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5

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17

18 **Abstract**

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

36

37 **1. Introduction**

38

39 The largest joint of the foot is the first metatarso-phalangeal joint (MPJ) (Leardini et al.,
40 1999), which is crossed by the flexor hallucis longus muscle (FHL). The FHL originates from
41 the distal 2/3 of the fibula and the membrana interossea, and inserts onto the distal phalanx of
42 the big toe while also spanning the ankle (Gray, 1980). Several functions have been attributed
43 to FHL, including supination of the foot (Ferris et al., 1995), maintenance of the medial
44 longitudinal arch of the foot (Thordarson et al., 1995), production of inversion torque on the
45 rearfoot (Hintermann et al., 1994) and plantarflexion torque at the ankle (Klein et al., 1996).
46 FHL may also play a role in accelerating the centre of mass in the push-off phase of
47 locomotion, as isolated strength training of the toe flexors has been shown to increase
48 horizontal jump performance (Goldmann et al., 2013).

49

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

61 cadaver studies support this idea, implying that proximo-distal energy conservation may be a
62 main role of this muscle (Hofmann et al., 2013; Kirane et al., 2008).

63

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

71

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

80

81 **2. Methods**

82

83 *2.1. Participants*

84 Eleven male subjects (age: 24.7 ± 3.7 yrs; height: 180.6 ± 6.6 cm; body mass: 79.2 ± 9.1 kg)
85 with no history of neuromuscular disorder or injury volunteered for this study. Subjects with

86 self-reported flat foot were excluded (Angin et al., 2014). The right (dominant) leg of each
87 subject was measured. The experimental procedures were approved by the ethics committee
88 of the University of Jyväskylä and all subjects gave written informed consent. Testing was
89 conducted according to the Declaration of Helsinki.

90 *2.2. Experimental protocol*

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

102 *Walking trials.* After warm-up and familiarisation trials, subjects were asked to walk at a self-
103 selected speed over the force plates five times. The slowest and fastest trials were discarded
104 and the remaining values were averaged to signify the preferred speed of walking (PW).
105 Subjects then walked at maximal walking speed (MW) as well as at slower (SW) and faster
106 (FW) speeds than PW: target durations over the force platforms were $PW + 30\%$ and $PW -$
107 30% , respectively. PW trials were always performed first, but subsequent trials were
108 performed in a random order. Three successful trials (within $\pm 5\%$ of target time) were
109 recorded for the MW, SW and FW conditions.

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

123 *2.3. Instrumentation*

124 *2.3.1. EMG activity*

125 EMG activity was recorded using a telemetric system (Noraxon Inc. Scottsdale, AZ, USA).
126 Silver-silver chloride Ambu *BlueSensor* N bipolar surface electrodes (Ambu A/S, Ballerup,
127 Denmark) were used to measure EMG activity according to the SENIAM recommendations
128 for TA, GL and GM (Hermens et al., 1999). For FHL, proper placement of electrodes was
129 determined using ultrasonography: the electrodes were placed between the insertion of SOL
130 and the calcaneus on the medial side, posterior to the medial malleolus where FHL was
131 identified. The inter-electrode distance for FHL was reduced to 16 mm to avoid cross-talk
132 from other muscles (Bojsen-Møller et al., 2010; Masood et al., 2014). For SOL the electrodes
133 were placed on the lateral side of the muscle to avoid cross-talk between SOL and FHL, but
134 otherwise the SENIAM recommendations were followed. Inter-electrode distance for TA and

135 triceps surae muscles was 22 mm. EMG signals were sampled at a frequency of 1500 Hz and
136 transmitted wirelessly to an A/D converter (Cambridge Electronic Design, Cambridge, UK)
137 connected to a personal computer. Digital signals were recorded and visualized online along
138 with the ground reaction force (GRF) signals in Spike 2.0 software (Cambridge Electronic
139 Design, Cambridge, UK).

140 2.3.2. *Force and torque*

141 Walking trials were performed on two 10 m x 0.5 m force plates (sampling frequency was
142 1000 Hz) parallel to each other, allowing 3 dimensional GRFs to be recorded for each step
143 and leg separately. GRF curves served to monitor the physical nature of steps (Figure 1). To
144 avoid recording the acceleration and deceleration phases, participants were asked to start
145 approximately 3 m prior to the force plates and continue for 2-3 meters after them, allowing
146 several steps to be recorded at a steady speed. Speeds were determined using photocells at the
147 start and end of the force plates. During isometric tasks torque was measured using a custom-
148 made ankle dynamometer. The hip, knee and ankle were fixed at 120, 0 and 90 degrees,
149 respectively. The right foot was firmly strapped to the pedal of the dynamometer, and the
150 thigh was strapped to the seat to avoid postural changes between trials and subjects.

151 2.3.3. *Plantar pressure*

152 Plantar pressure was recorded using the Pedar in-shoe dynamic pressure measuring system
153 (Novel, Munich, Germany) during walking and submaximal isometric trials. An insole
154 including 99 pressure sensors was placed into the subject's right shoe. Data were collected at
155 100 Hz via Bluetooth using Pedar's own software. At the start of the recordings a trigger
156 signal was sent to the Spike software to synchronize the data. Before every task the big toe
157 was pressed down by an assistant while the foot was unloaded, to help map the pressure area
158 covered by the big toe.

159 *2.4. Analysis*

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

162 *2.4.1. EMG activity*

163 TA activity served as a visual feedback from the antagonist activity during the step cycle.

164 Activity of the other muscles was analysed in the following way. EMG signals were band-
165 pass filtered between 20 and 450 Hz by a butterworth filter in MATLAB (MathWorks Inc,
166 Natick, MA, US.). Root mean square (RMS) EMG activity of each muscle was calculated in
167 the push-off phase of walking for each step based on the horizontal GRF, and values within
168 each trial were averaged. For submaximal isometric contractions, RMS was calculated from a
169 1-s stable plateau to characterize the EMG activity at different torque levels. The MVIC
170 where the peak torque was highest was used to normalise the EMG. After filtering, RMS was
171 calculated from a stable 1-s plateau around the peak MVIC torque and considered to be
172 maximal voluntary EMG activity. FHL activity was also defined relative to the individual
173 triceps surae muscles (FHL/LG, FHL/MG and FHL/SOL) and to the average activity of the
174 triceps surae (FHL/TS) to establish FHL activity relative to the primary plantarflexors.

175 *2.4.2. Plantar pressure analyses*

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

181 Forces were calculated within the push-off phase of walking by dividing the pressure by the
182 area of the pressed sensors. Peak forces were defined under the big toe and the whole foot for

183 each step and were averaged for every trial. Impulse was calculated as the integral of the
184 force-time curve for the sensors under the big toe and under the foot. All force and impulse
185 values in walking were normalized to body mass (BM). Average force during isometric
186 plantarflexion was calculated over the same time interval as the calculation of RMS EMG.
187 The contributions of the big toe to peak or average force and impulse were calculated by
188 dividing the big toe values by the respective values for the whole plantar surface (hereafter
189 these values are referred to as *relative big toe force* and *relative big toe impulse*).

190 2.4.3. Statistical analysis

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

202

203 **3. Results**

204

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

207 $P < 0.001$), but there were no significant changes in relative big toe force ($F = 0.836$ $P =$
208 0.529) (Figure 2). For walking trials, absolute peak force under the big toe increased
209 significantly with increasing speed ($F = 4.923$, $P = 0.005$), and the increase was greater than
210 in the other regions of the foot ($F = 3.801$, $P = 0.018$). Impulse under the big toe did not
211 increase with speed ($F = 0.078$, $P = 0.972$), however relative big toe impulse did ($F = 5.067$,
212 $P = 0.005$). Force under the big toe increased linearly as a function of increasing FHL
213 activity, both in isometric tasks ($r = 0.653$, $P < 0.001$) and during walking ($r = 0.439$, $P =$
214 0.003).

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

224

225 **4. Discussion**

226

227 Our results show that increased isometric plantarflexion torque and increased walking speed
228 were associated with higher FHL activity and concomitant higher forces under the big toe. It
229 is well known that an increase in walking speed results in a decrease in stance phase duration,
230 as well as the duration of all sub-phases (Liu et al. 2014) such as push-off (Table 1). On the
231 contrary, force under the big toe increased at a higher rate than force in other regions of the

232 plantar surface as walking speed increased. Thus, impulse was maintained across walking
233 speeds by an increasing force level under the big toe due largely to FHL activity. With
234 increasing walking speed, **big toe flexion impulse** relative to total foot impulse increased,
235 highlighting the importance of FHL at higher speeds. On the other hand, in isometric tasks, as
236 plantarflexion torque increased, significant changes in relative force under the big toe were
237 not observed, indicating that the role of FHL did not change across torque levels.

238

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

247

248 During isometric plantarflexion FHL functions as **an important** deep plantarflexor (Klein et
249 al., 1996). In this isolated contraction mode, forces are not generated at proximal joints, and
250 thus energy conservation from proximal joints is minimal compared to the situation during
251 locomotion. In this study, EMG activity increased in parallel in the gastrocnemius and FHL
252 muscles with increasing speed. Gastrocnemius has been reported to transmit forces towards
253 the forefoot (Chen et al., 2012; Fukunaga et al., 2001). In the second half of the stance phase
254 elastic energy reuse in the longitudinal arch buttresses the generation of positive work (Ker et
255 al., 1987). The main contributor to this so-called windlass mechanism (Hicks, 1954) is the
256 plantar aponeurosis (Erdemir et al., 2004), with an increased tension level at higher walking

257 speeds due to the more dorsiflexed toes in the push-off phase (Caravaggi et al., 2010). In vivo
258 motion analyses have shown that at higher speeds, the foot becomes stiffer and so foot length
259 decreases (Stolwijk et al., 2014). The decrease in foot length can be achieved by increased
260 tension of the tissues on the plantar surface. In addition to the passive elements, further
261 increases in foot stiffness are achieved by increasing the activity of the foot flexor muscles.
262 Kelly et al. (2014) found increased EMG activity of the abductor hallucis, flexor digitorum
263 brevis and quadratus plantae with increasing external load applied to the longitudinal arch,
264 suggesting that intrinsic foot muscles serve to decrease stress on the passive elements and
265 increase foot stiffness. In our study, we found that activity in the largest extrinsic foot flexor
266 also increased with increasing walking velocity. Since this muscle contributes to maintenance
267 of the medial longitudinal arch (Thordarson et al., 1995), it may also contribute to foot
268 stiffness.

269

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

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

288

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

298

299 In the isometric conditions, the FHL/SOL ratio at 20 Nm was significantly higher than at
300 higher isometric torque levels. However, absolute FHL activity increased gradually with
301 increasing torque level, suggesting that soleus may not contribute substantially to
302 plantarflexion torque at low torque levels. It should be noted that the FHL/SOL ratio was
303 about two times greater in isometric tasks than in walking, whereas FHL/MG and FHL/LG
304 were twofold higher in walking than in sustained isometric plantarflexions. Otherwise, FHL
305 activity relative to the individual triceps surae muscles did not change generally across torque

306 levels nor with increasing walking speed. These results highlight the task-dependent function
307 of the FHL muscle.

308 **LIMITATIONS**

309

310 **5. Conclusion**

311

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

318

319 **Conflict of interest statement**

320

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

322

323 **References**

324

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Figure Captions

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

404

405 Figure 2. Force and impulse in different tasks. A, C and E are calculated for the area under the
406 big toe. Relative values (B, D, F) present force and impulse under the big toe relative to the
407 same values for the entire plantar surface. For walking peak force and impulse are normalised
408 to body mass (BM) and calculated in the push-off phase (C, E). Statistical differences: *
409 $p < 0.05$, ** $p < 0.01$.

410

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

414

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