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Title: Foot structure is significantly associated to subtalar joint kinetics and mechanical energetics.

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Highlights

- Foot posture measurements did not predict subtalar joint (STJ) kinematics.

- Arch-height ratio was positively related to STJ moments and energy absorption.

- Step width may be a modifiable gait parameter that may reduce STJ moments.

ABSTRACT

Introduction/Aim: Foot structure has been implicated as a risk factor of numerous overuse injuries, however, the mechanism linking foot structure and the development of soft-tissue overuse injuries is not well understood. The aim of this study was to identify factors that could predict foot function during walking. **Methods:** A total of eleven variables (including measures of foot structure, anthropometry and spatiotemporal gait characteristics) were investigated for their predictive ability on identifying kinematic, kinetic and energetic components of the foot. Three-dimensional motion capture and force data were collected at preferred walking speed on an instrumented treadmill. Mechanical measures were subsequently assessed using a custom multi-segment foot model in Opensim. Factors with significant univariate associations were entered into multiple linear regression models to identify a group of factors

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independently associated with the mechanical measures. **Results:** Although no model could be created for any of the kinematic measures analysed, approximately 46% and 37% of the variance in the kinetic and energetic measures were associated with three or two factors respectively. Arch-height ratio, foot length and step width were associated with subtalar (STJ) joint moments, while greater STJ negative work was correlated to a low arch-height ratio and greater foot mobility. **Conclusion:** The models presented in this study suggest that the soft-tissue structures of a flat-arched, mobile foot are at a greater risk of injury as they have greater requirements to absorb energy and generate larger forces. However, as these associations are only moderate, other measures may also have an influence.

Keywords: Foot, Mechanical Energetics, Kinetics, Multi-segment model, Walking

INTRODUCTION

Structural foot deformities have frequently been implicated as intrinsic risk factors of lower extremity overuse injuries [1,2]. A low-arched foot has been implicated in medial tibial stress syndrome, tibialis posterior tendinopathy and other injuries involving the medial soft tissue structures of the lower extremity [2-4]. However, these perceptions are not universally accepted [5,6] primarily due to the inadequacies of commonly used measures of foot structure to account for foot function and a lack of consideration of other, perhaps more functional, measures of gait (e.g. spatiotemporal characteristics). Here we address the need to identify how foot structure relates to foot mechanics during walking, with a focus on the force and mechanical energy requirements for walking with different foot types.

It has long been postulated that structural variations in the foot alter foot kinematics, which may in turn increase the stresses and strains on soft tissue structures. However, a recent study found that measures of foot posture, or mobility, explains only a small amount of the variation in kinematic foot measures and found no association between measures indicative of increasing planus foot posture and dynamic pronatory kinematic characteristics [7]. Therefore, more meaningful measures of stresses and strains on tissues are required. When muscles, tendons and ligaments are loaded during the stance phase of gait, the amount of stress these tissues experience is most accurately inferred from measures of kinetics, e.g. joint moments [8]. The deformation of soft tissues can also be inferred from measures of mechanical work at the joints. For instance, when a joint power is negative it is absorbing energy (mechanical work) and the muscles and/or tendons and ligaments must be strained (stretched) to absorb this energy. Therefore, quantifying the relationship between foot posture and kinetics/energetics may be more fruitful in understanding overuse injuries of soft-tissues within the foot.

Incorporating structural and spatiotemporal characteristics of gait to create comprehensive models that best predict the variance in the kinematic, kinetic and mechanical energetics of the foot during walking has the potential to provide better insight into potential injury mechanisms. The aim of the present study was to develop such a model for each mechanical measure, which is shown schematically in Figure 1. The mechanical outcome measures we wished to predict were subtalar joint (STJ) pronation, STJ pronation velocity, STJ supination moments, STJ negative work and medial longitudinal arch (MLA) compression (an indicator of energy absorption in the foot) during barefoot walking due to their

relevance to injury of the lower limb [9]. Dynamic measures were calculated using a subject-specific, multi-segment foot model that enabled physiological tri-planar motion of pronation and supination rather than traditional models which report exclusive uniplanar motion. We hypothesised that measurements indicative of an increasing planus foot posture (low arch-height ratio, AHR; high foot posture index, FPI) and increased foot mobility would be associated with increased joint moments and energy absorption in the foot.

METHODS

Fifty-two subjects (29 men, 24 women) free of lower limb musculoskeletal injury for a minimum of two years gave informed written consent to participate in the study. The subjects' mean age, height, and body mass were 24.8 ± 3.9 years, 1.7 ± 0.1 m, 70.9 ± 12.7 kg, respectively. A priori sample size calculation was performed to achieve a power of 80% (effect size = 0.3, alpha level = 0.05), finding that a minimum of 46 subjects were required to include 4 factors into each model. The protocol was approved by the local university ethics committee and conducted according to the Declaration of Helsinki. The testing protocol was divided into two measurement sessions: 1) static foot measures and 2) dynamic gait measures. All static foot measures were collected by a qualified podiatrist (JNM). The static measures used in the study have high intra-rater reliability FPI [10], AHR and FMM[11].

Static foot measures. The six criteria of the FPI were measured in a relaxed standing position, summed and converted to Rasch transformed scores to be used as interval data in subsequent statistical analyses [12].

Arch-height (AH) and midfoot width (MFW) were measured at fifty percent of the total foot length using digital calipers while in a relaxed stance position on a foot posture platform [11]. AH and MFW were also measured in a non-weight bearing position using the same method while sitting. The change in AH (ΔAH) and MFW (ΔMFW) due to loading was calculated by subtracting the weight bearing and non-weight bearing measures (Figure 2). AHR was calculated as a ratio of AH during weight bearing to the participant's foot length.

Foot mobility magnitude (FMM) is a composite measure of ΔAH and ΔMFW and is calculated by taking the square root of the sum of both variables after each has been squared (Equation 1).

$$FMM = \sqrt{(\Delta AH^2 + \Delta MFW^2)} \quad \text{Equation 1}$$

Anthropometric and spatiotemporal measurements used as independent variables in the regression analyses included: body mass index (BMI = mass / body height^2), step length and step width. Step length and width were normalised to the leg length.

Mechanical measures.

Experimental Protocol. Participants were asked to walk barefoot on a force-instrumented, tandem treadmill (DBCEEWI, AMTI, USA) whilst kinematic and kinetic data were synchronously recorded. After 10-mins of familiarisation to the treadmill, participants selected their preferred walking velocity as the speed of the treadmill was incrementally increased and subsequently decreased until the subject consistently identified the same preferred velocity three times. Each subject's preferred walking velocity was used for data collection.

Motion Capture. Three-dimensional (3D) motion data were collected using an eight camera, opto-electronic motion capture system (Qualysis, Gothenburg, Sweden) at 200 Hz. Twenty-five infrared reflective markers (9 mm diameter) were placed on specific anatomical landmarks according to a multi-segment foot model developed to describe rear-, mid- and fore-foot motion [13].

Forces. Ground reaction forces and moments were collected at 4 kHz from the independent front and rear force platforms mounted beneath the spilt belts of the treadmill. Participants were required to walk such that each heel-strike landed on the front force plate and toe-off occurred from the rear force plate. This configuration allowed separate left- and right- limb force contributions and gait events to be determined.

Musculoskeletal modelling. A generic 3D musculoskeletal model available in OpenSim software [14] was modified so that each foot consisted of five segments: talus, calcaneus, mid-foot, fore-foot and toes. The segments were tracked using the same primary marker assignment as a previously published model [13], shown to have high reliability [15]. Our musculoskeletal model however, uses an inverse kinematics approach that assumes fixed degrees of freedom at each joint. The dimensions and masses of each segment were scaled using static calibration markers on the segment and body mass respectively. For further details, please see the supplementary information section.

Data analysis. Analyses were performed in Matlab (Mathworks, R2014b, Natick, MA) using custom written scripts. Raw marker positions and ground reaction forces were filtered at the same frequency, using a zero-lag, second-order, low-pass Butterworth filter with a cutoff frequency of 25 Hz to remove noise due to treadmill vibration. Inverse kinematics and inverse dynamic analyses were conducted in Opensim to output joint angles and moments, respectively.

Kinematic and kinetic data were filtered at a common frequency using a bi-directional, second-order low-pass Butterworth filter with a cut-off frequency of 10 Hz. Processed kinematic and kinetic data were time-normalised by linear interpolation from right heel strike to ipsilateral heel strike. The STJ angle is reported relative to its angle at heel strike and the mean angular velocity of the STJ was computed using 3-point difference method to calculate the 1st-order time derivative. These velocities were multiplied by STJ moments to obtain instantaneous powers of the STJ. Joint power was integrated with respect to time over discrete periods of positive and negative power, using the trapezoidal integration method, to calculate positive and negative STJ work. Moments, velocities, and work measures were normalised to non-dimensional forms to remove variability due to physical characteristics. Foot length was determined the most relevant factor to account for differences in foot dimensions using a method previously proposed [16].

The MLA angle was defined as rotation of the forefoot segment relative to the calcaneus segment about the z-axis (sagittal plane motion). An increase in MLA angle is indicative of a reduction in MLA height. Peak MLA compression was calculated as the difference between the peak MLA angle (relative to relaxed standing) and the angle at heel strike.

Statistical Analysis. Group means were computed from participant means, which were calculated over a minimum of three strides. A two-step approach was used to examine multivariate correlates of mechanical outcome. Firstly, univariate linear regression analyses were performed to investigate potential associations between each mechanical and static measure. A significance level of $P \leq 0.2$ was selected to ensure that the univariate analyses were sufficiently sensitive in identifying potential predicator variables[17]. Secondly, variables that showed significant associations on univariate analyses were entered into a series of stepwise multiple linear regression models with forward selection. Independent variables were retained in the final model if a significance of $P \leq 0.05$ was satisfied. Outliers were diagnosed and removed by checking the Cook's distance value. The strength of the identified factors was determined with standardised beta coefficients (β), while the predictive power of each final model was expressed as a percentage of explained variance (r^2). Regression models were checked for multicollinearity, heteroscedasticity and normality of residuals.

Sensitivity Analysis. A sensitivity analysis was performed to understand the effects of STJ axis variability on STJ mechanical measures. The two main sources of variability that we sought to understand were variances in the STJ inclination and the deviation angle, which were fixed in the musculoskeletal model. The sensitivity analysis covered the largest range of variation within the physiological range previously reported [18]. Intra-class correlation coefficient (ICC) was used to establish reliability and sensitivity of mechanical outcomes measures to STJ axis.

RESULTS

The mean anthropometric and static foot measurements, presented in Table 1, represent an average person, with a normal foot posture and mobility. The ranges for AHR (0.19- 0.3) and FMM (5 – 20 mm) indicate that a wide spectrum of foot types were included in the study.

Kinematic measures: The univariate analysis illustrated that peak STJ pronation had several significant ($P \leq 0.2$) predictors (step length, AHR, leg length, FPI and FMM) as did STJ pronation velocity (step length, ΔMFW, FMM, leg length and foot length). A multivariate model could not be established for

either peak STJ pronation displacement or STJ pronation velocity as no two variables were found to be significant ($P \leq 0.05$) in the final model.

Kinetic measures: Univariate linear regression revealed that AHR, step width, leg length, foot length and ΔDA were associated with peak STJ moments (normalised). All three predictors maintained significance in the multivariate model (n = 48, 3 missing data points, 1 outlier), which explained 46% of the total variance. A lower AHR and a narrower step width were associated with increased STJ supination moments. STJ moments for subjects with the highest, lowest and average AHR and with the widest and narrowest step width are shown in Figure 3A.

Energetic Measures: Univariate analysis showed FMM, ΔMFW, AHR, FPI and step length to be significantly associated with peak negative STJ work (normalized for body mass and foot length). FMM and AHR were retained in the multivariate model (n = 45, 4 missing data points, 3 outlier), to explain 38% of variance. A highly mobile foot and a low AHR resulted in greater negative work. Peak negative work for participants with the highest and lowest AHR and FMM is shown in Figure 3B.

Step length, AHR, ΔMFW and FMM had a significant univariate association with MLA compression during early stance. The multivariate analysis showed that AHR and FMM retained a significant association, with the model (n = 43, 4 missing data points, 5 outlier) explaining 37% of the total variance in peak MLA compression. A low AHR and a mobile foot were associated with greater MLA compression. Compression in the MLA for those subjects with the highest, lowest and average AHR and FMM is illustrated in Figure 3C.

Sensitivity to STJ axis: The ICC values (see Table 2) for the variations in the deviation angle ranged between (0.97 to 0.99) for all mechanical measures suggesting minimal influence of the deviation angle of the STJ axis. The ICC values for the inclination angle parameter on the peak negative work and pronation rotation were 0.93 and 0.96 respectively. STJ peak moments were most sensitive to variations in the inclination angle of the STJ axis with an ICC of 0.67. The peak STJ moments decreased as the inclination angle of the STJ axis increased.

DISCUSSION

The models developed in this study give insight into potential factors associated with STJ kinetics and energetics. The significant correlations suggest meaningful relationships between participants with low arches and mobile feet, with higher supination moments and greater absorption of energy at the STJ and MLA. However, the strength of the relationships ranged from fair to moderate, indicating the importance of other unmeasured factors. We believe these results may provide potential mechanisms linking static foot posture to soft-tissue overuse injuries

The results of this study indicate that static foot posture is associated with STJ kinetics and energetics (work), but not kinematic measures. Consistent with previous literature [7,19], no two individual measures were able to create a model to successfully predict kinematic foot mechanics. We therefore suggest that the relationship between foot posture and overuse injuries [3] is not driven by kinematic function. A recent systematic review supports this notion, reporting limited evidence associating kinematic variables with lower limb overuse injuries [20]. In contrast in-vivo joint moments and energy absorption have commonly been related to soft-tissue overuse-use injuries. For example, large abduction moments contribute to patellofemoral pain and anterior cruciate ligament injury [21]. Comparably, a narrower step width can substantially increase compression on the tibia, increasing its risk of tibial stress fractures [22] and iliotibial band strain [23]. Elastic tissue strains are essential for rapid absorption of mechanical energy [24] however they can also fail if strains are excessive [25]. As a low arch-height is linked to greater normalized STJ joint moments and STJ negative work, this may provide a potential link between foot posture and overuse injuries.

The regression models presented in this study suggest that soft-tissue structures of low-arched, mobile feet are at greater risk of overuse injury due to greater requirements for force generation and energy absorption. At the STJ, the tibialis posterior muscle is primarily responsible for generating supination moments and absorbing energy through active lengthening in the presence of STJ pronation [26]. The tendon of TP absorbs much of the energy at the STJ [26], and is therefore likely to experience larger stresses and strains in a low arched, mobile foot, potentially increasing its risk of tendinopathy. The peak STJ moment relationships also suggests that increasing step width could be a potential mechanism to

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reduce these moments and consequently the stresses and strains incurred by the tendon. However, regression models do not have the capacity to establish associations between predictors and therefore further research is required to understand if participants with low AHR walk with narrow step widths. Similarly, greater compression of the MLA indicates more energy absorption in the soft-tissue structures on the plantar aspect of the foot [27]. This may suggest that a low arched, mobile foot is at greater risk of conditions such as plantar fasciitis. While foot posture may be related to the required stress and strains in tissues across the foot, it is important that many other contributing factors such as tissue strength, training quantity and previous injury are likely to also contribute to the likelihood of injury.

Sensitivity of results to STJ axis: The large joint moments experienced in the low arched foot posture are a result of either a larger ground reaction force or larger STJ moment arm, the latter which may be influenced by the orientation of the STJ axis. It is widely accepted that the specific orientation of the STJ axis varies considerably among individuals [28] and therefore we investigated the influence of the orientation of the STJ axis on mechanical outcome measures to ensure that inaccuracies in assuming a constant STJ axis orientation would not influence our results. The sensitivity analysis suggested that increasing the inclination angle has a moderate, inverse effect on STJ moments. As the anatomy of the articulating facets of the talus and calcaneus cause the axis to move from a low pitch and medial orientation in pronation to a high pitch and forward orientation in supination [29], any individual variations in the STJ axis are likely to strengthen the results of this study. A low arch-height has been directly related to a low inclination angle of the STJ axis, which the results of this study suggest produces larger STJ moments. Therefore, if we were to calculate subject-specific STJ axes, we may have been able to explain greater variance in the joint moments.

Limitations: Although regression models are powerful tools for exploring associations among variables, it must be emphasised that it is not possible to establish a cause and effect relationship using this approach. The success of any prediction also depends on the investigator's ability to offer the critical variables to the regression process. For example, the static variables chosen in this study were restricted to the foot and it is possible that anatomical variations of more proximal bones (e.g. genu valgum) may also influence mechanical measures. Furthermore, healthy young individuals were recruited for this study, which may limit the generalisability of our findings to older or symptomatic populations. The findings may not be transferrable to overground gait, particularly during non-steady-state walking.

The musculoskeletal foot model has yet to be validated, primarily due to the difficulty in collecting a gold-standard for measuring foot dynamics without invasive measures. However, we have used the same marker set as Leardini et al [13], which has been shown to produce reliable kinematic results. Our energetic data is also comparable to that reported by Scott and Winter [30], who demonstrated that relative movements of the foot can be accurately reproduced when the STJ is modelled as a hinge joint. Furthermore, resultant joint moments do not determine the nature of soft-tissue loading, or which tissues are under stress. To determine the load transferred to the muscle-tendon structures, studies directly measuring muscle activation, force or strain are required. Greater negative joint work is however, believed to reflect larger soft tissue strain and hence we feel this measure is important in assessing links to potential factors leading to overuse injuries.

Conflict of Interest

The authors declare that they have no conflict of interest, financial or otherwise, related to the materials discussed in this manuscript.

Conflict of Interest None.

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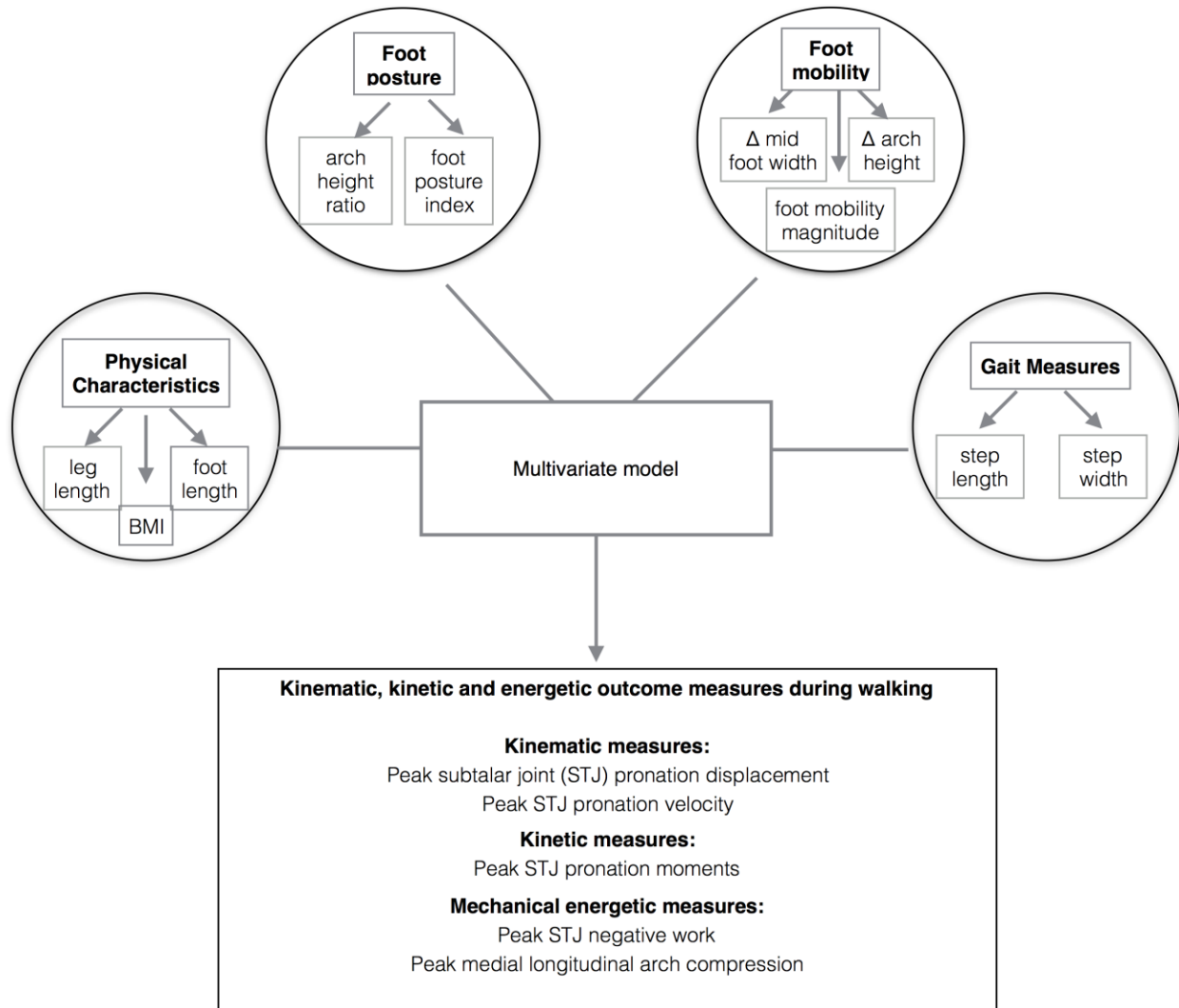
FIGURE CAPTIONS.

Figure 1: The conceptual model used to predict kinematic, kinetic and energetic outcome measures.
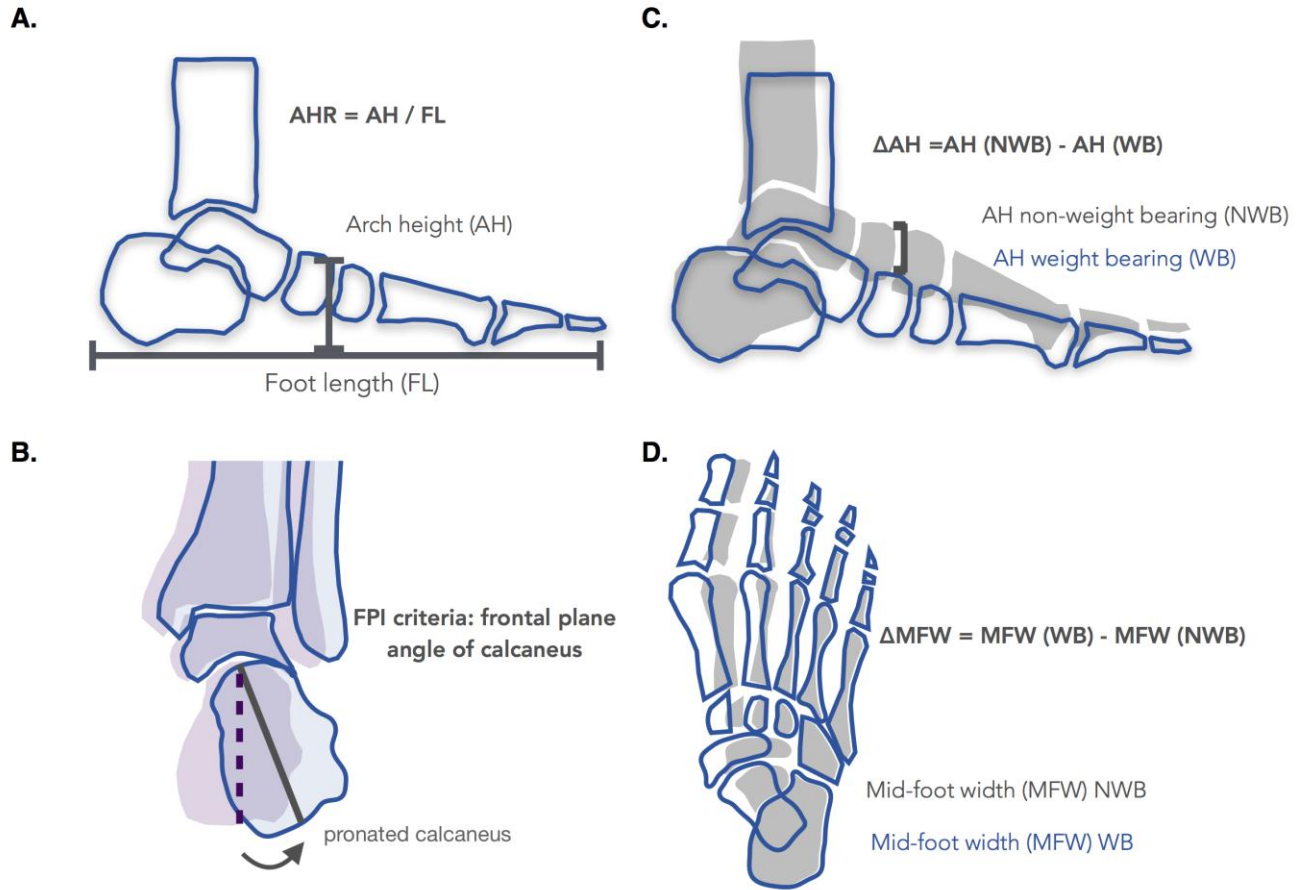
Figure 2: Image illustrating the measurements of certain foot structure measures. **A.** arch height ratio (AHR) **B.** foot posture index (FPI) **C.** Δ arch height (ΔAH) **D.** Δ midfoot width (ΔMFW). The grey scale bony anatomy represents a non-weight bearing foot.

A. Peak subtalar joint moments

Predictors	Effect	β	p	r	r²
Step width	- ive	-0.010	<0.01		
Foot length	- ive	-0.012	<0.01	**0.678**	**0.460**
Arch height ratio	- ive	-0.012	<0.01		
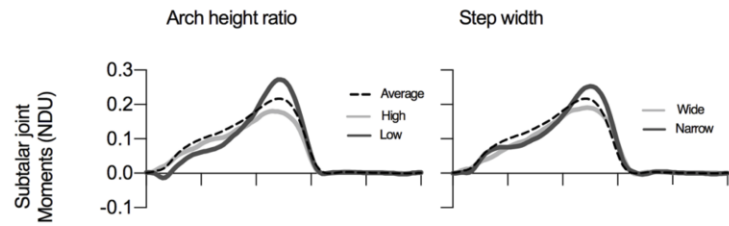
B. Peak subtalar joint negative work

Predictors	Effect	β	p	r	r²
Foot mobility magnitude	- ive	-0.001	0.01		
				0.617	**0.381**
Arch heigh ratio	+ ive	0.001	<0.01		
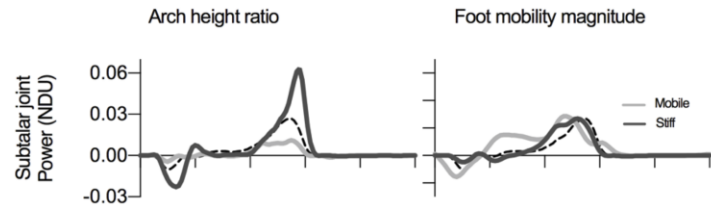
C. Peak medial longitudinal arch compression

Predictors	Effect	β	p	r	r²
Foot mobility magnitude	+ ive	0.348	<0.01		
				0.612	**0.374**
Arch heigh ratio	- ive	-0.186	0.048		

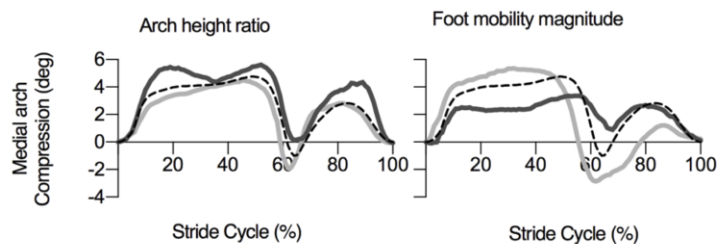**Figure 3:** Tables on the left illustrate the contribution of significant static measures (P ≤ 0.05) to **A.** peak STJ moments **B.** peak STJ negative work **C.** peak medial arch compression during early stance. β indicates the standardized beta coefficient, p represents the significance (P-value) of associated static measures; r is the correlation coefficient and r^2 is the correlation of determination of the multiple linear regression model. The standardized beta coefficient represents the standard deviations a dependent variable will change, per standard deviation increase in the associated variable. The higher relative absolute value of the beta coefficient, the stronger the effect. The plots on the right illustrate time-series data of group ensembles (dashed line) and example subjects (solid trace) during a stride cycle at preferred walking velocity. **A.** Effect of arch height ratio and step width on subtalar joint moments. **B.** Effect of arch height ratio and foot mobility magnitude on subtalar joint work (area under the negative power plot) **C.** Effect of arch height ratio and foot mobility magnitude on medial arch compression.

Table 1 Participant characteristics, values are mean (range, min:max)

Gender (n)	Male: 28; Female: 23
Age (years)	24.8 (20 - 33)
Height (m)	1.7 (1.5 - 1.95)
Body Mass (kg)	71.2 (48 - 102)
Body mass index (kg/m^2)	23.6 (16.8 - 31.4)
Foot length (cm)	25.6 (0.23 – 0.29)
Leg length (cm)	89.2 (0.78 – 1.0)
Foot posture	
Foot posture index	3.6 (-8 - 12)
Arch height ratio	0.25 (0.2 – 0.3)
Foot mobility	
Foot mobility magnitude (mm)	14.6 (5.6 – 20.2)
Δ Arch height (mm)	8.0 (2.7 - 12)

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Δ Midfoot width (mm)   11.3 (2.0 – 18.8)

Spatiotemporal measurements

Step length (cm)   66 (40 - 69)

Step width (cm)   10.4 (4.9 – 15.5)

Walking speed (m/s)   1.2 (0.9 – 1.5)

Table 2: Intra-class coefficients (ICC), p-values (p), mean and standard deviation (SD) for varying STJ axis

	ICC	p	-25⁰ Deviation		-21⁰ Deviation		-15⁰ Deviation	
			Mean	SD	Mean	SD	Mean	SD
Supination moments	0.99	<0.01	0.22	0.03	0.22	0.03	0.22	0.03
Negative work	0.97	<0.01	-0.005	0.002	-0.005	0.003	-0.004	0.002
Pronation rotation	0.99	<0.01	-6.19	2.58	-6.13	2.57	-5.86	2.62

	ICC	p	33⁰ Inclination		38⁰ Inclination		48⁰ Inclination	
			Mean	SD	Mean	SD	Mean	SD
Supination moments	0.67	<0.01	0.24	0.03	0.22	0.03	0.18	0.03
Negative work	0.93	<0.01	-0.005	0.003	-0.005	0.003	-0.004	0.002
Pronation rotation	0.96	<0.01	-5.66	2.28	-6.13	2.57	-5.64	2.48