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Effect of static foot posture on the dynamic stiffness of foot joints during walking

E. Sanchis-Sales[a,*], J.L. Sancho-Bru[b], A. Roda-Sales[b], J. Pascual-Huerta[c]

[a] Facultad de Enfermería y Podología, Universidad de Valencia, C/Jaume Roig s/n, 46010, Valencia, Spain
[b] Departamento de Ingeniería Mecánica y Construcción, Universitat Jaume I, Av. Vicent Sos Baynat, s/n, 12071, Castellón, Spain
[c] Clínica del Pie Elcano, Plaza Euskaltzaindia, 5, 48910, Barakaldo, Bilbao, Spain

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ABSTRACT

Background: The static foot posture has been related to the development of lower limb injuries.
Research question: This study aimed to investigate the dynamic stiffness of foot joints during gait in the sagittal plane to understand the role of the static foot posture in the development of injuries.
Methods: Seventy healthy adult male subjects with different static postures, assessed by the Foot Posture Index (FPI) (30 normal, 20 highly pronated and 20 highly supinated), were recruited. Kinematic and kinetic data were recorded using an optical motion capture system and a pressure platform, and dynamic stiffness at the different stages of the stance was calculated from the slopes of the linear regression on the flexion moment-angle curves. The effect of foot type on dynamic stiffness and on ranges of motion and moments was analysed using ANOVAs and post-hoc tests, and linear correlation between dynamic stiffness and FPI was also tested.
Results: Highly pronated feet showed a significantly smaller range of motion at the ankle and metatarsophalangeal joints and also a larger range of moments at the metatarsophalangeal joint than highly supinated feet. Dynamic stiffness during propulsion was significantly greater at all foot joints for highly pronated feet, with positive significant correlations with the squared FPI. Highly supinated feet showed greater dynamic stiffness than normal feet, although to a lesser extent. Highly pronated feet during normal gait experienced the greatest decrease in the dorsiflexor moments during propulsion, normal feet being the most balanced regarding work generated and absorbed.
Significance: Extreme static foot postures show greater dynamic stiffness during propulsion and greater absorbed work, which increases the risk of developing injuries. The data presented may be used when designing orthotics or prostheses, and also when planning surgery that modifies joint stiffness.

1. Introduction

Foot injuries, such as hallux valgus or plantar fasciitis (prevalences of 37% and 7%, respectively [1]), are related to abnormal motion and, more relevantly, to abnormal forces on the foot joints [2]. Analysis of the foot joint dynamics during gait can help to understand the development of these injuries [3]. Different works undertook this analysis by looking at the dynamic joint stiffness [4,5], defined as the ratio between the external moment applied to the joint and the joint angle, at a specific time, assessed while performing activities that require muscle activation, such as walking. This stiffness combines the effect of muscle forces, inertia and deformation of soft tissue, and was already applied to the ankle in the sagittal plane with different purposes [4,6,7]. High and low dynamic stiffness has been related with a higher incidence of bone injuries [8,9], and with excessive joint motion and less joint stability [10], respectively. Moreover, the analysis of dynamic stiffness is also valuable for providing mechanical properties of the foot joints to be used when designing orthotics or prostheses, and also to check the effect of surgery that may modify joint stiffness.

Recently, the authors analysed the flexion stiffness of the ankle, midtarsal and metatarsophalangeal joints during normal walking in normal healthy subjects [11], identifying different stance phases in which moment and angle changes were linearly related, i.e. with an approximately constant dynamic stiffness: early and late midstance phases and propulsion phase at the ankle and midtarsal joints, and propulsion phase at the metatarsophalangeal joint. The study of these dynamic stiffnesses in feet with different static postures may help us to understand the well-known relationship between the static foot posture and the development of lower limb injuries [12]. To date, only the effect of the static posture on foot kinematics during gait has been

* Corresponding author.
E-mail addresses: ensansa@uv.es (E. Sanchis-Sales), sancho@uji.es (J.L. Sancho-Bru), al132053@alumail.uji.es (A. Roda-Sales), info@clinicadelpieelcano.es (J. Pascual-Huerta).

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studied [13–16], but reporting contradictory data. These works found inconsistent data in peak value dependency, probably because they are affected by the reference posture [12,14], but also by the approach applied to determine relative motion. They also reported different results regarding range of motion (ROM). While some studies observed a decreased ankle ROM of pronated feet in the sagittal plane [15,16] and increased in the frontal plane [15,17,18], other works found no significant differences in any motion plane [19,20]. At the midtarsal joint, one study observed a decrease in the ROM for pronated feet in the transverse plane [13], although others found no significant differences in any motion plane [19]. And at the metatarsophalangeal joint one study observed a reduction in the ROM in the sagittal plane for pronated feet [15].

Differences in ROM results among works may also be due to differences in the static foot posture index used in each study, and to differences in the samples (age, sex, etc.). There are currently different methods available for quantifying the static foot posture [21], the foot posture index (FPI) being reported to be more reliable than other indices to estimate the foot dynamic function [22].

As the analysis of the effect of the FPI on the foot joint dynamics has been limited to their kinematics and reported contradictory data, this study aimed to analyse the effect of FPI on the foot dynamics in the principal plane of motion, the sagittal plane, during normal gait. The analysis included the comparison of the ROM, the moment ranges and the dynamic stiffnesses throughout the stance phase, and their relationship with FPI.

2. Material and methods

2.1. Experiment description

The study was carried out on 70 adult male subjects without a history of neuromuscular problems, diabetes or foot or ankle surgery, and who did not use orthotics or report pain in the lower extremity. The subjects were recruited with normal (from 0 to +5), highly pronated (HP) (higher than +10) or highly supinated (HS) (lower than −5) static FPI on both feet, as measured by Redmond et al. [22], all participants presenting very similar FPI values in both feet. All of them provided written informed consent to participate in the study, which was approved by the ethical committee of the Universitat Jaume I (Castellón, Spain).

The subjects were asked to walk barefoot along a 7-m walkway at a comfortable self-selected speed, stepping with their right foot on a pressure platform located in the middle of the walkway. Before data collection, the subjects were familiarized with the conditions by walking on the walkway several times. The subjects had to look forwards while walking, to avoid platform targeting, and they repeated the activity as many times as needed to have five valid trials, trials where they did not step on the platform with the right foot being discarded.

2.2. Data acquisition

The dynamics of the ankle and of the midtarsal and metatarsophalangeal joints of the right foot were registered using an adaptation of the model proposed by Bruening et al. [23], as presented in Sanchis-Sales et al. [11]. This model considered the midtarsal and metatarsophalangeal joints globally, rather than one particular midtarsal or metatarsophalangeal joint.

Segment position and orientation were tracked at a 100 Hz sampling rate by an eight infrared camera motion analysis system (Vicon Motion Systems Ltd., Oxford, UK). Joint angles were calculated, from the upright standing static reference posture, using a Cardan rotation sequence between distal and proximal segments: 1 - dorsiflexion/plantarflexion (DF/PF), 2 - abduction/adduction (AB/AD), and 3 - inversion/eversion (IN/EV). All kinematic data were low-pass filtered with a 4th-order Butterworth filter and cut-off frequency of 10 Hz.

Contact pressures of the right foot were recorded at a 100 Hz sampling rate with a Podoprint pressure platform (Namrol Group, Barcelona, Spain) synchronized with the infrared camera system. In each frame, pressure data were segmented by comparing the contact-cell coordinates with the anteroposterior location of the joint centres for the time when the foot was in full contact with the platform (e.g., cells with anteroposterior-coordinate between those of the midtarsal and metatarsophalangeal joint centres were assigned to the forefoot segment). The normal component of the ground reaction forces and centre of pressure (CoPs) were calculated on each foot segment (taking into account the area of the contact cells), and joint moments in the sagittal plane were then calculated from them and expressed relative to the orientation of the local coordinate system of the proximal segment. Calculated joint moments were normalized to body-weight, consistently with previous publications [4,5] and were low-pass filtered with a 4th-order Butterworth filter and cut-off frequency of 50 Hz.

2.3. Dynamic stiffness calculation

As in a previous work [11], dynamic stiffnesses were computed as the slopes of the linear regressions at those phases where the dorsiflexion moment-angle relationship was approximately linear: early midstance and propulsion for the ankle (K_{ankle}^{EMSP} and K_{ankle}^{PP}), late midstance and propulsion for the midtarsal joint (K_{MT}^{LMSP} and K_{MT}^{PP}) and propulsion for the metatarsophalangeal joint (K_{MP}^{PP}).

Phases were trimmed by 5% at both the onset and ending of each phase, and then the dynamic stiffness was calculated as the slope of the linear regression of the joint moment versus the joint angle, i.e. the tangent of the angle from the horizontal to the interpolated straight line. However, the tangent function is non-linear and presents a discontinuity at 90°, which may introduce errors when calculating mean values and when applying ANOVAs. To avoid these problems, mean calculations and ANOVAs were performed directly on the angles (θ), and results were finally transformed into dynamic stiffness data by computing the tangent of the angle data.

2.4. Statistical analysis

For each foot type, and in each foot joint, plots with the means and 95% confidence intervals (CI) were presented for the dorsiflexion angle and moment along the stance phase from all the trials and subjects. And mean joint moments were plotted versus mean joint angles, along with the linear regressions representing the dynamic stiffnesses in each of the above-mentioned phases.

For each subject, the ROMs, the ranges of joint moments and the angles representing the dynamic stiffness in each phase were averaged across the five trials recorded, as in a previous work [11]. Three sets of ANOVAs were performed to check for the effect of foot type, considering statistical significance at 0.05 level: (i) one ANOVA on the ROM with foot type as factor (normal, HP or HS) in each joint; (ii) one ANOVA per joint on the range of joint moments with foot type as factor, in each joint; and (iii) a set of ANOVAs (one for each phase at each joint) on the angles representing the dynamic stiffness as the dependent variable, with foot type as factor. Tukey post-hoc tests were performed for a deeper understanding when significant differences were detected. Finally, Pearson's correlations between dynamic stiffnesses and FPI and squared FPI were also calculated.

3. Results

The plots of the joint dorsiflexion moments versus the joint dorsiflexion angles during the stance phase (Fig. 1) showed a counter-clockwise loop at the ankle for the normal FPI feet, in agreement with previous works [6,11], and a clockwise loop for both HP and HS feet, enclosing less area in the case of normal FPI feet. At the midtarsal and metatarsophalangeal joints, all loops were clockwise, with normal FPI