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BIOMECHANICAL FUNCTION OF THE PLANTAR INTRINSIC FOOT MUSCLES

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INTRODUCTION

The human foot is a flexible multi-segment structure, capable of conforming to variations in load and surface to maintain effective force transmission between the lower limb and the ground. This functionality is achieved via an intricate interaction of movements occurring in a series of small joints (Leardini *et al.*, 2007) which allows the longitudinal arch (LA) to deform during early stance, attenuating impact forces. Later in stance the LA stiffens via the windlass mechanism (Hicks, 1954). Regulation of foot stiffness has traditionally been considered to be due to passive mechanisms of the plantar aponeurosis (Ker et. al., 1987), however recent studies have suggested that active muscular control may also contribute (Caravaggi et al., 2010).

The plantar intrinsic foot muscles possess both origins and insertions that are contained within the foot. Electromyographic studies have suggested that these muscles may provide active support for the LA during gait and postural tasks (Mann & Inman, 1964; Kelly et al., 2012). However, it is still unknown if these relatively small muscles are able to produce a significant alteration in foot biomechanics under loaded conditions that may influence the stiffness of the LA. Here we tested the hypothesis that abductor hallucis (AH), flexor digitorum brevis (FDB) and quadratus plantae (QP) are capable of generating sufficient forces to actively stiffen the LA and influence the ground reaction forces beneath the foot, with the idea that such effects would have important consequences for how energy is absorbed and dissipated during gait.

METHODS

Nine healthy males volunteered to participate in the study (mean ± standard deviation (SD) for age, height and mass were 30 ± 4yrs, 179 ± 7cm and 80 ± 6 kg, respectively). All participants were informed of the study requirements, benefits and risks before giving written informed consent. The procedures were approved by the local scientific ethics committee and performed according to the Declaration of Helsinki.

Participants were seated with their right foot placed flat on a marked area in the centre of a force plate (Kistler 9286A, Zurich, Switzerland). The shank was positioned at approximately 10 degrees of flexion (relative to vertical) and the femur perpendicular to the shank. Masses corresponding to 50% and 100% of body mass were loaded to the distal aspect of thigh using a custom built loading device in order to simulate the single and double support load demands of stance, whilst limiting the effects of postural sway on muscle activity.

To stimulate the three muscles independently, fine-wire electrodes were inserted using delivery needles into the proximal and distal portions of AH, FDB and QP muscles under ultrasound guidance. A surface ground electrode was attached to the medial malleolus of the right ankle and secured with adhesive tape. A constant current electrical stimulator (Digitimer DS7AH, Digitmer, Herfordshire, UK) was programmed to deliver trains of electrical stimulation (20 rectangular pulses, at 40Hz) across the muscle. One experimental condition consisted of three evoked muscle contractions (in one muscle) using the above stimulation parameters separated by 15s. Current intensity was determined for each muscle prior to data collection, by delivering trains of stimulation (using the above configuration) starting at 1mA and increasing incrementally by 1mA, until a mechanical response could be clearly determined as a minimum change of 10N in either the vertical ground reaction forces.

Three-dimensional motion data (Vicon MX, Oxford, UK) was also collected in order to quantify the magnitude and direction of foot motion arising from the electrically evoked muscle contractions. Retro-reflective markers were placed on the skin of the right foot and ankle according to a multi-segment foot model developed to describe rear-, mid- and fore-foot motion (Leardini et al., 2007). Motion and force plate data were processed in Visual 3D (C-Motion Inc., Germantown, USA). Assumed rigid segments were created according to a previously described multi segment foot model (Leardini et al. 2007) including the calcaneus and metatarsals.

LA length was defined as the antero-posterior distance between the sustentaculum tali marker, located on the medial calcaneus and the marker located on the medial head of the first metatarsal. LA length and height were calculated prior to simulation (load), as well as the length at peak displacement during stimulation (stimulated), for each muscle, for both 50% and 100% loading conditions. LA length and height were normalised to seated, unloaded and non-stimulated LA length and height, with positive values indicating an increase in LA length and height. For each participant, LA length and height were calculated as an average of the three stimulations.

Foot segment angles were calculated prior to (load) and during loaded stimulations in the sagittal, frontal and axial planes, for the calcaneus and metatarsal segments. Angular rotations were defined relative to the laboratory co-ordinate system and normalized to the seated, unloaded calibration position, with zero degrees being equal to the seated unloaded angle. Segment angles were calculated according to an x-y-z cardan sequence, ie extension-flexion (positive extension) as the rotation about the x-axis, inversion-eversion (positive inversion) as the rotation about the y-axis and abduction-adduction (positive adduction) as the rotation about the z-axis.

Displacement of the centre of pressure (COP) was calculated in both medio-lateral (COP_{ML}) and antero-posterior (COP_{AP}) directions, as well as the magnitude of the vertical ground reaction force (Fz). For each participant the mean of three trials was used to determine the value of each variable used for statistical analysis.

A two-way, repeated measures ANOVA was used to determine the effect of loading (50% or 100% body mass) and muscle stimulation on normalised LA length and height, segment angle, COP and Fz for each muscle stimulated. Pairwise comparisons, with Bonferroni corrections for multiple comparisons were conducted as post hoc analysis. The level of significance was set at $P<0.05$. Results are presented as group means ± standard deviation.

RESULTS AND DISCUSSION

Both increasing load and stimulation had a significant main effect on LA length and height ($P\leq0.05$, Figure 1). The MLA was significantly longer and lower when loaded with 100%, compared to 50% body mass (all $P\leq0.05$). Electrical stimulation reduced LA length and increased LA height in both 50% and 100% body mass conditions (all $P \leq 0.05$).

There was no main effect of load on calcaneal segment angle ($P\geq0.05$), however a main effect of stimulation was evident for all muscles (all $P\leq0.05$). Stimulation of AH produced extension, inversion and abduction of the calcaneus, FDB produced inversion and abduction and stimulation of QP led to abduction of calcaneus. Significant main effects of load and stimulation on metatarsal segment angle were apparent for all muscles (all $P\leq0.05$). Loading the foot with 100% body mass led to extension and abduction of the metatarsals (all $P\leq0.05$). Muscle stimulation produced flexion (AH, $P\leq0.05$) and adduction of the metatarsals (AH, FDB and QP, all $P\leq0.05$). Group means for kinematic variables are presented in table 1.

COP_{AP} and COP_{ML} location did not change between 50% and 100% loading conditions ($P>0.05$). Stimulation of AH, FDB, and QP in both 50% and 100% loading led to a significant posterior shift in COP (mean difference ± standard error for AH 6±2mm, FDB 11±3mm and QP 7±1mm, all $P\leq0.05$). Stimulation of AH also produced a significant lateral shift in COP (5±1mm, $P\leq0.05$). An increase in Fz was observed as a result of stimulation, in both 50% (AH 23.09±8.7N, FDB 21.89±13.2N and QP 20.43±11.4 N, all $P \leq 0.05$) and 100% (AH 22.73±12.1 N, FDB 20.97±21.5N and QP 20.36±21.8N, all $P \leq 0.05$) body mass loading conditions

CONCLUSIONS

We have shown that the foot is indeed a flexible structure that deforms when loaded with masses equivalent to those experienced during single and double leg support. Contraction of the intrinsic foot muscles altered foot segment biomechanics and ultimately led to an increase in the stiffness of the LA. This finding could have important implications for how energy is stored, returned and dissipated during gait.

Furthermore, contraction of these muscles produce significant alteration in the location in COP under loads consistent with single and double leg stance, thus indicating that these muscles are capable of generating sufficient force to aid in postural stabilization.

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	abductor hallucis				flexor digitorum brevis				quadratus plantae			
	50% body mass		100% body mass		50% body mass		100% body mass		50% body mass		100% body mass	
	load	stim	load	stim	load	stim	load	stim	load	stim	load	stim
LA length	15.7±1.7	15.1±1.7	15.9±1.9	15.5±1.9	15.8±1.7	15.4±1.8	15.9±1.7	15.6±1.7	15.9±1.7	15.5±1.7	16.0±1.8	15.8±1.8
LA height	4.8±0.7	5.1±6.3	4.6±0.7	4.8±0.6	4.8±0.7	5.1±0.6	4.6±0.7	4.8±0.6	4.6±0.7	4.7±0.7	4.4±0.6	4.5±0.6
Calcaneus X	2.1±3.1	2.5±3.2	2.7±2.9	2.8±2.9	1.6±5.2	2.6±5.3	1.1±5.0	1.7 (5.1)	0.5±1.6	1.2±1.3	1.7±5.5	1.8±5.6
Calcaneus Y	-1.3±4.2	-0.4±4.2	-1.3±4.9	-0.6±4.9	1.9±2.1	1.0±2.7	-1.6±2.6	-0.8 (2.9)	-2.0±2.8	-2.0±3.0	-2.8±3.4	2.5±3.1
Calcaneus Z	0.4±5.1	-1.2±5.2	0.8±5.8	-0.4±5.7	2.1±4.1	1.2±4.0	3.4±5.2	2.7 (5.0)	4.0±3.9	3.4±4.3	3.2±5.8	2.7±6.0
Metatarsal X	1.5±1.7	0.4±1.9	1.7±1.5	0.9±1.5	1.3±1.7	0.9±2.0	1.8±1.9	1.5 (1.7)	0.9±2.1	0.6±2.0	2.1±1.8	1.7±1.9
Metatarsal Y	0.7±2.3	1.1±2.5	0.3±2.4	0.6±2.2	0.5±1.8	1.7±4.0	0.4±2.1	1.2 (1.7)	0.0±2.1	0.1±2.1	0.9±2.1	1.1±2.0
Metatarsal Z	0.5±2.1	3.3±3.1	-0.9±1.6	1.2±2.0	-0.8±2.0	0.8±2.6	-1.2±2.1	0.2 (2.5)	0.1±1.5	1.0±1.4	-1.9±2.9	1.2±3.1

Table 1: Group means ± standard deviation for change in longitudinal arch (LA) length and height (cm), and calcaneal and metatarsal segment angles (degrees) due to loading of body mass and subsequent stimulation of the plantar intrinsic muscles.