American Society of Biomechanics Journal of Biomechanics Award 2015

Elastic energy within the human plantar aponeurosis contributes to arch shortening during the push-off phase of running

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ABSTRACT

During locomotion, the lower limb tendons undergo stretch and recoil, functioning like springs that recycle energy with each step. Cadaveric testing has demonstrated that the arch of the foot operates in this capacity during simple loading, yet it remains unclear whether this function exists during locomotion. In this study, one of the arch's passive elastic tissues (the plantar aponeurosis; PA) was investigated to glean insights about it and the entire arch of the foot during running. Subject specific computer models of the foot were driven using the kinematics of eight subjects running at 3.1 m/s using two initial contact patterns (rearfoot and non-rearfoot). These models were used to estimate PA strain, force, and elastic energy storage during the stance phase. To examine the release of stored energy, the foot joint moments, powers, and work created by the PA were computed. Mean elastic energy stored in the PA was 3.1 ± 1.6 J, which was comparable to *in situ* testing values. Changes to the initial contact pattern did not change elastic energy storage or late stance PA function, but did alter PA pre-tensioning and function during early stance. In both initial contact patterns conditions, the PA power was positive during late stance, which reveals that the release of the stored elastic energy assists with shortening of the arch during push-off. As the PA is just one of the arch's passive elastic tissues, the entire arch may store additional energy and impact the metabolic cost of running.

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1. Introduction

Elastic tissues (e.g., tendons and aponeuroses) in humans and other animals serve a variety of purposes, which range from preserving favorable muscle conditions during locomotion to acting as an energy buffer during impacts, and enhancing power output during explosive activities (Bobbert et al., 1986; Konow et al., 2011; Lichtwark et al., 2007). During the cyclic motions of running, elastic tissues aid in the conservation of metabolic energy as they store elastic energy (EE) during early stance and release it during push-off (Hof et al., 2002). This exchange between kinetic and elastic potential energy has been compared to that of a bouncing ball and has been suggested to increase mechanical efficiency beyond what is possible by muscle alone (Cavagna et al., 1964). Understanding the precise mechanisms by which these elastic tissues contribute to human locomotion remains an important area of research.

In situ testing has suggested that during running, the elastic tissues within the arch of the human foot can store 17 J of EE and contribute significantly to metabolic energy savings (Ker et al., 1987). To provide

context, the Achilles tendon is considered to be the primary site of EE storage and release during gait, contributing approximately 30–40 J per step (Ker et al., 1987; Rubenson et al., 2011). While Ker et al. (1987) have established the arch's energy storing potential *in situ*, there has been little work describing this EE storage and potential release during locomotion. Furthermore, questions remain as to how the release of this EE contributes to joint work within the foot during running. Given the complexity and variety of foot motion during running, further research is required to examine the influence of foot kinematics on the ability of the arch to store and return EE.

In addition to neglecting the kinematics of running, previous studies examining the role of the arch of the foot have not accounted for variations in initial foot contact, which can occur with the rearfoot, midfoot, or forefoot and can differ both with running speed and between individuals (Breine et al., 2014; Hasegawa et al., 2007). These variations in initial foot contact could, in turn, alter EE storage within the arch. Several studies have suggested this, referencing that use of a non-rearfoot contact pattern increases center of mass work without a concomitant rise in metabolic cost (Ardigo et al., 1995; Perl et al., 2012); this increased efficiency has been attributed to additional EE storage in the arch. In light of these previous studies, the influence of initial foot contact patterns on the EE in the arch of the foot warrants investigation.

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The arch contains four main elastic tissues: the long and short plantar ligaments, the spring ligament, and the plantar aponeurosis (PA). Of these four, the PA has the largest moment arm with respect to the joints of the arch, and provides the greatest support (Huang et al., 1993). Due to their small moment arms, the other three arch tissues likely make very small contributions to the joint moments. Therefore, this study estimated the EE stored and released from the PA to provide further scrutiny of the mechanism proposed by Ker et al. (1987), and used the PA as a surrogate to investigate the role of EE within the entire arch in running. The role of this EE in generating moments at the major articulations of the arch of the foot, the talonavicular joint (TN) and the metatarsophalangeal joint (MTP), was determined. Analysis of running indicates that the MTP joint absorbs energy (Stefanyshyn and Nigg, 1997), therefore as the PA crosses this joint its role in this energy absorption was examined. Two initial contact patterns (rearfoot and non-rearfoot) were investigated to encompass the various ways in which humans run. It was hypothesized that during running, EE would be stored within the PA during early stance and released during push-off. There were two secondary hypotheses: 1) use of a non-rearfoot initial contact pattern would result in greater EE storage within the PA than a rearfoot initial contact pattern, and 2) during push-off the PA would generate power at the TN joint and absorb power at the MTP joint.

2. Methods

2.1. Experimental protocol

Eight healthy runners participated as voluntary subjects (Table 1), each giving informed consent. All subjects were healthy and ran a minimum of 15 km per week, with running abilities that ranged from recreationally fit to regionally competitive. All experimental procedures were approved by the Institutional Review Board.

To obtain kinematics with which to drive a model of the foot, subjects ran barefoot at 3.1 m/s along a 20 m runway using both a rearfoot initial contact pattern (RF), and a non-rearfoot initial contact pattern (NRF). Subjects first performed three trials using their self-selected initial contact pattern, and then were asked to perform an additional three trials using the other initial contact pattern. Ground reaction forces (GRF) and moments were collected at 1500 Hz using a 90×60 cm force plate (Kistler Instrument Corp., Amherst, NY). Bony landmarks of the left foot were tracked by a 7-camera motion analysis system sampling at 150 Hz (Motion Analysis Corp., Mountain View, CA), using passive markers placed directly on the skin. Detailed foot segment kinematics were determined using a marker set adapted from Caravaggi et al. (2009), with the medial calcaneus marker moved to denote the sagittal plane location of the PA origin (Fig. 1). Kinematic data were filtered with a third order infinite impulse response Butterworth filter with a cutoff frequency of 6 Hz selected by analysis of residuals between filtered and unfiltered signals (Challis, 1999).

One footfall from each trial was identified using a vertical GRF threshold of 5 N. To identify initial foot contact pattern, the foot contact angle was measured in the global x–y plane, using the calcaneus and 2nd metatarsal markers, with positive angles indicating a RF contact and negative angles a NRF contact. Detailed

mechanical analyses were performed on the stance phase, while swing phase kinematics were used to estimate PA slack length.

2.2. Foot and plantar aponeurosis model

To estimate the function of the PA during running, a 10 degree of freedom rigid body model of the left lower leg and foot was developed in OpenSim 3.2 and driven using the experimental kinematics. The model comprised five rigid segments (shank, talus, calcaneus, midfoot, and combined toes) connected by four joints (Fig. 2). The global and segment coordinate system definitions followed ISB recommendations (Wu and Cavanagh, 1995). The ankle and subtalar joints were defined as in Arnold et al. (2010). The midfoot-calcaneus joint was a revolute joint located at the navicular tuberosity and permitted rotation about the foot's transverse axis (henceforth referred to as the talonavicular (TN) joint). The metatarsophalangeal (MTP) joint was represented as an ellipsoid joint that permitted hallux rotation about the transverse axis of the foot (Bojsen-Moller and Flagstad, 1976). The radii of this ellipsoid joint were determined from sagittal plane ultrasound imaging of the first metatarsal head of two of the subjects, collected using a 7.5 MHz linear array transducer (Hitachi-Aloka Medical America, Inc., Wallingford, CT). The boundary of the metatarsal head was identified with 20 points and modeled as an ellipse, with the resulting radii used as the radii of the model's ellipsoid MTP joint (Fig. 3). The PA was modeled as a non-linear spring that connected the PA origin (medial tubercle of the calcaneus) to the PA insertion (a point on the inferior surface of the hallux). The path of the PA was constrained to the MTP joint ellipsoid's surface (Fig. 2), thus representing the windlass mechanism.

The generic foot model was scaled to each subject using a static pose similar to Caravaggi et al. (2009), with the subject's left foot on the ground but unweighted. To estimate each subject's PA slack length (l_0), the PA was assumed to be slack when it was at its shortest length during the swing phase. This was determined by randomly selecting three swing phase trials for each subject and obtaining the 10 shortest PA lengths from each trial. The mean of these 30 values gave subject-specific l_0 values.

2.3. Analysis of PA function and EE storage

Inverse kinematics determined the joint angles by minimizing error between the experimental markers and the corresponding virtual markers on the model using a least squares formulation (Delp et al., 2007). Inverse dynamics determined the resultant joint moments, using SimBody within OpenSim 3.2 (Sherman et al., 2011). The tensile force carried by the PA was estimated using a non-linear force–deformation relationship (Fung, 1967),

$$F_{PA} = 19.65 \cdot e^{529.88 x_{PA}} \qquad 0 \leq x_{PA} \leq 0.004 \text{ m} \tag{1}$$

$$F_{PA} = 102613.74 \cdot x_{PA} - 246.81 \qquad 0.004 \text{ m} < x_{PA} \leq x_{PA,max} \tag{2}$$

where F_{PA} is PA force, and x_{PA} is PA deformation. Coefficient values were obtained by fitting the model to PA force-deformation data obtained from *in situ* loading of six cadaveric specimens (Erdemir et al., 2004). The mean body mass of the cadaver specimens was 59.6 kg, which is within one standard deviation of the mean body mass of the subjects. Joint moments created by the PA were computed as the product of the PA force and PA moment arm throughout stance. The PA generated joint moments were expressed as a fraction of the peak resultant joint moment. EE stored in the PA was calculated as,

$$EE = \int_0^{l_{PA}} F_{PA} \, dl \tag{3}$$

where l_{PA} is the PA length at peak PA strain.

Table 1
Subject characteristics.

Subject	Sex	Age (yrs)	Mass (kg)	Height (m)	Foot length[a] (cm)	Foot width[b] (cm)	Navicular height (cm)	Preferred contact[c]
1	F	22	65.0	1.81	24.1	9.8	6.1	NRF
2	F	21	57.5	1.61	21.8	7.7	4.9	RF
3	F	48	58.5	1.58	23.4	8.3	4.5	NRF
4	M	42	80.5	1.85	25.8	9.8	4.1	NRF
5	M	24	66.5	1.85	24.6	9.4	5.0	NRF
6	F	26	59.0	1.63	23.0	9.2	5.7	RF
7	M	29	80.1	1.90	25.6	8.4	6.3	NRF
8	F	24	53.9	1.62	21.3	7.3	5.2	RF
Mean	–	29.5	65.1	1.73	23.7	8.7	5.2	–
SD	–	10	10.2	0.13	1.6	1.0	0.8	–

[a] Foot length was measured as the distance from the heel marker to the marker on the longest toe.
[b] Foot width was measured as the distance between the 1st and 5th metatarsal head markers.
[c] Preferred contact was determined through practice trials. RF=rearfoot initial contact pattern and NRF=non-rearfoot initial contact pattern.