbraced or stabilizing brace condition. There was a main effect of foot type during walking for maximum force in the medial forefoot (p=0.014, η_p^2=0.229), and middle forefoot (p=0.003, η_p^2=0.302). In general, a higher arch resulted in lower loads in the midfoot and higher loads in the forefoot.

During running (Figure 2.4, Table 2.2) there was a brace by foot type interaction for the maximum force in the medial forefoot (p=0.045, η_p^2=0.136). There was a significant main effect of brace on the maximum force in the medial midfoot (p<0.001, η_p^2=0.827), lateral midfoot (p<0.001, η_p^2=0.379), and middle forefoot (p=0.001, η_p^2=0.349). Significant brace main effects also existed for the force-time integral in the medial midfoot (p<0.001, η_p^2=0.570) and the medial forefoot (p=0.006, η_p^2=0.276). Compared to the unbraced and stabilizing brace condition, the rigid brace usually resulted in higher loads in the midfoot and lower loads in the forefoot. Maximum force was significantly different in the medial midfoot (p=0.039, η_p^2=0.179) and middle forefoot (p=0.012, η_p^2=0.235) based on foot type. Foot type also affected force-time integral in the medial forefoot (p=0.032, η_p^2=0.188) and middle forefoot (p=0.047, η_p^2=0.169). Overall, values were decreased in the midfoot and increased in the forefoot with a higher arch.

During cutting (Figure 2.5, Table 2.3), there was no interaction between brace and foot type. There was a significant main effect of brace for contact area in the medial midfoot (p=0.014, η_p^2=0.234) and lateral midfoot (p=0.006, η_p^2=0.275), and on maximum force in the medial midfoot (p<0.001, η_p^2=0.785), lateral midfoot (p<0.001, η_p^2=0.546), and medial forefoot (p=0.043, η_p^2=0.178). The force-time integral in the medial midfoot (p<0.001, η_p^2=0.786), lateral midfoot (p<0.001, η_p^2=0.465), and medial forefoot (p=0.015, η_p^2=0.230) were different based on brace type. On average, the rigid brace resulted in lower loads in the forefoot and higher loads in the midfoot compared to the unbraced and stabilizing brace condition. Foot type had a significant main effect on maximum force in the medial midfoot (p=0.013, η_p^2=0.232), lateral midfoot (p=0.001, η_p^2=0.341), medial forefoot (p=0.004, η_p^2=0.282), and middle forefoot (p=0.016, η_p^2=0.221), and a main effect on the force-time integral in the medial forefoot (p=0.005, η_p^2=0.276) and the middle forefoot (p=0.008, η_p^2=0.256). A higher arch usually led to decreased loads in the midfoot and increased loads in the forefoot.

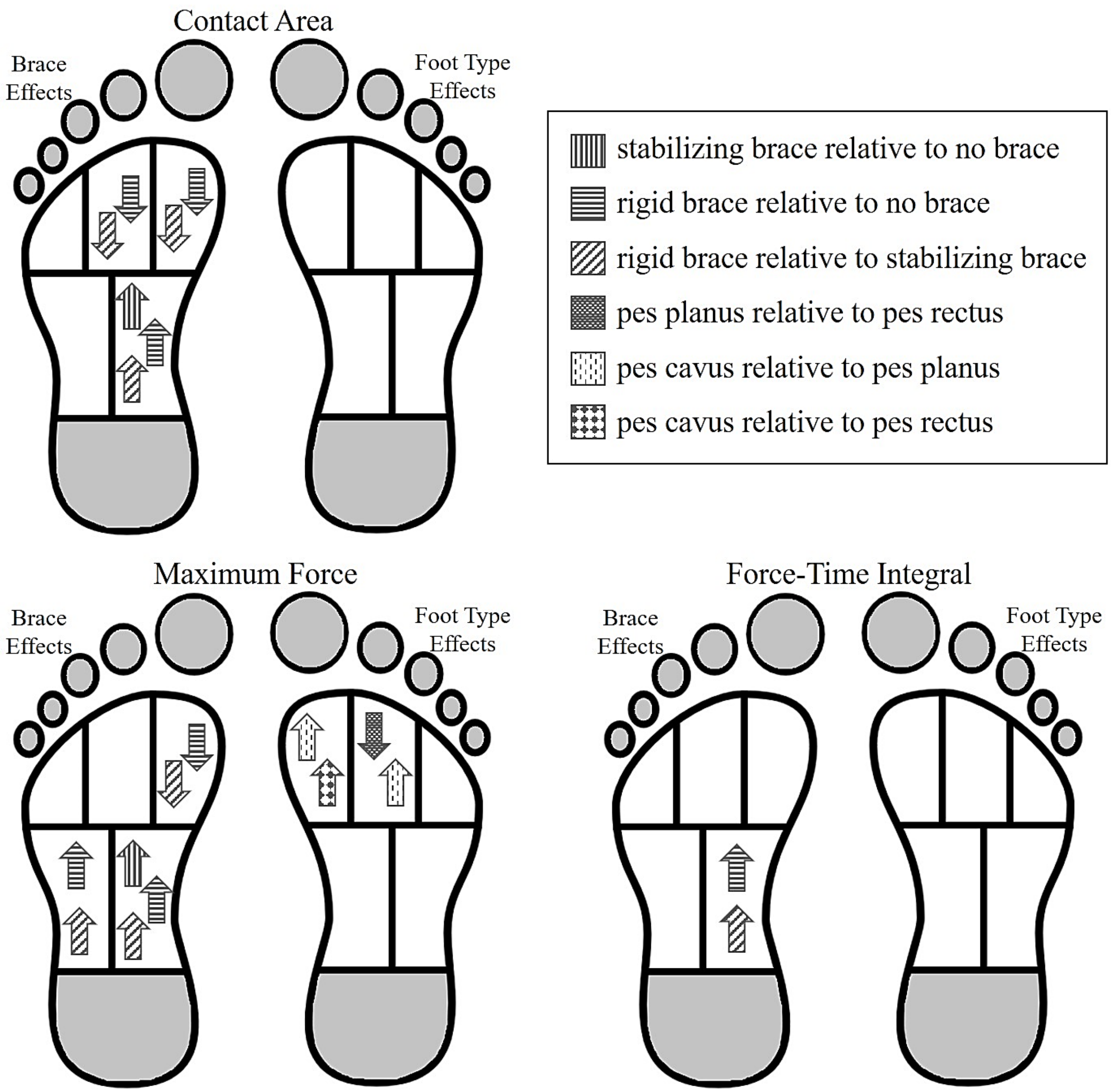

Figure 2.3: Results from a 3x3 mixed model repeated measures ANOVA and post-hoc testing comparing brace condition and foot type for contact area, maximum force, and force-time integral in the five regions of interest during the walk.

Table 2.1: Mean (SD) and results from a 3x3 mixed model repeated measures ANOVA comparing brace condition and foot type for contact area, maximum force, and force-time integral in the five regions of interest during the walk.

		Pes Planus			Pes Rectus			Pes Cavus					
		NB	SB	RB	NB	SB	RB	NB	SB	RB	Interaction	ME Brace	ME Foot Type
Contact Area (NICA)	Medial Midfoot[a,b,c]	0.140 (0.042)	0.144 (0.033)	0.155 (0.006)	0.137 (0.024)	0.142 (0.025)	0.155 (0.004)	0.129 (0.021)	0.138 (0.015)	0.154 (0.004)	p=0.932 η_p^2=0.013	**p=0.005*** **η_p^2=0.280**	p=0.924 η_p^2=0.005
	Lateral Midfoot	0.161 (0.007)	0.162 (0.005)	0.162 (0.003)	0.163 (0.007)	0.162 (0.008)	0.161 (0.003)	0.165 (0.005)	0.165 (0.005)	0.160 (0.002)	p=0.092 η_p^2=0.112	p=0.071 η_p^2=0.152	p=0.662 η_p^2=0.025
	Medial Forefoot[b,c]	0.077 (0.002)	0.078 (0.002)	0.072 (0.006)	0.077 (0.007)	0.078 (0.004)	0.073 (0.007)	0.079 (0.004)	0.078 (0.004)	0.076 (0.006)	p=0.673 η_p^2=0.034	**p=0.001*** **η_p^2=0.338**	p=0.464 η_p^2=0.045
	Middle Forefoot[b,c]	0.093 (0.005)	0.093 (0.006)	0.092 (0.001)	0.094 (0.005)	0.094 (0.005)	0.092 (0.002)	0.095 (0.003)	0.095 (0.004)	0.092 (0.002)	p=0.837 η_p^2=0.021	**p=0.007*** **η_p^2=0.264**	p=0.731 η_p^2=0.019
	Lateral Forefoot	0.085 (0.007)	0.086 (0.005)	0.085 (0.002)	0.087 (0.003)	0.084 (0.008)	0.085 (0.001)	0.087 (0.003)	0.086 (0.002)	0.084 (0.002)	p=0.229 η_p^2=0.079	p=0.138 η_p^2=0.116	p=0.843 η_p^2=0.010
Maximum Force (BW)	Medial Midfoot[a,b,c]	0.171 (0.065)	0.173 (0.063)	0.236 (0.039)	0.120 (0.036)	0.132 (0.047)	0.201 (0.055)	0.094 (0.024)	0.107 (0.016)	0.176 (0.023)	p=0.720 η_p^2=0.031	**p<0.001*** **η_p^2=0.710**	p=0.094 η_p^2=0.134
	Lateral Midfoot[b,c]	0.230 (0.033)	0.235 (0.036)	0.269 (0.040)	0.220 (0.048)	0.217 (0.032)	0.261 (0.032)	0.198 (0.043)	0.220 (0.052)	0.240 (0.048)	p=0.486 η_p^2=0.050	**p<0.001*** **η_p^2=0.561**	p=0.284 η_p^2=0.073
	Medial Forefoot[b,c,2,3]	0.164 (0.038)	0.149 (0.031)	0.125 (0.038)	0.176 (0.055)	0.168 (0.063)	0.151 (0.064)	0.234 (0.077)	0.242 (0.088)	0.228 (0.083)	p=0.561 η_p^2=0.044	**p=0.011*** **η_p^2=0.247**	**p=0.014*** **η_p^2=0.229**
	Middle Forefoot[1,2]	0.206 (0.060)	0.209 (0.043)	0.194 (0.044)	0.262 (0.039)	0.260 (0.046)	0.243 (0.049)	0.276 (0.054)	0.277 (0.062)	0.260 (0.057)	p=0.921 η_p^2=0.014	p=0.132 η_p^2=0.119	**p=0.003*** **η_p^2=0.302**
	Lateral Forefoot	0.153 (0.050)	0.164 (0.052)	0.164 (0.042)	0.201 (0.051)	0.203 (0.049)	0.213 (0.048)	0.185 (0.045)	0.193 (0.055)	0.173 (0.047)	p=0.302 η_p^2=0.070	p=0.153 η_p^2=0.111	p=0.073 η_p^2=0.147
Force Time Integral (N*s)	Medial Midfoot[b,c]	130.3 (182.9)	127.5 (157.9)	199.7 (261.3)	67.8 (65.1)	99.2 (105.9)	173.9 (147.4)	81.0 (100.1)	108.3 (103.1)	127.4 (134.1)	p=0.752 η_p^2=0.028	**p<0.001*** **η_p^2=0.476**	p=0.935 η_p^2=0.004
	Lateral Midfoot	187.7 (207.9)	198.5 (239.7)	216.9 (267.3)	168.2 (174.5)	174.2 (165.3)	226.5 (181.1)	132.5 (143.3)	174.3 (052.3)	147.8 (146.1)	p=0.605 η_p^2=0.040	p=0.204 η_p^2=0.095	p=0.806 η_p^2=0.013
	Medial Forefoot	81.3 (75.7)	74.7 (83.1)	48.6 (42.0)	80.5 (72.7)	84.2 (80.1)	76.7 (64.0)	116.9 (134.2)	145.6 (145.6)	110.4 (129.1)	p=0.588 η_p^2=0.041	p=0.348 η_p^2=0.064	p=0.599 η_p^2=0.031
	Middle Forefoot	108.8 (101.6)	105.1 (104.4)	104.9 (115.2)	129.3 (114.0)	140.8 (123.0)	140.4 (114.1)	156.4 (185.1)	198.4 (188.5)	145.1 (175.0)	p=0.726 η_p^2=0.030	p=0.619 η_p^2=0.030	p=0.585 η_p^2=0.032
	Lateral Forefoot	77.3 (72.0)	80.1 (84.5)	72.4 (64.6)	120.2 (125.3)	111.1 (104.8)	128.2 (108.3)	110.5 (126.9)	143.6 (132.8)	99.8 (118.8)	p=0.562 η_p^2=0.043	p=0.709 η_p^2=0.021	p=0.506 η_p^2=0.040

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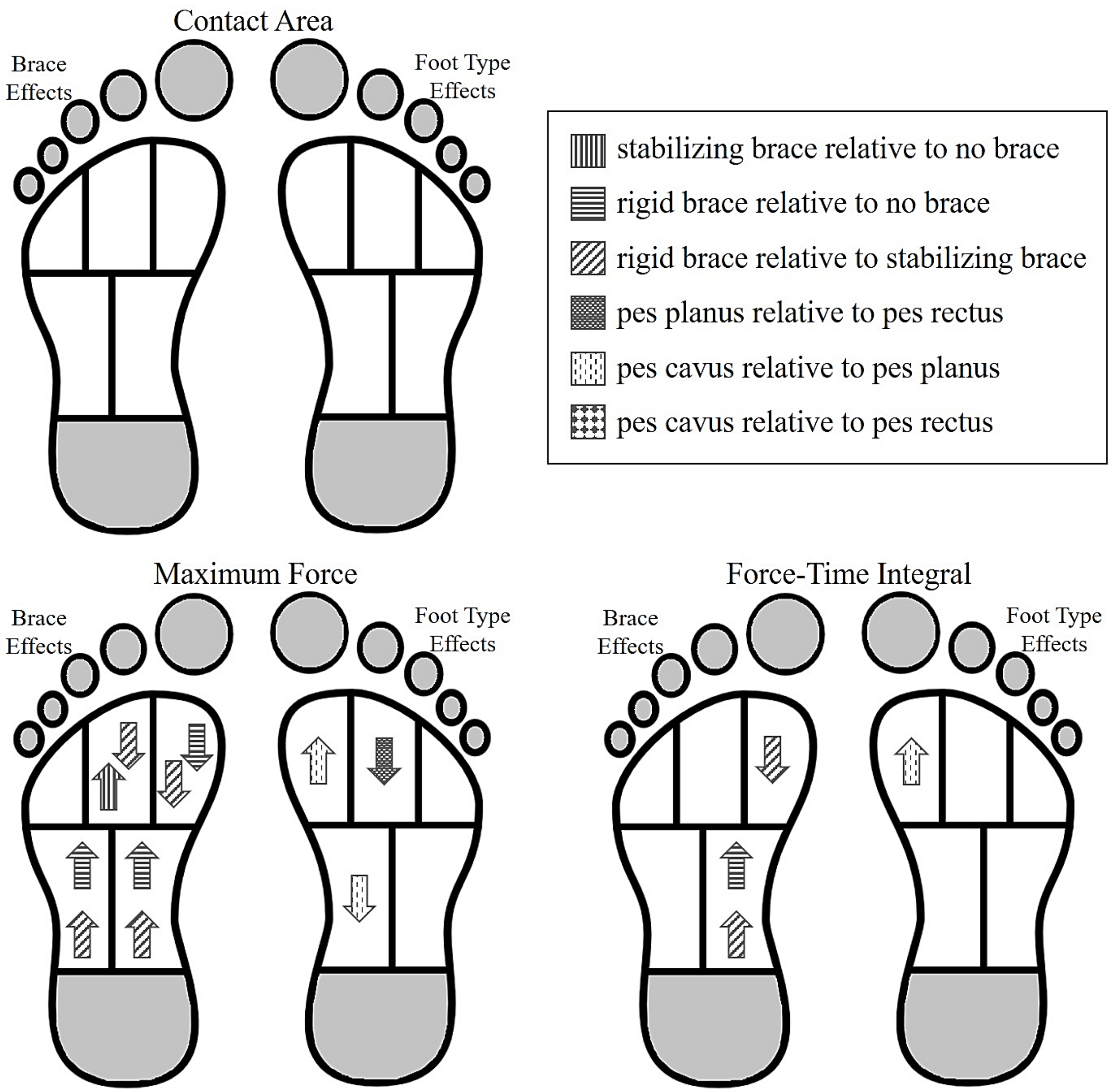

Figure 2.4: Results from a 3x3 mixed model repeated measures ANOVA and post-hoc testing comparing brace condition and foot type for contact area, maximum force, and force-time integral in the five regions of interest during the run.

Table 2.2: Mean (SD) and results from a 3x3 mixed model repeated measures ANOVA comparing brace condition and foot type for contact area, maximum force, and force-time integral in the five regions of interest during the run.

		Pes Planus			Pes Rectus			Pes Cavus					
		NB	**SB**	**RB**	**NB**	**SB**	**RB**	**NB**	**SB**	**RB**	**Interaction**	**ME Brace**	**ME Foot Type**
Contact Area (NICA)	Medial Midfoot	0.151 (0.013)	0.154 (0.007)	0.158 (0.003)	0.157 (0.004)	0.159 (0.005)	0.159 (0.006)	0.153 (0.009)	0.154 (0.008)	0.157 (0.002)	p=0.447 η_p^2=0.054	p=0.054 η_p^2=0.167	p=0.105 η_p^2=0.128
	Lateral Midfoot	0.162 (0.006)	0.160 (0.004)	0.162 (0.004)	0.160 (0.011)	0.160 (0.007)	0.162 (0.007)	0.161 (0.002)	0.160 (0.003)	0.160 (0.001)	p=0.359 η_p^2=0.063	p=0.320 η_p^2=0.069	p=0.927 η_p^2=0.005
	Medial Forefoot	0.075 (0.006)	0.077 (0.001)	0.075 (0.009)	0.079 (0.003)	0.079 (0.003)	0.079 (0.003)	0.078 (0.001)	0.078 (0.001)	0.078 (0.001)	p=0.638 η_p^2=0.037	p=0.448 η_p^2=0.049	p=0.080 η_p^2=0.142
	Middle Forefoot	0.091 (0.002)	0.092 (0.003)	0.093 (0.003)	0.092 (0.003)	0.093 (0.003)	0.092 (0.004)	0.092 (0.001)	0.092 (0.001)	0.092 (0.001)	p=0.367 η_p^2=0.062	p=0.677 η_p^2=0.024	p=0.838 η_p^2=0.011
	Lateral Forefoot	0.085 (0.003)	0.084 (0.002)	0.085 (0.002)	0.086 (0.003)	0.084 (0.008)	0.086 (0.003)	0.084 (0.002)	0.084 (0.002)	0.084 (0.001)	p=0.823 η_p^2=0.022	p=0.617 η_p^2=0.030	p=0.924 η_p^2=0.005
Maximum Force (BW)	Medial Midfoot[b,c,2]	0.324 (0.092)	0.334 (0.093)	0.400 (0.073)	0.275 (0.069)	0.279 (0.064)	0.359 (0.083)	0.219 (0.043)	0.232 (0.048)	0.327 (0.057)	p=0.147 η_p^2=0.096	**p<0.001* η_p^2=0.827**	**p=0.039* η_p^2=0.179**
	Lateral Midfoot[b,c]	0.390 (0.085)	0.424 (0.099)	0.443 (0.075)	0.364 (0.098)	0.369 (0.097)	0.419 (0.087)	0.319 (0.050)	0.338 (0.051)	0.376 (0.055)	p=0.748 η_p^2=0.028	**p<0.001* η_p^2=0.379**	p=0.200 η_p^2=0.093
	Medial Forefoot[b,c,2]	0.234 (0.066)	0.237 (0.046)	0.183 (0.050)	0.312 (0.094)	0.312 (0.105)	0.277 (0.095)	0.319 (0.091)	0.331 (0.091)	0.304 (0.063)	**p=0.045* η_p^2=0.136**	p<0.001 η_p^2=0.497	p=0.028 η_p^2=0.195
	Middle Forefoot[a,c,1]	0.283 (0.078)	0.305 (0.065)	0.265 (0.072)	0.382 (0.082)	0.393 (0.079)	0.376 (0.085)	0.372 (0.107)	0.381 (0.100)	0.351 (0.080)	p=0.420 η_p^2=0.056	**p=0.001* η_p^2=0.349**	**p=0.012* η_p^2=0.235**
	Lateral Forefoot	0.214 (0.083)	0.230 (0.073)	0.227 (0.070)	0.246 (0.067)	0.255 (0.082)	0.268 (0.067)	0.243 (0.064)	0.248 (0.059)	0.227 (0.054)	p=0.198 η_p^2=0.086	p=0.191 η_p^2=0.098	p=0.482 η_p^2=0.043
Force Time Integral (N*s)	Medial Midfoot[b,c]	23.9 (10.8)	26.6 (12.2)	35.0 (13.7)	24.6 (10.4)	26.5 (14.3)	43.4 (39.9)	38.1 (61.8)	50.4 (66.0)	63.3 (75.5)	p=0.779 η_p^2=0.026	**p<0.001* η_p^2=0.570**	p=0.725 η_p^2=0.019
	Lateral Midfoot	36.2 (11.7)	40.3 (14.3)	39.1 (9.9)	41.3 (16.6)	40.2 (19.1)	51.5 (40.4)	57.5 (87.5)	71.9 (88.9)	70.9 (89.9)	p=0.569 η_p^2=0.043	p=0.073 η_p^2=0.151	p=0.824 η_p^2=0.012
	Medial Forefoot[c,2]	23.0 (14.2)	23.3 (11.1)	17.4 (9.3)	34.8 (13.8)	35.2 (16.1)	32.9 (16.5)	60.7 (89.9)	76.7 (109.7)	55.5 (60.9)	p=0.184 η_p^2=0.089	**p=0.006* η_p^2=0.276**	**p=0.032* η_p^2=0.188**
	Middle Forefoot	27.4 (16.0)	30.7 (15.9)	26.6 (15.0)	45.1 (17.5)	46.9 (19.8)	47.6 (23.2)	83.4 (150.2)	106.8 (176.0)	77.5 (105.6)	p=0.636 η_p^2=0.037	p=0.072 η_p^2=0.151	**p=0.047* η_p^2=0.169**
	Lateral Forefoot	19.2 (8.5)	22.2 (11.6)	19.9 (9.0)	31.1 (12.8)	31.0 (15.8)	34.2 (17.7)	55.8 (101.1)	70.7 (111.5)	52.0 (74.1)	p=0.503 η_p^2=0.049	p=0.408 η_p^2=0.054	p=0.179 η_p^2=0.099

NB: no brace, SB: stabilizing brace, RB: rigid brace
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3 indicates significant difference between pes rectus and pes cavus

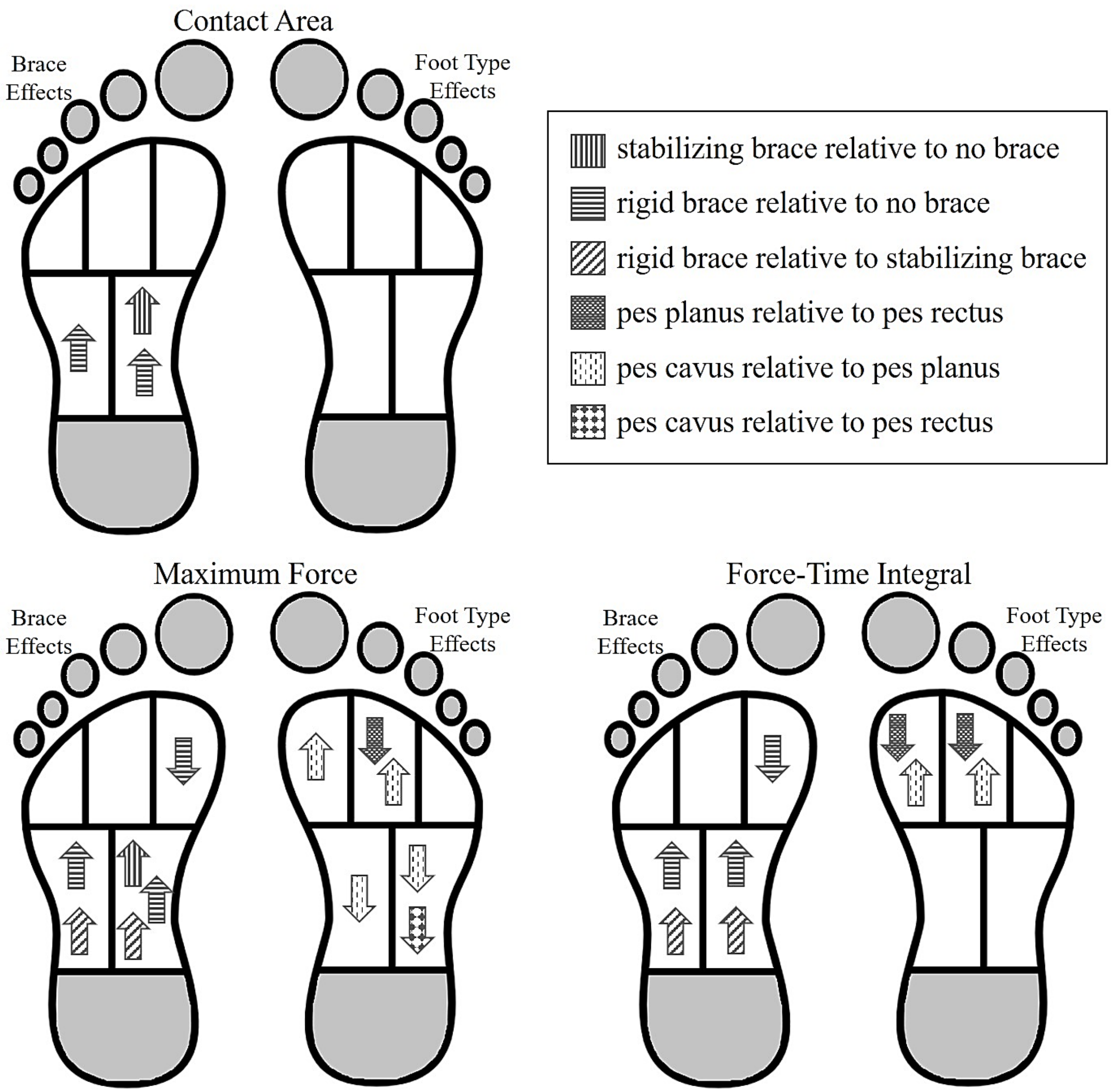

Figure 2.5: Results from a 3x3 mixed model repeated measures ANOVA and post-hoc testing comparing brace condition and foot type for contact area, maximum force, and force-time integral in the five regions of interest during the cut.

Table 2.3: Mean (SD) and results from a 3x3 mixed model repeated measures ANOVA comparing brace condition and foot type for contact area, maximum force, and force-time integral in the five regions of interest during the cut.

		Pes Planus			**Pes Rectus**			**Pes Cavus**					
		NB	**SB**	**RB**	**NB**	**SB**	**RB**	**NB**	**SB**	**RB**	**Interaction**	**ME Brace**	**ME Foot Type**
Contact Area (NICA)	Medial Midfoot[a,b]	0.153 (0.012)	0.155 (0.005)	0.160 (0.005)	0.154 (0.007)	0.156 (0.008)	0.158 (0.003)	0.154 (0.007)	0.155 (0.004)	0.157 (0.003)	p=0.826 η_p^2=0.022	**p=0.014*** **η_p^2=0.234**	p=0.976 η_p^2=0.001
	Lateral Midfoot[b]	0.161 (0.005)	0.160 (0.002)	0.163 (0.005)	0.158 (0.005)	0.156 (0.013)	0.161 (0.002)	0.159 (0.002)	0.158 (0.003)	0.160 (0.003)	p=0.917 η_p^2=0.014	**p=0.006*** **η_p^2=0.275**	p=0.257 η_p^2=0.079
	Medial Forefoot	0.076 (0.006)	0.077 (0.003)	0.073 (0.009)	0.077 (0.002)	0.077 (0.004)	0.076 (0.004)	0.077 (0.002)	0.076 (0.004)	0.077 (0.002)	p=0.429 η_p^2=0.056	p=0.186 η_p^2=0.100	p=0.426 η_p^2=0.050
	Middle Forefoot	0.092 (0.004)	0.092 (0.002)	0.093 (0.004)	0.091 (0.002)	0.092 (0.003)	0.091 (0.001)	0.092 (0.002)	0.092 (0.003)	0.092 (0.002)	p=0.055 η_p^2=0.129	p=0.276 η_p^2=0.077	p=0.526 η_p^2=0.038
	Lateral Forefoot	0.085 (0.002)	0.084 (0.001)	0.086 (0.003)	0.083 (0.004)	0.082 (0.008)	0.084 (0.003)	0.084 (0.002)	0.085 (0.002)	0.084 (0.002)	p=0.684 η_p^2=0.034	p=0.157 η_p^2=0.109	p=0.292 η_p^2=0.072
Maximum Force (BW)	Medial Midfoot[a,b,c,2]	0.366 (0.126)	0.386 (0.107)	0.448 (0.088)	0.289 (0.088)	0.301 (0.088)	0.407 (0.097)	0.224 (0.051)	0.240 (0.049)	0.348 (0.050)	p=0.094 η_p^2=0.112	**p<0.001*** **η_p^2=0.785**	**p=0.013*** **η_p^2=0.232**
	Lateral Midfoot[b,c,2,3]	0.344 (0.078)	0.352 (0.077)	0.407 (0.083)	0.298 (0.080)	0.275 (0.086)	0.344 (0.066)	0.203 (0.035)	0.214 (0.052)	0.268 (0.053)	p=0.107 η_p^2=0.107	**p<0.001*** **η_p^2=0.546**	**p=0.001*** **η_p^2=0.341**
	Medial Forefoot[b,2]	0.262 (0.061)	0.243 (0.067)	0.210 (0.069)	0.367 (0.126)	0.364 (0.142)	0.332 (0.125)	0.430 (0.141)	0.431 (0.142)	0.439 (0.129)	p=0.206 η_p^2=0.084	**p=0.043*** **η_p^2=0.178**	**p=0.004*** **η_p^2=0.282**
	Middle Forefoot[1,2]	0.272 (0.045)	0.265 (0.051)	0.268 (0.067)	0.347 (0.082)	0.363 (0.097)	0.360 (0.100)	0.373 (0.098)	0.366 (0.104)	0.350 (0.094)	p=0.543 η_p^2=0.045	p=0.850 η_p^2=0.010	**p=0.016*** **η_p^2=0.221**
	Lateral Forefoot	0.224 (0.056)	0.218 (0.054)	0.219 (0.046)	0.221 (0.063)	0.220 (0.065)	0.226 (0.064)	0.209 (0.049)	0.193 (0.057)	0.189 (0.052)	p=0.589 η_p^2=0.041	p=0.186 η_p^2=0.100	p=0.442 η_p^2=0.048
Force Time Integral (N*s)	Medial Midfoot[b,c]	34.6 (17.0)	34.6 (14.7)	48.7 (19.4)	29.6 (12.4)	32.2 (13.2)	46.0 (14.3)	26.5 (14.7)	26.3 (10.8)	42.8 (14.8)	p=0.541 η_p^2=0.045	**p<0.001*** **η_p^2=0.786**	p=0.495 η_p^2=0.042
	Lateral Midfoot[b,c]	35.7 (13.3)	35.7 (12.7)	41.5 (13.8)	33.3 (13.5)	32.0 (15.0)	37.6 (13.7)	26.1 (13.7)	25.8 (9.7)	31.1 (10.7)	p=0.716 η_p^2=0.031	**p<0.001*** **η_p^2=0.465**	p=0.239 η_p^2=0.083
	Medial Forefoot[b,1,2]	50.9 (23.9)	27.4 (15.3)	23.5 (11.2)	30.5 (18.1)	51.4 (24.0)	46.5 (22.1)	59.4 (31.4)	57.8 (27.5)	58.4 (27.4)	p=0.152 η_p^2=0.095	**p=0.015*** **η_p^2=0.230**	**p=0.005*** **η_p^2=0.276**
	Middle Forefoot[1,2]	30.6 (19.4)	28.6 (18.0)	31.3 (17.7)	46.1 (13.5)	49.4 (16.6)	49.2 (16.6)	49.7 (24.8)	46.7 (21.9)	47.2 (19.5)	p=0.338 η_p^2=0.066	p=0.777 η_p^2=0.016	**p=0.008*** **η_p^2=0.256**
	Lateral Forefoot	23.2 (12.7)	21.4 (11.3)	22.4 (10.1)	30.8 (10.5)	31.1 (11.4)	30.8 (11.1)	26.8 (13.2)	23.8 (11.0)	24.7 (11.4)	p=0.582 η_p^2=0.042	p=0.121 η_p^2=0.124	p=0.157 η_p^2=0.106

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A CMC analysis of the COP displacement curves revealed that the traces of the COP were highly correlated across brace conditions (Table 2.4). This is evidenced by mean CMC values being close to 1 for all foot types in both the walking and running tasks. However, a SPM analysis using a mixed model repeated measures ANOVA revealed that brace condition resulted in significantly different COP displacement for certain portions of the COP trace during both the walking and running tasks (p<0.05) (Table 2.5, Figure 2.6). Neither foot type nor the foot type by brace interaction had a significant effect on any portion the COP curves.

Table 2.4: Correlation of Multiple Coefficient (CMC) outputs (mean ± SD) for a comparison of COP curves among brace conditions. Values close to 1 indicate similar curves while values close to 0 indicate dissimilar curves.

		Medial-Lateral COP	**Anterior-Posterior COP**
Walk	*Pes Planus*	0.933 ± 0.071	0.991 ± 0.005
	Pes Rectus	0.948 ± 0.050	0.992 ± 0.009
	Pes Cavus	0.955 ± 0.042	0.989 ± 0.007
Run	*Pes Planus*	0.958 ± 0.040	0.992 ± 0.008
	Pes Rectus	0.934 ± 0.074	0.987 ± 0.022
	Pes Cavus	0.961 ± 0.042	0.979 ± 0.034

Table 2.5: Statistical Parametric Mapping (SPM) outputs of a 3x3 repeated measures ANOVA, indicating the regions of difference in COP traces based on each factor.

		Regions of Difference			
		Foot Type	*Brace Condition*		*Interaction*
Walk	*Medial-Lateral COP*	N/A	0-76.6%	p=0.001	N/A
	Anterior-Posterior COP	N/A	0-45.5% 63.9-74.6% 92.9-98.0%	p=0.007 p=0.045 p=0.049	N/A
Run	*Medial-Lateral COP*	N/A	0-3.0% 20.9-71.0%	p=0.050 p=0.021	N/A
	Anterior-Posterior COP	N/A	0-29.2%	p=0.033	N/A

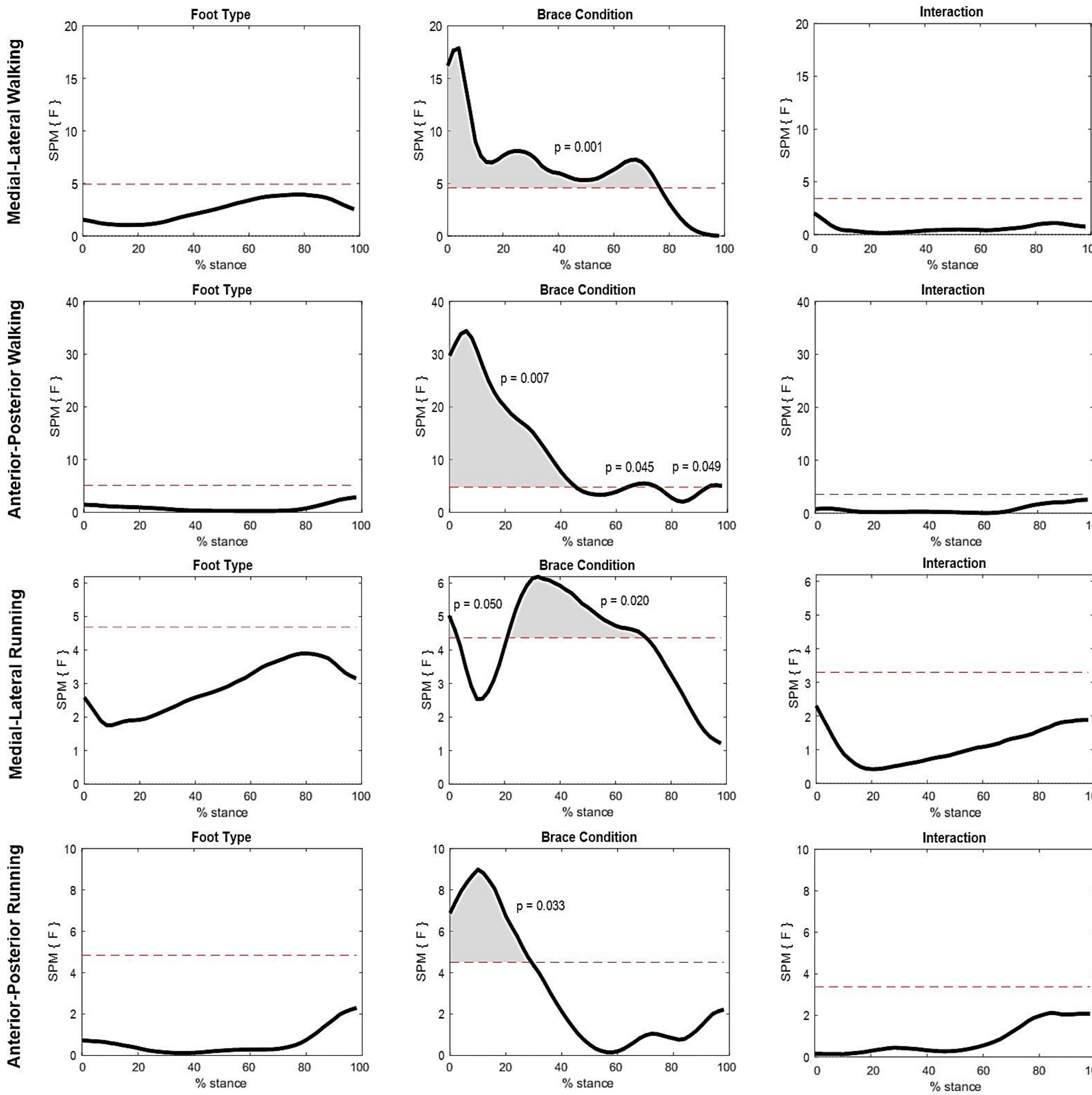

Figure 2.6: Graphical representation of SPM analysis of COP as a percent of the stance phase, with significant effects indicated by the shaded regions and associated p-values.

A Wilcoxon signed-rank test between comfort ratings of the stabilizing brace and the rigid brace revealed that participants found the stabilizing brace significantly more comfortable than the rigid brace (p=0.026). However, when broken down by foot type classification, there were no significant differences in comfort ratings between the two braces.

Discussion

The purpose of this study was to determine the effects of two ankle braces on plantar loading between different foot types during three movement tasks. This was accomplished through the analysis of plantar loading variables in five different regions of the foot during walking, running, and cutting. It was hypothesized that ankle braces would help to reduce irregular plantar loads caused by abnormal foot postures by shifting plantar loading patterns medially in pes cavus individuals and laterally in pes planus individuals.

The present study found that foot type did affect the plantar loading variables of maximum force and the force-time integral (FTI), largely independent from brace. Specifically, maximum force was significantly affected by foot type during walking, running, and cutting in the medial midfoot, medial forefoot, and middle forefoot, and force in the lateral midfoot was affected by arch during the cutting tasks. Post hoc testing revealed that most of these differences were between pes planus and pes cavus individuals. This is in agreement with previous studies, which have found that maximum force in the medial midfoot is higher for pes planus individuals and lower for pes cavus individuals and that pes planus individuals have reduced forces in the forefoot [6], [7], [35]. The present study also found that the FTI was affected by foot type in the medial forefoot during running and cutting, and in the middle forefoot during cutting. This agrees with previous work, which has reported FTI to be lower in the forefoot of planus individuals and higher in the forefoot of cavus individuals [6], [7], [36]. However, previous research has also found that FTI in the midfoot and lateral forefoot is affected by different foot types, which was not found in the present study [6], [7], [36]. Furthermore, the present study did not find any differences in contact area due to foot type for any region of the foot during any task. This does not agree with previous literature, which reports that pes planus individuals have increased contact area in the medial midfoot and lateral forefoot, and that pes cavus individuals also have increased contact area in the lateral forefoot when compared to the normal foot [6], [7], [35]. This could be because the present study had individuals in the normal category that exhibited contact area patterns that resembled the patterns of a planus or cavus foot, or because the brace conditions resulted in fewer differences between foot types. Despite inconsistent agreement with other studies, the present study detected differences in force distribution due to foot type during different athletic tasks. This could contribute to the increased risk of injury that is related to abnormal foot postures [2], which highlights the need for a prevention mechanism.

One mechanism that could be used to prevent lower extremity injuries by reducing irregular plantar loads are ankle braces. The present study also examined the effects of a stabilizing brace and a semi-rigid brace on plantar loading during various athletic tasks. These results revealed that the brace increased the contact area in the midfoot during walking and cutting and decreased the contact area in the medial and middle forefoot during walking independent of foot type. Ankle braces also increased the maximum force in the medial and lateral midfoot during walking, running, and cutting, and decreased maximum force in the medial forefoot during all tasks. Finally, FTI was increased by brace wear in the medial midfoot during all three tasks and was increased in the lateral midfoot and decreased in the medial forefoot during cutting. Overall, contact area, maximum force, and the FTI were increased in the midfoot and decreased in the forefoot when braced, largely independent from foot type. While this does not support the original hypothesis of a medial or lateral shift, it indicates that braces do cause a redistribution of plantar loads. Post hoc testing revealed that most of these differences occurred when wearing the semi-rigid brace when compared to both the unbraced and the stabilizing brace condition. While no previous studies have examined plantar loading differences due to ankle braces, other studies have examined the impact of ankle taping and found inconsistent results on whether it does [13], [19] or does not [22] affect plantar loading. While bracing and taping have been shown to reduce the risk of lower extremity injuries [8], [9], [17], ankle bracing is more effective. This could be due to the redistribution of plantar pressures, as evidenced by the present study.

Furthermore, in addition to regional differences in contact area, maximum force, and FTI, braces were also shown to alter center of pressure (COP) distribution. The COP was analyzed in the AP and ML directions, as is commonly done in studies of this nature [49], [50], since curves were being compared across multiple conditions. While a CMC analysis revealed that the COP distribution was highly correlated across the three brace conditions, the SPM analysis identified portions of the COP distribution that were affected by brace-wear. In the AP direction, the brace affected the beginning of the COP trace in both the walk and run, with a larger portion of the curve being altered during walking. This indicates that the majority of the brace effects in the AP direction were at heel strike, suggesting that participants contacted the ground with a different heel strike angle while wearing the brace. In the ML direction, the brace altered the middle portion of the curve during both walking and running, with a larger portion of the curve being affected by braces during walking. This shows that most of the ML changes due to brace occurred at midstance, which is likely a result of the brace controlling the pronation and supination of the foot during the highest weight bearing phase. The ML shifts are important to consider due to the fact that pes planus individuals have been shown to have a more medial COP while pes cavus individuals have a more lateral COP when compared to someone with a normal foot posture [6]. The fact that braces may correct the abnormal COP displacement in pes planus and pes cavus individuals in combination with the fact that braces are easily adjustable, can be used without an athletic trainer, and are reusable [9], [17] support further research into the possibility of using ankle braces as an injury prevention method.

There were a few limitations in the present study. The first is that there were an unequal number of participants for each foot type group, with 8 participants being pes planus, 17 being normal, and 11 being pes cavus. While this is expected based on the fact that a larger portion of the population has normal arches [1], a more equally distributed sample could have resulted in a more robust analysis. A second limitation is that the AirLift PTTD brace was designed with an inflatable pocket under the arch, which was not inflated during the present study. This was done to prevent variations in inflation levels between participants from biasing the results, but is not standard for the intended use of the brace. One final limitation in this study is the fact that not all participants had cutting experience, despite this being an inclusion criterion. Furthermore, cutting speed for each participant was not standardized between trials. This could have resulted in speed variations that affected the plantar loading measures; however this risk was minimized by averaging plantar loading values across multiple trials. Nevertheless, these aspects should be considered in future studies.

In conclusion, this study provides one of the first analyses of the effects of ankle braces, in combination with foot type, on plantar loading variables. It was determined that foot type did affect plantar forces in the foot independent of brace wear, which agrees with previous work. Additionally, it was found that ankle braces, specifically rigid braces, can significantly alter COP and plantar forces within the midfoot and forefoot, independent of foot type. While no conclusions could be drawn on the use of ankle braces for prevention of injuries related to foot type, the results of this study support the use of braces for redistribution of plantar loads. Therefore, ankle braces could be used to prevent some overuse injuries and additional research should be done to further quantify the effects of ankle braces on foot type.

Conclusions

Foot type can affect lower extremity kinematics, plantar loading, and contribute to injury risk. There are many different aspects of foot posture, including foot mobility, rearfoot angle, and arch height [3], [6], [29]. Arch height is one of the most commonly used methods to determine foot posture and allows an individual to be classified as pes planus (flat-footed or low arched), pes rectus (normal), or pes cavus (high arched). Previous research has found that about 60% of the population is normal, 20% has flat feet, and 20% has high arches [1]. There are many different measurements that can be used to classify arch height, including the medial longitudinal arch, navicular height and navicular drop, longitudinal arch angle, rearfoot angle, arch height index, or foot print measures [10]–[12], [26], [29]. However, current technologies to measure these aspects of foot posture are expensive and not easily accessible to smaller clinics or research settings.

Therefore, the first purpose of this study was to develop a new inexpensive device that could accurately measure foot posture characteristics. As a result, the Foot Posture Measurement System (FPMS) was built for less than $100 and can measure total foot length, truncated foot length, foot width, dorsum height, and navicular height. This system was validated against a 3D scanner and was shown to have excellent validity (ICC=0.908-0.994) and good to excellent between day repeatability (ICC=0.867-0.996). This system can therefore be used as an inexpensive method to accurately and reliably measure foot posture characteristics in clinics or research labs.

One reason that foot posture classification is so important is due to its known relationship to lower extremity injuries [2]. More specifically, pes planus individuals are more susceptible to soft tissue injuries or injuries on the medial sides of the legs due to the increased mobility in the foot [3], [4]. On the other hand, pes cavus individuals are more prone to bone injuries or lateral leg injuries due to increased rigidity and reduced shock absorption in the foot [3], [4]. Ankle braces and ankle taping are both commonly used during athletic activities to reduce the occurrence and severity of these injuries [8], [9]. While taping has the advantage of a more personalized fit [17] and has been shown to control arch height and plantar loads [13], [19]–[22], its usefulness may be limited due to the fact that it loosens over time and must be applied by an athletic trainer [9], [13], [17]. Ankle bracing may be a suitable alternative, however no research has been done into whether it can effectively control arch height and plantar loads during athletic activity.

As a result, the second purpose of this research was to examine the effects of different ankle braces and foot type on plantar loading variables during different activities. Contact area, maximum force, force-time integral, and center of pressure were measured during walking, running, and cutting. The results of this analysis revealed that foot type did have a significant effect on forces in the foot ($p=0.003$-0.033), as expected. Furthermore, it was revealed that ankle braces did alter contact area ($p=0.001$-0.0011), maximum force ($p<0.001$ – $p=0.006$), force-time integral ($p<0.001$ – $p=0.040$), and center of pressure ($p=0.001$-0.050) during all athletic tasks, independent of brace condition. Specifically, the forces were reduced in the forefoot and increased in the midfoot independent of foot type, and the semi-rigid brace had a greater effect on these variables than a lace-up stabilizing brace. Nevertheless, this indicates that ankle braces are a viable option for redistributing plantar loads in the foot and possibly preventing some lower extremity injuries, regardless of foot type.

Despite these promising findings, there is still research that remains to be done in the field. First, as this was the first study of its kind, a larger study on this topic would be beneficial in order to confirm these conclusions and expand upon their generalizability. This study could include additional types of ankle braces and should look at their effects on both plantar loading and static foot posture. Second, while it has been reported that bracing has a limited effect on other joints in the lower extremity [17], further research on this topic would be beneficial to ensure that this recommendation has no other harmful side effects. Nevertheless, the present study provided novel information regarding the effects of ankle braces on foot posture and plantar loading patterns.

The findings from the present study also helped to provide direction for future brace development that could bring relief to individuals suffering from injuries due to abnormal foot conformation. The fact that the rigid brace had a more significant effect on plantar loading patterns indicates that this design should serve as the foundation for future ankle braces. However, this brace was also found to be less comfortable than the stabilizing brace, and many participants commented that it was extremely bulky. When considering the stabilizing brace, participants favored the support provided from the figure-eight strapping but did not like the fact that the laces made it difficult to get on and off. Therefore, future braces should incorporate figure-eight strapping with slimmer rigid supports in order to appropriately support irregular foot postures with the goal of preventing injuries. These rigid stabilizers should be able to support up to 56 kg in order to properly protect the ligaments in the ankle [24]. Furthermore, the fastening mechanisms should largely consist of straps, both elastic and inelastic, in order to also improve brace-wear compliance. The combination of these aspects should appropriately balance both comfort and support for an improved ankle brace design.

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