Author(s): Peter, Annamaria; Hegyi, Andras; Stenroth, Lauri; Finni Juutinen, Taija; Cronin, Neil

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Title: EMG and force production of the flexor hallucis longus muscle in isometric plantarflexion and the push-off phase of walking

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Authors: Annamária Péter, András Hegyi, Lauri Stenroth, Taija Finni, Neil J Cronin.

Annamária Péter and András Hegyi contributed equally to this paper.

Affiliation: Neuromuscular Research Centre, Department of Biology of Physical Activity, University of Jyväskylä, Finland.

Corresponding author: Annamária Péter,

Address: Viveca, Rautpohjankatu 8, 40700 Jyväskylä, Finland (e-mail: annamaria.a.peter@jyu.fi).

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Abstract

Flexor hallucis longus (FHL) is the largest toe flexor muscle of the foot. Previous studies suggest that the multiarticular FHL may contribute to forefoot force transmission via energy conservation. The aim of this study was to examine FHL function at different levels of isometric plantarflexion torque and in the push-off phase at different speeds of walking. FHL and calf muscle activity were measured by surface EMG and plantar pressure was recorded by pressure insoles. FHL activity was compared to the activity of the calf muscles. Force and impulse values were calculated under the big toe, and were compared to the entire pressed area of the insole to determine the relative contribution of FHL to forefoot force transmission. FHL activity increased with increasing plantarflexion torque level (F = 2.8, P = 0.024) and with increasing walking speed (F = 11.608, P < 0.001). No differences were observed in the relative contribution of the force under the big toe to the entire sole between different plantarflexion torque levels (F = 0.836 P = 0.529). On the contrary, in the push-off phase of walking peak force under the big toe increased at a higher rate than force under the other areas of the plantar surface (F = 3.801, P = 0.018). Our results indicate that the relative functional importance of FHL increases with increasing walking speed. However, substantial differences were found between isometric plantarflexion and walking concerning FHL activity relative to that of the calf muscles, highlighting the task-dependent behaviour of FHL.

1. Introduction

The largest joint of the foot is the first metatarso-phalangeal joint (MPJ) (Leardini et al., 1999), which is crossed by the flexor hallucis longus muscle (FHL). The FHL originates from the distal 2/3 of the fibula and the membrana interossea, and inserts onto the distal phalanx of the big toe while also spanning the ankle (Gray, 1980). Several functions have been attributed to FHL, including supination of the foot (Ferris et al., 1995), maintenance of the medial longitudinal arch of the foot (Thordarson et al., 1995), production of inversion torque on the rearfoot (Hintermann et al., 1994) and plantarflexion torque at the ankle (Klein et al., 1996). FHL may also play a role in accelerating the centre of mass in the push-off phase of locomotion, as isolated strength training of the toe flexors has been shown to increase horizontal jump performance (Goldmann et al., 2013).

Modelling data have shown that triceps surae forces are transmitted to the forefoot, even though the origin of the Achilles tendon is on the calcaneus (Chen et al., 2012). The plantar aponeurosis (Erdemir et al., 2004; Ker et al., 1987) and intrinsic muscles of the foot (Kelly et al., 2014) have both been found to play a role in forefoot force transmission from the calcaneus. In the push-off phase of cyclic movements, energy conservation via bi- and multiarticular muscles exists as they have the capacity to transmit energy from proximal to distal joints and vice versa (Prilutsky and Zatsiorsky, 1994) while functioning relatively isometrically. This so-called proximo-distal energy conservation has been observed for the rectus femoris (Bobbert and van Ingen Schenau, 1988) and gastrocnemius muscles (Fukunaga et al., 2001) during locomotion. The low fibre length to muscle length ratio of FHL (Ward et al., 2009) suggests that FHL may work near-isometrically during walking. In fact, recent cadaver studies support this idea, implying that proximo-distal energy conservation may be a main role of this muscle (Hofmann et al., 2013; Kirane et al., 2008).

It has been estimated that during walking the forces within the foot are largest in the first ray, with 29% body weight under the first MPJ and 24% under the big toe (Jacob, 2001). FHL may make a major contribution to these large forces, since it has the largest physiological cross-sectional area of the foot muscles (Friederich and Brand, 1990). Jacob (2001) estimated that the force along the FHL tendon is around 52% of body weight at the second peak of the vertical ground reaction force in the stance phase of walking. FHL muscle activity is also maximal at the terminal stance phase of gait (Perry, 1992).

Numerous studies have shown the multifunctionality of FHL, but all of these studies used cadaver or indirect methods to estimate muscle function. Direct in vivo methods have not been used except for the study of Perry (1992), which did not examine the effects of activity on movement outcomes. The aim of this study was to examine FHL activity and the resulting force under the big toe during sustained isometric plantarflexions at different contraction levels, and in the push-off phase of walking at different speeds. This allowed us to determine differences in the use of this muscle between an isolated and a functional task, as well as changes with increasing load.

2. Methods

2.1. Participants

Eleven male subjects (age: 24.7 ± 3.7 yrs; height: 180.6 ± 6.6 cm; body mass: 79.2 ± 9.1 kg) with no history of neuromuscular disorder or injury volunteered for this study. Subjects with self-reported flat foot were excluded (Angin et al., 2014). The right (dominant) leg of each subject was measured. The experimental procedures were approved by the ethics committee of the University of Jyväskylä and all subjects gave written informed consent. Testing was conducted according to the Declaration of Helsinki.

2.2. Experimental protocol

After shaving and abrading the skin lightly then cleansing with alcohol, bipolar electromyography (EMG) electrodes were placed over tibialis anterior (TA), gastrocnemius lateralis (GL), gastrocnemius medialis (GM), soleus (SOL) and flexor hallucis longus (FHL) muscles. After checking that electrode impedance was below 5kΩ the signal quality was confirmed with a few plantarflexion and toe flexion tasks. Subjects then performed overground walking trials over two 10m instrumented force platforms parallel to each other. Thereafter, submaximal isometric plantarflexions were executed in a custom-made ankle dynamometer (University of Jyväskylä, Finland). Finally, for normalization of EMG signals, maximal voluntary isometric contractions (MVIC) were performed. During all walking and plantarflexion trials, foot pressures (Novel, Munich, Germany) were recorded using pressure insoles.

Walking trials. After warm-up and familiarisation trials, subjects were asked to walk at a self-selected speed over the force plates five times. The slowest and fastest trials were discarded and the remaining values were averaged to signify the preferred speed of walking (PW). Subjects then walked at maximal walking speed (MW) as well as at slower (SW) and faster (FW) speeds than PW: target durations over the force platforms were PW + 30% and PW – 30%, respectively. PW trials were always performed first, but subsequent trials were performed in a random order. Three successful trials (within ± 5% of target time) were recorded for the MW, SW and FW conditions.

Submaximal and maximal plantarflexions. After some familiarisation trials, subjects were asked to reach different isometric plantarflexion torque levels and maintain each of them for 2 seconds. Torque-time curves served as visual feedback for the subjects. Torque levels were set at 20, 40, 60, 80, 100 and 120 Nm and were performed in a random order. The reason for choosing absolute values instead of relative to each individual's maximum was that low torques within this range were expected during walking. Submaximal trials served as an initial warm up for the maximal contractions. In addition, 3 submaximal contractions lasting 2-3 s at around 70% MVIC were performed before maximal trials. For TA, triceps surae and FHL muscles, dorsiflexion, plantarflexion and big toe flexion superimposed on plantarflexion were performed, respectively. Strong verbal encouragement was given to achieve maximal effort. MVICs were performed 3 times with 1.5 minute rest between contractions. In cases where the greatest torque was recorded in the third trial, additional contraction(s) were performed until maximal torque no longer increased.

2.3.Instrumentation

2.3.1. EMG activity

EMG activity was recorded using a telemetric system (Noraxon Inc. Scottsdale, AZ, USA). Silver-silver chloride Ambu *BlueSensor* N bipolar surface electrodes (Ambu A/S, Ballerup, Denmark) were used to measure EMG activity according to the SENIAM recommendations for TA, GL and GM (Hermens et al., 1999). For FHL, proper placement of electrodes was determined using ultrasonography: the electrodes were placed between the insertion of SOL and the calcaneus on the medial side, posterior to the medial malleolus where FHL was identified. The inter-electrode distance for FHL was reduced to 16 mm to avoid cross-talk from other muscles (Bojsen-Møller et al., 2010; Masood et al., 2014). For SOL the electrodes were placed on the lateral side of the muscle to avoid cross-talk between SOL and FHL, but otherwise the SENIAM recommendations were followed. Inter-electrode distance for TA and triceps surae muscles was 22 mm. EMG signals were sampled at a frequency of 1500 Hz and transmitted wirelessly to an A/D converter (Cambridge Electronic Design, Cambridge, UK) connected to a personal computer. Digital signals were recorded and visualized online along with the ground reaction force (GRF) signals in Spike 2.0 software (Cambridge Electronic Design, Cambridge, UK).

2.3.2. Force and torque

Walking trials were performed on two 10 m x 0.5 m force plates (sampling frequency was 1000 Hz) parallel to each other, allowing 3 dimensional GRFs to be recorded for each step and leg separately. GRF curves served to monitor the physical nature of steps (Figure 1). To avoid recording the acceleration and deceleration phases, participants were asked to start approximately 3 m prior to the force plates and continue for 2-3 meters after them, allowing several steps to be recorded at a steady speed. Speeds were determined using photocells at the start and end of the force plates. During isometric tasks torque was measured using a custom-made ankle dynamometer. The hip, knee and ankle were fixed at 120, 0 and 90 degrees, respectively. The right foot was firmly strapped to the pedal of the dynamometer, and the thigh was strapped to the seat to avoid postural changes between trials and subjects.

2.3.3. Plantar pressure

Plantar pressure was recorded using the Pedar in-shoe dynamic pressure measuring system (Novel, Munich, Germany) during walking and submaximal isometric trials. An insole including 99 pressure sensors was placed into the subject's right shoe. Data were collected at 100 Hz via Bluetooth using Pedar's own software. At the start of the recordings a trigger signal was sent to the Spike software to synchronize the data. Before every task the big toe was pressed down by an assistant while the foot was unloaded, to help map the pressure area covered by the big toe.

2.4. Analysis

Trials including steps that deviated by more than ±5% of the average duration of all gait cycles within a trial were excluded from the analysis for each subject.

2.4.1. EMG activity

TA activity served as a visual feedback from the antagonist activity during the step cycle. Activity of the other muscles was analysed in the following way. EMG signals were band-pass filtered between 20 and 450 Hz by a butterworth filter in MATLAB (MathWorks Inc, Natick, MA, US.). Root mean square (RMS) EMG activity of each muscle was calculated in the push-off phase of walking for each step based on the horizontal GRF, and values within each trial were averaged. For submaximal isometric contractions, RMS was calculated from a 1-s stable plateau to characterize the EMG activity at different torque levels. The MVIC where the peak torque was highest was used to normalise the EMG. After filtering, RMS was calculated from a stable 1-s plateau around the peak MVIC torque and considered to be maximal voluntary EMG activity. FHL activity was also defined relative to the individual triceps surae muscles (FHL/LG, FHL/MG and FHL/SOL) and to the average activity of the triceps surae (FHL/TS) to establish FHL activity relative to the primary plantarflexors.

2.4.2. Plantar pressure analyses

Before each trial, an investigator manually pressed down on the big toe. In the subsequent analysis, this was used to define the area of the big toe. The sensors at the edge of this area were excluded from the analysis to avoid interference from other toes or foot regions. Pressing the toe down before each trial also allowed us to ensure that the insole did not move inside the shoe between tasks.

Forces were calculated within the push-off phase of walking by dividing the pressure by the area of the pressed sensors. Peak forces were defined under the big toe and the whole foot for each step and were averaged for every trial. Impulse was calculated as the integral of the force-time curve for the sensors under the big toe and under the foot. All force and impulse values in walking were normalized to body mass (BM). Average force during isometric plantarflexion was calculated over the same time interval as the calculation of RMS EMG. The contributions of the big toe to peak or average force and impulse were calculated by dividing the big toe values by the respective values for the whole plantar surface (hereafter these values are referred to as *relative big toe force* and *relative big toe impulse*).

2.4.3. Statistical analysis

All statistical analyses were performed using IBM SPSS software (IBM New York, US). Normality of distribution was tested by Shapiro Wilk's W test, while homogeneity of variance was determined with Levene's test. Spearman's rank was used to perform correlations between FHL activity and force under the big toe in isometric and walking trials. Differences between EMG, force or impulse values at submaximal plantarflexion levels and different speeds of walking were defined by one-way ANOVA and Tukey HSD post-hoc test if normal distribution and homogeneity of variances were confirmed. One of the subjects did not perform MW trials so differences between MW and other walking tasks were defined using Gabriel's procedure. In all other cases non-parametric Kruskal-Wallis ANOVA and Games-Howell post-hoc tests were used to signify differences. Significance level was set at $P < 0.05$. Values in the text are expressed as mean ± standard deviation.

3. Results

Maximal isometric plantarflexion torque was 362.39 ± 78.35 Nm. In isometric tasks with varying torque level, average force under the big toe generally increased with torque (F = 5.41

$P < 0.001$), but there were no significant changes in relative big toe force ($F = 0.836$ $P = 0.529$) (Figure 2). For walking trials, absolute peak force under the big toe increased significantly with increasing speed ($F = 4.923$, $P = 0.005$), and the increase was greater than in the other regions of the foot ($F = 3.801$, $P = 0.018$). Impulse under the big toe did not increase with speed ($F = 0.078$, $P = 0.972$), however relative big toe impulse did ($F = 5.067$, $P = 0.005$). Force under the big toe increased linearly as a function of increasing FHL activity, both in isometric tasks ($r = 0.653$, $P < 0.001$) and during walking ($r = 0.439$, $P = 0.003$).

Walking velocity, number of analysed steps for each walking trial and duration of the push-off phase are described in Table 1. Table 2 presents FHL EMG activity normalized to maximal voluntary activity, and compared to the activity of the superficial plantarflexors in different tasks. FHL activity increased with increasing plantarflexion level and walking speed ($F = 2.8$, $P = 0.024$ and $F = 11.608$, $P < 0.001$, respectively). In sustained isometric plantarflexions the FHL/SOL activity ratio was significantly higher at 20 Nm compared to all other torque levels ($F = 6.298$, $P < 0.001$). FHL/MG, FHL/LG and FHL/TS did not change with increasing torque. For walking trials, none of the activity ratios altered significantly with increasing speed.

4. Discussion

Our results show that increased isometric plantarflexion torque and increased walking speed were associated with higher FHL activity and concomitant higher forces under the big toe. It is well known that an increase in walking speed results in a decrease in stance phase duration, as well as the duration of all sub-phases (Liu et al. 2014) such as push-off (Table 1). On the contrary, force under the big toe increased at a higher rate than force in other regions of the plantar surface as walking speed increased. Thus, impulse was maintained across walking speeds by an increasing force level under the big toe due largely to FHL activity. With increasing walking speed, big toe flexion impulse relative to total foot impulse increased, highlighting the importance of FHL at higher speeds. On the other hand, in isometric tasks, as plantarflexion torque increased, significant changes in relative force under the big toe were not observed, indicating that the role of FHL did not change across torque levels.

Previously, based on surface EMG, Masood et al. (2014) found that the contribution of FHL to plantarflexion was about 31 ± 13% during intermittent isometric exercise at 30% of MVIC. In the current study, FHL activity was 16 ± 10% of the cumulative EMG activity of all four plantarflexors at a torque level of 100 Nm (28.72 ± 5.94% MVIC). Peak force under the big toe at preferred walking speed (11 ± 4% BM) was also lower than 24 ± 8% BM estimated by Jacob (2001). However, it should be emphasized that the method we used to map the big toe presumably underestimated the size of the target area, since we attempted to eliminate interference from other parts of the plantar surface.

During isometric plantarflexion FHL functions as an important deep plantarflexor (Klein et al., 1996). In this isolated contraction mode, forces are not generated at proximal joints, and thus energy conservation from proximal joints is minimal compared to the situation during locomotion. In this study, EMG activity increased in parallel in the gastrocnemius and FHL muscles with increasing speed. Gastrocnemius has been reported to transmit forces towards the forefoot (Chen et al., 2012; Fukunaga et al., 2001). In the second half of the stance phase elastic energy reuse in the longitudinal arch buttresses the generation of positive work (Ker et al., 1987). The main contributor to this so-called windlass mechanism (Hicks, 1954) is the plantar aponeurosis (Erdemir et al., 2004), with an increased tension level at higher walking speeds due to the more dorsiflexed toes in the push-off phase (Caravaggi et al., 2010). In vivo motion analyses have shown that at higher speeds, the foot becomes stiffer and so foot length decreases (Stolwijk et al., 2014). The decrease in foot length can be achieved by increased tension of the tissues on the plantar surface. In addition to the passive elements, further increases in foot stiffness are achieved by increasing the activity of the foot flexor muscles. Kelly et al. (2014) found increased EMG activity of the abductor hallucis, flexor digitorum brevis and quadratus plantae with increasing external load applied to the longitudinal arch, suggesting that intrinsic foot muscles serve to decrease stress on the passive elements and increase foot stiffness. In our study, we found that activity in the largest extrinsic foot flexor also increased with increasing walking velocity. Since this muscle contributes to maintenance of the medial longitudinal arch (Thordarson et al., 1995), it may also contribute to foot stiffness.

Stolwijk et al. (2014) found that the foot begins to shorten upon heel elevation, achieves its peak rate of shortening after contralateral heel strike, and then rapidly lengthens just before toe off. Perry (1992) also found that FHL activity is highest in the terminal stance phase but rapidly decreases just before toe off. Thus, changes in foot length and FHL activity seem to be linked. We found that FHL activity and force under the big toe correlated significantly. If FHL contributes substantially to foot stiffness and propulsion of the body, the force under the big toe might be expected to increase at a higher rate than the force under the other parts of the foot as speed increases. Indeed, this is what we observed for peak force and impulse under the big toe. Higher relative force under the big toe at faster velocities can be achieved by a stiffer first ray compared to the other parts of the foot. This may enhance force transmission via FHL compared to other structures. The optimal joint angle combination for FHL is 0°-10° ankle dorsiflexion and 25°-45° MPJ dorsiflexion (Goldmann and Bruggemann, 2012).

Previous studies suggest that the optimal length of FHL can be maintained across velocities in the late stance phase of walking (Caravaggi et al., 2010) and that this muscle works near-isometrically (Hofmann et al., 2013; Kirane et al., 2008). This in turn suggests efficient energy conservation via FHL, as reflected by the increasing relative contribution of FHL force to total force under the foot regions in contact with the ground as walking speed increased in this study.

However, the potential for force transmission from the calf muscles to the ground via the FHL seems to be quite low. Before reaching the calcaneus, FHL descends via a fibro-osseous tunnel (Gray, 1980) that limits its physical connections to other structures before reaching the distal phalanx of the big toe. Intramuscular force transmission between SOL and FHL has also been found to be minimal or even absent (Bojsen-Møller et al., 2010). Thus, force transmission via the FHL seems to be distinct from force transmission from the calf muscles. In support of this, in our study the force under the big toe relative to the plantar surface was significantly higher at faster walking velocities, which suggests an increasing relative importance of energy conservation via FHL.

In the isometric conditions, the FHL/SOL ratio at 20 Nm was significantly higher than at higher isometric torque levels. However, absolute FHL activity increased gradually with increasing torque level, suggesting that soleus may not contribute substantially to plantarflexion torque at low torque levels. It should be noted that the FHL/SOL ratio was about two times greater in isometric tasks than in walking, whereas FHL/MG and FHL/LG were twofold higher in walking than in sustained isometric plantarflexions. Otherwise, FHL activity relative to the individual triceps surae muscles did not change generally across torque levels nor with increasing walking speed. These results highlight the task-dependent function of the FHL muscle.

LIMITATIONS

5. Conclusion

Our results have shown for the first time how FHL force and EMG activity vary over different isometric plantarflexion levels and different speeds of locomotion. We found that FHL activity generally increases in parallel with the force under the big toe during isometric plantarflexion and walking. Our results suggest that FHL plays an important role in energy conservation from the shank to the ground, and that this contribution increases with increasing walking speed.

Conflict of interest statement

None of the authors have any conflict of interest to declare.

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Figure Captions

Figure 1. Raw data from one walking trial and one subject. Panel A-C show ground reaction forces (GRF) measured from the force platform. In addition, the black line in A represents the force calculated for the area under the big toe. Panels D-H show EMG activity from the FHL, SOL, MG, LG and TA muscles. Shaded areas represent the push-off phase of steps which were analysed.

Figure 2. Force and impulse in different tasks. A, C and E are calculated for the area under the big toe. Relative values (B, D, F) present force and impulse under the big toe relative to the same values for the entire plantar surface. For walking peak force and impulse are normalised to body mass (BM) and calculated in the push-off phase (C, E). Statistical differences: * $p<0.05$, ** $p<0.01$.

Table 1. Descriptive statistics of walking trials included in the analysis. SW = slow walking, PW = preferred speed walking, FW = fast walking, MW = maximal speed walking. For SW and FW target times were PW + 30% and PW - 30%, respectively (± 5%).

Table 2. FHL EMG activity alone and compared to EMG in the other calf muscles. For plantarflexion trials values refer to a 1-sec plateau; for walking trials values were calculated as the mean in the push-off phase. Ratios were calculated after normalising to MVC.