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Segmental foot and ankle kinematic differences between rectus, planus, and cavus foot types

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ABSTRACT

The presence of multiple foot types has been used to explain the variability of foot structure observed among healthy adults. These foot types were determined by specific static morphologic features and included rectus (well aligned hindfoot/forefoot), planus (low arched), and cavus (high arched) foot types. Unique biomechanical characteristics of these foot types have been identified but reported differences in segmental foot kinematics among them has been inconsistent due to differences in neutral referencing and evaluation of only select discrete variables. This study used the radiographically-indexed Milwaukee Foot Model to evaluate differences in segmental foot kinematics among healthy adults with rectus, planus, and cavus feet based on the true bony alignment between segments. Based on the definitions of the individual foot types and due to conflicting results in previous literature, the primary study outcome was peak coronal hindfoot position during stance phase. Additionally, locally weighted regression smoothing with alpha-adjusted serial *t*-test analysis (LAAST) was used to compare these foot types across the entire gait cycle. Average peak hindfoot inversion was $-1.6^\circ \pm 5.1^\circ$, $6.7^\circ \pm 3.5^\circ$, and $13.6^\circ \pm 4.6^\circ$, for the Planus, Rectus, and Cavus Groups, respectively. There were significant differences among all comparisons. Differences were observed between the Rectus and Planus Groups and Cavus and Planus Groups throughout the gait cycle. Additionally, the Planus Group had a premature peak velocity toward coronal varus and early transition toward valgus, likely due to a deficient windlass mechanism. This assessment of kinematic data across the gait cycle can help understand differences in dynamic foot function among foot types.

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1. Introduction

The foot's complex structure leads to morphologic variability. Even among the healthy population, foot structure varies widely. Root et al. described the following foot types in the healthy adult population based on specific morphologic features: rectus (well aligned hindfoot and forefoot), planus (low arch, valgus hindfoot and/or varus forefoot, and cavus (high arch, varus hindfoot and/or valgus forefoot) (Root et al., 1971). Cavus foot types have been associated with clawfoot deformity (Sugathan & Sherlock, 2009) and hammertoes (De Doncker & Kowalski, 1970). Planus feet are associated with hallux valgus, hallux rigidus, and posterior tibial tendon dysfunction (Ledoux et al., 2003, Erol et al., 2015). Both planus and cavus foot types are also at increased risk of

stress fractures (Levy et al., 2006, Kaufman et al. 1999, Matheson et al. 1987).

Further investigation identified that these foot types not only had characteristic static morphologic features but also unique biomechanical characteristics during ambulation (Burns et al., 2005; Hillstrom et al., 2013; Ledoux & Hillstrom, 2002). Plantar pressure studies demonstrated differences in foot loading metrics such as the center of pressure excursion index (CPEI) (Diaz et al., 2018). Findings indicated planus feet maintain a more medial CPEI while cavus feet have a more lateral CPEI than rectus (Hillstrom et al., 2013). Individuals with planus feet also presented with higher medial pressures and forces than individuals with cavus feet.

With different anatomical structure and loading patterns during gait, different segmental foot kinematics among foot types are expected; however, results have been inconsistent and in some cases, conflicting. Buldt et al. (2015) concluded individuals with cavus feet presented with greater peak hindfoot eversion and peak midfoot abduction at initial contact when compared to individuals

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with rectus or planus feet. In contrast, [Houck et al. \(2008\)](#) identified that individuals with planus feet presented with greater peak hindfoot eversion during stance when compared to individuals with “normal” feet.

One reason for these conflicts may be the methods of neutral referencing. Reliable and consistent methods of neutral referencing are crucial for repeatable kinematic results. Additionally, accurate marker placement is critical to establish a mathematical relationship between surface markers and the underlying skeletal anatomy. This becomes problematic when modeling segments such as the hindfoot as reliable bony landmarks do not exist on the calcaneus. Bone-based referencing via the Milwaukee Foot Model (MFM) addresses these limitations by referencing marker locations to actual bony anatomy using radiographic measurements ([Kidder et al., 1996](#)). A literature review of segmental foot models concluded that of all published models, the MFM was one of only two that demonstrated external validity in the literature ([Bishop et al., 2012](#)). The MFM was validated for use in pediatric and adult populations ([Long et al., 2011](#); [Myers et al., 2004](#)) and assessment of these radiographic measures showed excellent repeatability ([Davids et al., 2005](#)).

Another reason for inconsistent results across studies is limiting primary variables of interest to discrete values (e.g. peak max/min) across the gait cycle. There is a potential loss of information when only analyzing a single datapoint across an entire time series. During a normal gait cycle, ankle kinematics are split into three rockers ([Brockett & Chapman, 2016](#)). Changes to rocker function are often used to evaluate foot pathology but kinematic differences between these rockers may not be appreciated when only comparing kinematic peaks. Statistical methods have recently been used to compare biomechanical data over continuous time series ([Niiler, 2017](#); [Pataky, 2010](#)).

Recent model-based analysis of the calcaneus has shown that the orientation of the hindfoot is highly variable and markers placed posterior calcaneal tuberosity alone, as is the case with most segmental foot models, may not represent the orientation of the body of the calcaneus ([Zavatsky et al., 2019](#)). This inability to accurately represent calcaneal alignment is likely the cause of variability in hindfoot kinematics among previous studies. [Zavatsky et al.](#) recommended imaging of the foot could be used to realign the marker-based coordinate systems with those of the bones. Therefore, the purpose of this work was to compare segmental foot kinematic data among healthy adult individuals with rectus, cavus and planus foot types using the MFM. The use of this radiographically-indexed model allows for differences in bony alignment to be accurately measured. The primary outcome measure for the current study was peak coronal hindfoot position during stance. This variable was chosen based on definitions of the individual foot types ([Root et al., 1971](#)) and due to the conflicting results from previous segmental foot model studies among foot types ([Buldt et al., 2015](#), [Houck et al., 2008](#)). It was hypothesized that the Planus Group would present with greater stance phase peak hindfoot eversion; whereas the Cavus Group would present with greater stance phase peak hindfoot inversion. Additionally, the current work employed novel statistical strategies to compare kinematics across the entire gait cycle. Such results will demonstrate if there are not only differences in static foot morphology but also the dynamic characteristics of the kinematic curves during locomotion which is not possible when only select datapoints are analyzed across an entire time series.

2. Methods

Volunteer subjects were recruited who reported no symptoms or history of foot problems. The sample population included 33

asymptomatic, healthy adults (Cavus Group: n = 11, Female: 5, Average Age: 24.1 yrs, SD: 4.7 yrs, Rectus Group: n = 11, Female: 8, Average Age: 26.0 yrs, SD: 3.5 yrs, Planus Group: n = 11, Female: 3, Average Age: 24.3 yrs, SD: 5.3 yrs). Study numbers were based on previous segmental foot kinematic studies ([Levinger et al., 2010](#); [Houck et al., 2008](#)). Consent was obtained from all participants using an IRB approved protocol. Subjects were assigned to respective groups following enrollment. As more subjects were tested and initial findings suggested differences, enrollment was targeted to ensure a balanced distribution of participants among foot type groups. Foot type was determined using measures of resting calcaneal stance position (RCSP) and forefoot to rearfoot relationship (FF-RF) ([Hillstrom et al., 2013, Table 1](#)). RCSP values obtained from clinical examination were confirmed using measurements from static weight-bearing radiographs. Clinical examination was performed for all subjects by a licensed physical therapist.

All individuals participated in an instrumented gait analysis which included segmental foot analysis using the MFM ([Long et al., 2011](#)). A 14-camera Vicon system (Oxford Metrics, UK) was used to record motion data at 120 frames/second. During the static trial, subjects were instructed to stand comfortably on a piece of cardboard. Their foot position was traced to create a foot positioning template. Participants were instructed to walk “at a comfortable walking speed” with 10–15 trials collected and three randomly chosen for processing. Once the representative trials were processed, the resulting kinematics were visually inspected by a senior investigator to identify any data contamination. To evaluate consistency of included trials for each participant, a robust scoring (Rscore) algorithm ([Sangeux & Polak, 2015](#)) was also employed to identify outliers. If contaminated data or outliers were identified, another trial was randomly chosen and processed.

Following gait data collection, anterior-posterior, lateral view, and modified hindfoot coronal alignment weight-bearing radiographs of the foot were captured while standing on the foot positioning template. The MFM calibrates neutral positions (zero positions) based on absolute bony alignment and relies on measures of angles between adjacent bone segments as measured from these weightbearing radiographs. Based on these measures, the orientation of bone-based axes in the global space is established. The use of these radiographic offset angles doesn't require a “zero reference level” to be established by neutral referencing, rather it allows for calculation of kinematics based on the true bony alignment between segments.

The motion data and radiographic offset measurements were input into a custom software model ([Sullivan, 2019](#)). To identify gait cycles, two consecutive foot strike events were identified from force plate strikes. Consistent with Vicon's Nexus software used in clinical gait analysis, those points were down sampled to 51 points and reported as percentage of gait cycle. Kinematics were calculated for the tibia relative to the global coordinate system (tibia angle), hindfoot relative to tibia (hindfoot angle), forefoot relative to hindfoot (forefoot angle).

The average of the three representative trials kinematic curves was calculated for each subject. To test the hypothesis that the Planus Group would present with greater peak hindfoot eversion; whereas the Cavus Group would present with greater peak

Table 1
Hindfoot and forefoot alignment used to identify foot types.

Foot type identification	
Planus	RCSP $\geq 4^\circ$ valgus OR FF-RF $\geq 4^\circ$ varus
Rectus	$0^\circ \leq$ RCSP $\leq 2^\circ$ valgus AND $0^\circ \leq$ FF-RF $\leq 4^\circ$ varus
Cavus	RCSP $\geq 0^\circ$ varus AND FF-RF $\geq 1^\circ$ valgus

hindfoot inversion, an analysis of variance (ANOVA) was performed to evaluate the effect of foot type on kinematic peaks. When a significant group effect was identified based on the omnibus F-test, pair-wise comparisons using Tukey-HSD were performed to evaluate differences between foot types ($\alpha < 0.05$).

For the evaluation of kinematic curves, between group comparisons were assessed using the method of LAAS (Niiler, 2017). LAAS uses locally weighted scatterplot smoothing, serial correlation to make alpha adjustments, and point-wise Welch's t-tests to make comparisons between groups throughout the gait cycle. To correct for correlation in adjacent P-values, a new adjusted α (α') based on the Bonferroni method was introduced. In this case, α' is:

$$\alpha' = 0.05N(1 - \rho^2) + \rho^2$$

where ρ is the correlation coefficient between a time series time-shifted by one point and itself. By this formula, data that are perfectly correlated will have no α adjustment, while those with zero correlation receive full Bonferroni correction (Nicholson et al., 2018). P values less than α' indicated that the kinematics for the two groups being compared were statistically different at that time point. This analysis was performed using custom scripts in MATLAB (Mathworks, Natick, MA).

3. Results

Post-hoc power analysis showed the sample size of 11 subjects per group was sufficient to power the study to >80% power for all comparisons (Table 2, Chow et al., 2008). Results of the ANOVA showed an effect of group and the post-hoc Tukey tests showed differences among all comparisons of our primary outcome variable.

The average segmental kinematics of each group are plotted in Figs. 1–3. Heat maps below each plot were used to show where significant kinematics differences were identified among individual groups. The color gradient in each heat map provides an indicator of the magnitude of the p value when compared to α' . White bands indicate where p values were not less than the adjusted alpha values. As the color darkens, the calculated p value becomes lesser than the adjusted alpha and represent greater differences between groups.

Tibia kinematics showed the tibia progresses anteriorly during stance with a peak anterior tilt occurring at the transition between stance and swing (Fig. 1). In the coronal plane, the tibia is mildly adducted at initial contact and processes toward a more neutral position throughout midstance. It rapidly progresses into abduction during terminal stance and then adducts through swing. In the transverse plane, it progresses into internal rotation following initial contact and progresses into external rotation from terminal stance through mid-swing.

When compared to the Rectus Group, the Cavus Group presented with less external rotation throughout stance ($p < 0.025$). Only minor differences were identified in the sagittal and coronal planes. No differences were identified between Rectus and Planus Groups.

Table 2

Peak coronal position of the hindfoot and forefoot during stance for Rectus, Planus, and Cavus Groups. Post-hoc power analysis using group means, pooled standard deviation, and sample size showed sufficient power for each comparison (Chow, Shao, & Wang, 2008).

Foot type	Peak Hindfoot inversion during stance, Mean (S.D.)
Rectus	6.7° (3.2°)*^
Planus	-1.6° (5.1°)**
Cavus	13.6° (4.3°)^*

* Rectus vs Planus ($p < 0.05$), Pooled S.D. = 4.4°, Power = 98%.

^ Rectus vs Cavus ($p < 0.05$), Pooled S.D. = 4.1°, Power = 95%.

* Cavus vs Planus ($p < 0.05$), Pooled S.D. = 4.8°, Power > 99%.

Hindfoot kinematics showed the hindfoot was dorsiflexed relative to the tibia throughout the gait cycle (Fig. 2). Similar to ankle kinematics reported in single segment models, the hindfoot progresses towards plantarflexion following initial contact as the foot becomes plantigrade. It then dorsiflexes throughout mid-stance as the tibia advances over the plantigrade foot. The hindfoot then rapidly plantarflexes at terminal stance and recovers into dorsiflexion through swing. In the coronal plane, the hindfoot of the Rectus Group is in a relatively neutral position at initial contact and gradually progresses into slight eversion during mid-stance then progresses into inversion during terminal stance. In the transverse plane, the hindfoot is slightly externally rotated with minimal excursion.

In the sagittal plane, the Cavus Group presented with a dorsiflexion shift of the hindfoot throughout the gait cycle compared to the Rectus Group ($p < 0.004$) while the Planus Group presented with a shift toward plantarflexion throughout stance ($p < 0.008$). In the coronal and transverse planes when compared to the Rectus Group, hindfoot kinematics shifted toward inversion among the Cavus Group ($p < 0.004$) and eversion and external rotation among the Planus Group ($p < 0.011$ and $p < 0.010$, respectively).

Relative the hindfoot, the forefoot is plantarflexed throughout stance (Fig. 3). There is a rapid increase in plantarflexion during terminal stance followed a partial rebound towards dorsiflexion during swing. In the coronal and transverse planes of the Rectus Group, the forefoot is in varus/adduction at initial contact and gradually progresses into slight valgus/toward abduction during mid-stance as the first metatarsal approximates with the support surface. The forefoot then progresses into varus and increased adduction during terminal stance.

In the sagittal plane, the Cavus Group presented with a plantarflexion shift of the forefoot throughout the gait cycle compared to the Rectus Group ($p < 0.006$) while the Planus Group presented with a shift toward dorsiflexion ($p < 0.008$). In the coronal and transverse planes when compared to the Rectus Group, forefoot kinematics are shifted toward valgus and adduction among the Cavus Group ($p < 0.017$ and $p < 0.006$, respectively) and varus among the Planus Group ($p < 0.018$).

Due to observed difference in both timing and amplitude of forefoot kinematics in the coronal plane, a secondary analysis of velocity was performed to further evaluate kinematic curve morphology among groups (Fig. 4). Forefoot velocities showed that the Planus Group had a premature peak velocity toward varus and early transition toward valgus when compared to the Rectus and Cavus Groups ($p < 0.002$).

4. Discussion

This study compared segmental foot kinematics among rectus, planus, and cavus feet using the radiographically-indexed MFM. Our results showed triplanar differences in hindfoot and forefoot motion among these foot types. The Cavus Group showed increased dorsiflexion and inversion in the hindfoot and increased plantarflexion, valgus, and adduction in the forefoot when compared to the Rectus Group. The Planus Group had less dorsiflexion, more eversion, and more external rotation in the hindfoot as well as less plantarflexion and increased varus in the forefoot when compared to the Rectus Group. Understanding differences in function among foot types is important given the increased injury risk associated with planus and cavus foot types (Levy et al., 2006; Sugathan & Sherlock, 2009).

Past literature stressed that in order to systematically study foot pathologies, objective measures of foot structure and motion sensitive to different foot types are needed. A previous study reported differences in foot structure and dynamic pronation (Hillstrom

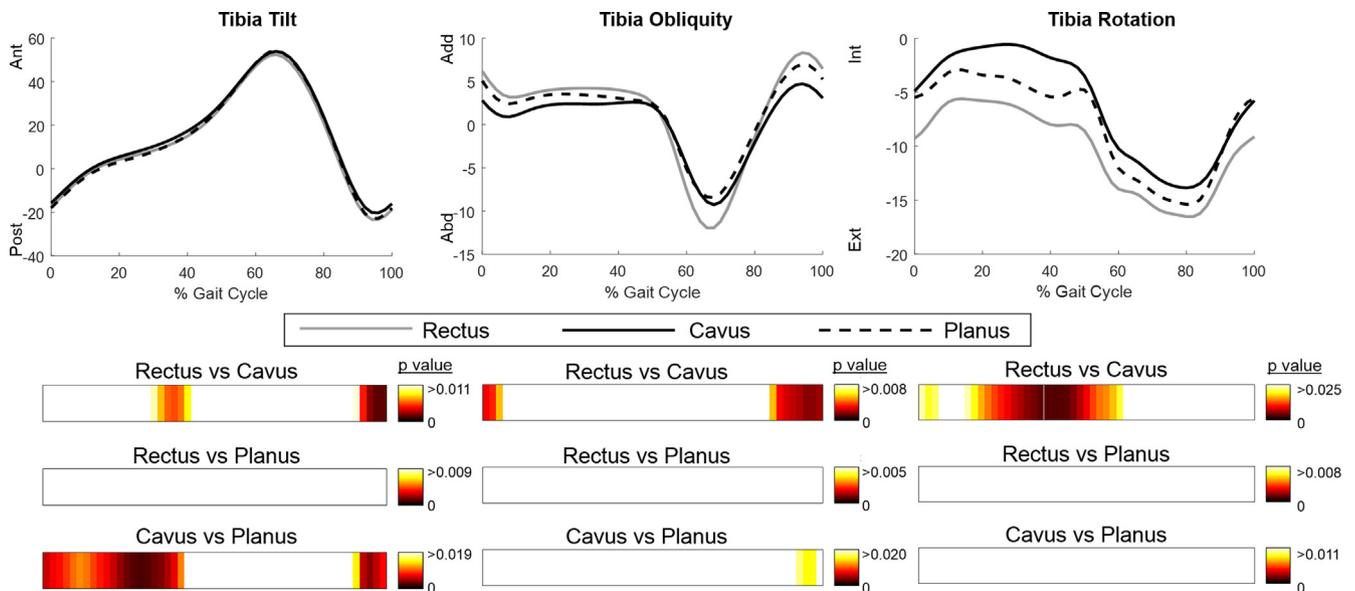


Fig. 1. Mean sagittal, coronal, and transverse tibia kinematics (with respect to the laboratory (global) coordinate axes) among the Rectus, Planus, and Cavus Groups. Heat maps represent the period of the gait cycle where significant differences were identified (darker colors indicate greater difference). The adjusted alpha was used as the upper bound of each heat map.

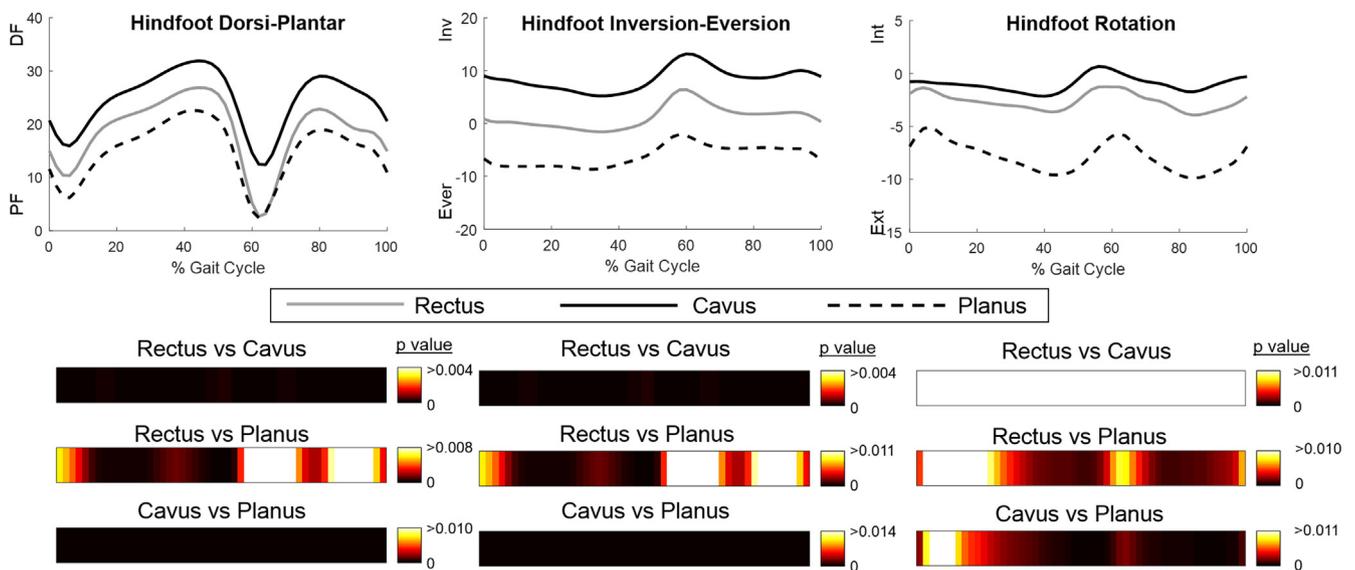


Fig. 2. Mean sagittal, coronal, and transverse hindfoot kinematics (with respect to the tibia) among the Rectus, Planus, and Cavus Groups. Heat maps represent the period of the gait cycle where significant differences were identified (darker colors indicate greater difference). The adjusted alpha was used as the upper bound of each heat map.

et al., 2013). While this previous study reported that gait pattern parameters (e.g. temporal-spatial parameters) were invariant across foot types, segmental foot kinematics were not compared.

Increased peak coronal hindfoot eversion in the Planus Group is consistent with previous static comparisons which reported significantly larger RCSP when compared with rectus and cavus feet (Hillstrom et al., 2013). Differences in segmental foot kinematics among foot types have been reported but with conflicting results. In the sagittal plane, Levinger et al. reported the forefoot of individuals with flat arches to be more plantarflexed than those with normal arches (Levinger et al., 2010) while Buldt et al. reported no sagittal differences (Buldt et al., 2015). This contradicts our results which showed the forefoot of the Planus Group to be in less plantarflexion than the Rectus and Cavus Groups.

Coronal plane motion and alignment has also been variable among healthy adults with some studies reporting no difference (Levinger et al., 2010; Powell et al., 2011) and others reporting increased inversion among cavus feet (Buldt et al., 2015; Powell et al., 2011). The current work demonstrated coronal forefoot differences among the foot types. In the coronal plane when compared to the Rectus Group, forefoot kinematics shifted toward valgus in the Cavus Group and varus in the Planus Group. Differences in transverse plane motion were also reported by Buldt with increased adduction of both the hindfoot and forefoot of the cavus foot type and abduction of both segments in the planus foot type. Our results were consistent with these alignment differences although differences were only observed in the forefoot of the Cavus Group and the hindfoot of the Planus Group. Although the

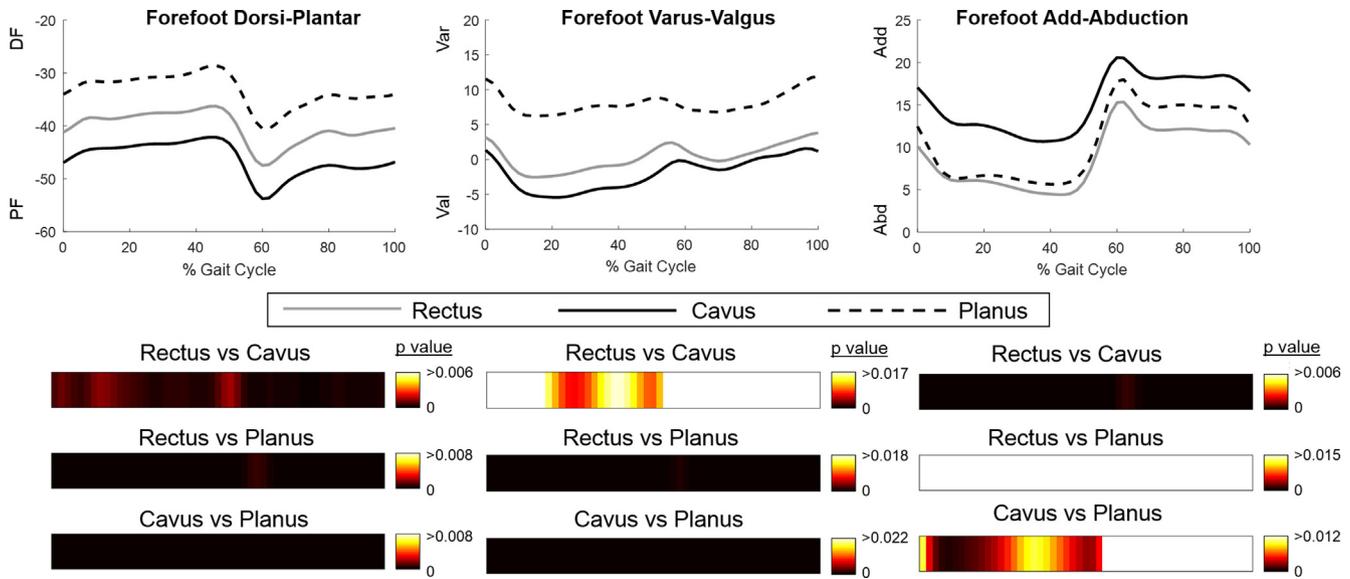


Fig. 3. Mean sagittal, coronal, and transverse forefoot kinematics (with respect to the hindfoot) among the Rectus, Planus, and Cavus Groups. Heat maps represent the period of the gait cycle where significant differences were identified (darker colors indicate greater difference). The adjusted alpha was used as the upper bound of each heat map.

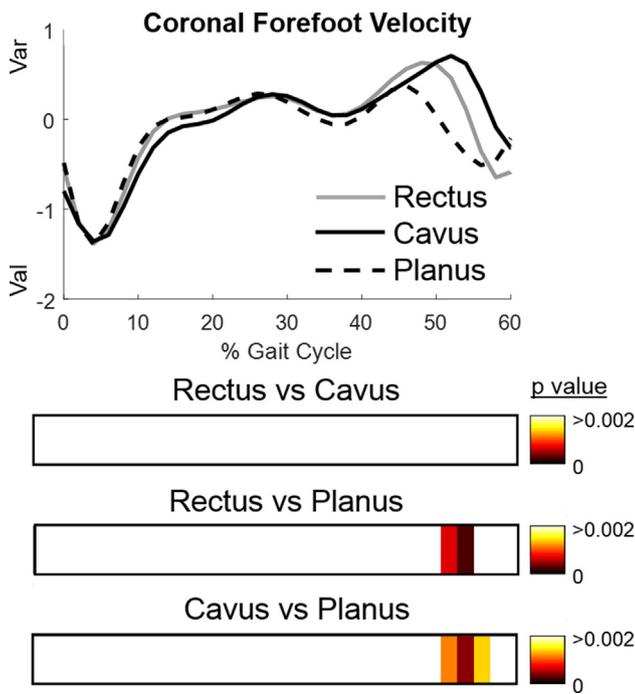


Fig. 4. Mean coronal forefoot angular velocities (with respect to the hindfoot) among the Rectus, Planus, and Cavus Groups across the stance phase of the gait cycle (0–60%). Heat maps represent the period of the gait cycle where significant differences were identified (darker colors indicate greater difference). The adjusted alpha was used as the upper bound of each heat map.

above studies used similar criteria to assign participants to respective foot type groups, each study used different methods for calculating segmental foot kinematics. Specifically, each study employed different strategies for neutral referencing.

It has been stressed that differences in kinematics between studies are likely due to the different methods used for neutral referencing (Buldt et al., 2015). Use of comfortable standing has been justified due to the difficulties in placing subjects in a standardized reference posture (e.g. subtalar neutral) and because this position many not be reached during walking (Buldt et al., 2015; McPoil

& Hunt, 1995; Menz, 1995). The use of a “zero reference level” from neutral referencing achieved via either comfortable standing or subtalar neutral is limited because it removes joint alignment differences which are frequently used as part of clinical assessment and should be included in the analysis (Leardini & Caravaggi, 2017).

Additionally, most studies of segmental foot kinematics rely on standard skin markers on the foot which may not accurately represent the orientation of the underlying bony anatomy (Zavatsky et al., 2019; Kruger et al., 2017). This is especially evident in the hindfoot because the calcaneus lacks easily identifiable landmarks, making repeatable marker placement and thus, accurate measurement of kinematics, difficult. The MFM used in the current work addresses these limitations of existing segmental foot models by using radiographic offset measurements for neutral referencing. Specific measurements are taken from weight-bearing x-rays to mathematically relate position and orientation of segmental local coordinate axes established from surface marker data with the underlying bony anatomy. Therefore, the MFM has the capability to detect subtle changes in foot segment orientation which may not be accessible by visual inspection such as those identified in rectus, cavus, and planus foot types.

The shape and timing of the kinematic curve is another potential difference of clinical relevance among foot types. In a comparison of high- and low-arched females during dynamic loading tasks, differences in forefoot kinematics between groups were observed in position but not range of motion (Powell et al., 2011). While our results demonstrated shifts in forefoot kinematics, the analysis of coronal forefoot velocity showed additional differences in kinematic trajectories including the timing of peak velocity. The secondary velocity analysis showed that during terminal stance, individuals in the Planus Group had a premature peak velocity toward coronal varus and early transition toward valgus.

It is postulated that this kinematic nuance can be explained by a deficient windlass mechanism. The windlass mechanism is characterized by metatarsophalangeal extension which tensions the plantar aponeurosis resulting in concomitant medial-longitudinal arch rise and forefoot supination, which is thought to be particularly important during late-stance propulsion (Hicks, 1954). During this period of the gait cycle, the forefoot supinates as the foot

transitions from a supple structure capable of accommodating support surface irregularities to a rigid lever capable of generating power to accelerate the body forward. Results from the current study demonstrated that instead of forefoot supination (varus), the Planus Group actually progressed toward valgus during terminal stance. Forefoot valgus associated with a deficient windlass mechanism can result because the planar aponeurosis was not sufficiently tensioned with metatarsophalangeal extension during terminal stance. Previous studies of foot morphology and plantar loading reported reduced medial longitudinal arch height and deficient windlass mechanism among individuals with flat feet (Mootanah et al., 2013). Future studies can explore the associations between these morphologic and plantar loading metrics and segmental foot kinematics. Additionally, differences in plantar fascia material properties can impact arch height and midfoot pronation (Cheung et al., 2004). Future work could also explore if decreased plantar fascia stiffness in the Planus Group contributed to the deficient windlass mechanism.

Our study was limited in that only 11 subjects within a limited age group for each foot type were analyzed with three trials per subject being analyzed. Therefore, the reader should be cautioned about generalizing results to other individuals throughout the lifespan. Due to the differences observed between in the primary outcome variable, we feel the sample size in this study is justified. Additionally, the MFM is limited in that it reports hindfoot motion as the combined motion of the talocrural and subtalar joints. Future work can use novel technology such as biplane fluoroscopy (Cross et al., 2017) to evaluate the individual contributions of talocrural and subtalar joint motion to overall hindfoot motion and the effect of foot type on hindfoot motion.

In conclusion, this study has identified differences in segmental foot kinematics among rectus, planus, and cavus foot types. Findings of this study included both shifts of the kinematic curves and differences in timing and magnitude of coronal forefoot velocity. Identification of these unique kinematics among foot types explains the variability among healthy adults and could result in the development of multiple reference ranges for identification of atypical motion and goal setting during clinical care, as well as identifying an appropriate comparison group for future clinical research studies. Additionally, the use of the novel statistical analysis method via LAAST allowed differences in both segmental alignment and kinematic timing among these foot types to be measured across the gait cycle.

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Declaration of Competing Interest

The authors report no conflicts of interest.

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