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Original article

Effect of footwear on intramuscular EMG activity of plantar flexor muscles in walking

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Abstract

One of the purposes of footwear is to assist locomotion, but some footwear types seem to restrict natural foot motion, which may affect the contribution of ankle plantar flexor muscles to propulsion. This study examined the effects of different footwear conditions on the activity of ankle plantar flexors during walking. Ten healthy habitually shod individuals walked overground in shoes, barefoot and in flip-flops while fine-wire electromyography (EMG) activity was recorded from flexor hallucis longus (FHL), soleus (SOL), and medial and lateral gastrocnemius (MG and LG) muscles. EMG signals were peak-normalised and analysed in the stance phase using Statistical Parametric Mapping (SPM). We found highly individual EMG patterns. Although walking with shoes required higher muscle activity for propulsion than walking barefoot or with flip-flops in most participants, this did not result in statistically significant differences in EMG amplitude between footwear conditions in any muscle ($p > 0.05$). Time to peak activity showed the lowest coefficient of variation in shod walking (3.5, 7.0, 8.0 and 3.4 for FHL, SOL, MG and LG, respectively). Future studies should clarify the sources and consequences of individual EMG responses to different footwear.

1. Introduction

Walking is a fundamental activity of everyday life, and is often performed whilst wearing footwear. Modern footwear is typically designed to assist with locomotion by providing protection, stability, and comfort to the ankle-foot complex. To achieve these aims, typical footwear often has characteristics such as arch support, cushioning, a closed nature, toe spring, elevated and cushioned heel, stiff heel counter and a somewhat stiff sole. However, because of its supporting properties, modern footwear may also restrict natural foot motion in locomotion (Morio et al., 2009). For example, built-in arch support seems to restrict the physiological movement of the medial longitudinal arch during locomotion, which may lead to stiffer forefeet, large inter-individual variability in arch height, and arch-related problems, as observed in

habitually shod individuals compared to habitually barefoot individuals (D'Août et al., 2009; Kadambande et al., 2006). The restrictive nature of some footwear can also contribute to the development of forefoot pathologies such as toe deformities (e.g. Zipfel and Berger, 2007), which can in turn increase the risk of falling in the elderly (Mickle et al., 2009). Wearing stable and supportive shoes seems to also be disadvantageous for the extrinsic foot muscles (e.g. flexor hallucis longus, FHL), since excess foot support may cause them to be under-utilized (McDonald et al., 2016). This can result in reduced strength of foot flexor muscles, leading to an imbalance between foot flexor and extensor muscles, which has also been suggested to cause toe deformities (Myerson and Shereff, 1989). Additionally, FHL functions as an important ankle plantar flexor muscle (Friederich and Brand, 1990; Goldmann et al., 2013; Kura et al., 1997), so changes in FHL function presumably affect the function of other plantar flexor muscles such as the triceps surae (e.g. Heikkinen et al., 2017).

Contrary to shod walking, conditions that are more similar to barefoot walking, such as walking in minimalistic footwear and flip-flops, seem to restrict foot function less. Flip-flops, characterized by a thin, flat-shaped, and flexible rubber sole attached to the foot with a Y-shaped strap in the forefoot region, are often used in everyday life. Walking in flip-flops results in increased ankle joint and subtalar joint range of motion compared to shod walking (Chen et al., 2018). Increased ankle joint range of motion has also been observed when walking barefoot compared to shod walking (Chen et al., 2018; Wirth et al., 2011; Zhang et al., 2013). Presumably, altered kinematics is associated with an altered relative contribution of ankle and foot muscles (Farris and Sawicki, 2012; Winter, 1983). For example, regularly wearing minimalistic footwear for five months increased the cross-sectional area and strength of long and short toe flexor muscles, with the biggest improvements observed in FHL, but comparably smaller improvements in the triceps surae (Brüggemann et al., 2005). Nonetheless, barefoot and shod walking seems to show generally similar electromyography (EMG) activity of tibialis

posterior, flexor digitorum longus and peroneus longus (Akuzawa et al., 2016). Previous comparisons of walking in flip-flops and barefoot have found no difference in medial gastrocnemius (MG) (Burgess and Swinton, 2012) and peroneus longus (Price et al., 2014) EMG activity, and general gait kinematics (such as knee, ankle and subtalar joint kinematics, and knee and ankle joint loading) (Chen et al., 2018). It is currently unclear whether EMG activity patterns differ between FHL and triceps surae muscles when walking in flip-flops versus walking barefoot. Altered FHL function has been linked to Achilles tendon rupture (Finni et al., 2006) and flatfoot (Angin et al., 2014), highlighting the need to understand the relative activity of FHL and triceps surae muscles in these conditions. It is reasonable to assume that the toes are responsible for holding the flip-flops on the foot, which may require a larger relative contribution from toe flexor muscles, of which FHL has the largest cross-sectional area (Friederich and Brand, 1990; Kura et al., 1997).

The aim of this study was to examine the acute effect of shod, barefoot and flip-flops walking on FHL and triceps surae EMG activity. Due to increased leg compliance in shoes, we hypothesized that there would be higher triceps surae EMG activity in shoes compared to the other conditions. Nonetheless, we expected that walking in shoes would result in lower FHL EMG activity because of the restrictive nature and medial longitudinal arch support of shoes. Conversely, walking in flip-flops was expected to result in higher FHL EMG activity compared to the other conditions because of the suspected role of FHL to hold the flip-flops on the foot and due to less restricted foot motion in flip-flops than in shoes. Due to the restricted forefoot motion in shoes, we further expected that the time to peak activity would vary less between individuals in shod walking than in other conditions.

2. Materials and methods

2.1. Participants

Ten healthy volunteers participated in this study (age: 29.6 ± 7.4 years; height: 1.74 ± 0.12 m; body mass: 70.6 ± 12.7 kg), six males and four females. Participants had no known current or recent (< 6 months) leg or foot injuries or deformities (e.g. self-reported flatfoot), and were habitually shod walkers. The Stockholm regional ethics committee approved the study protocol (approval nr: 2017/261-31/4). All subjects gave written informed consent before participation. The study was performed in accordance with the Declaration of Helsinki.

2.2. Study protocol

In the familiarization session (1-3 days before data collection) EMG electrode locations were marked with a permanent marker. During the subsequent measurement session, intramuscular EMG electrodes were inserted and hereafter participants warmed up by performing submaximal and maximal isometric plantarflexion contractions in an isokinetic dynamometer. In addition, they walked at their preferred-speed for 5 minutes. Then, data were collected during overground steady-speed walking in shoes, barefoot, and in flip-flops, while intramuscular EMG activity was recorded from the FHL, soleus (SOL), MG, and lateral gastrocnemius (LG) muscles of the right leg.

2.3. Walking conditions

To define walking speeds, custom-made photocells were placed at the beginning and end of the 7 m long measurement area. To maintain quasi-constant speed over this area, participants started and ended the walking trials 2-3 meters before and after the photocells where they could accelerate and decelerate, respectively. Participants walked in their own sports shoes, barefoot (in socks), and in standardized flip-flops (without socks) (Figure 1). Walking in one's own shoes aimed to reflect real-life situation and eliminate the effects of adaptation to a new shoe that might interfere with the obtained data (Sacco et al., 2010). Self-selected walking speed seems to be highly repeatable (Boonstra et al., 1993; Kadaba et al., 1989; Stolze et al., 1998; Wirth et al., 2011). Thus, participants first walked at self-selected speed in shoes through the

measurement area five times. The slowest and fastest trials were discarded and the remaining three trials were averaged to define preferred shod walking speed. In previous studies, walking barefoot compared to shod resulted in slower preferred walking speed (Chen et al., 2018; Wirth et al., 2011), while barefoot and flip-flops walking showed similar preferred walking speeds (Chen et al., 2018; Price et al., 2014; Sharpe et al., 2016). Therefore, participants of this study walked barefoot and in flip-flops at two speeds in random order: the same speed as preferred shod walking (matched) and at self-selected speed, three successful trials from each. From the matched conditions, trials were accepted within $\pm 5\%$ of target speed.

2.4. Instrumentation

2.4.1. EMG activity

Intramuscular EMG activity of FHL, SOL, MG, and LG was measured with a telemetric system (MyoSystem 1400A, Noraxon Inc. Scottsdale, AZ, USA; sampling frequency: 3000 Hz). A surface reference electrode (silver-silver chloride BlueSensor N; Ambu, Ballerup, Denmark) was placed on the tibia. EMG signals were wirelessly transmitted to an A/D converter (Cambridge Electronic Design, Cambridge, UK), which was connected to a personal computer. Digital signals were collected in Spike2 (Cambridge Electronic Design, Cambridge, UK).

After cleaning the skin with alcohol, intramuscular fine-wires were inserted by an experienced radiologist under the guidance of real-time, high-resolution B-mode and doppler ultrasonography (Logiq E9, GE, USA). Two wires were inserted in each muscle in bipolar configuration with hypodermic needles (0.8 mm diameter, used for wire insertion only) with an inter-tip distance of ~ 5 mm. Wires were Teflon-coated seven-stranded silver hook-wire electrodes with a diameter of 0.25 mm. At the end of the wires a 2 mm region of Teflon coating was stripped off to form the recording surface. Wires and needles were previously sterilized. For MG and LG, electrodes were inserted at the locations where surface electrodes are typically placed (Hermens et al., 2000), with small adjustments in cases of rich vascularity seen by

Doppler ultrasonography. Surface EMG electrodes were also placed over each examined muscle (data not included in this paper) and intramuscular electrodes were inserted right underneath the surface electrodes for the triceps surae muscles (Péter et al., 2019). Since SOL surface EMG may be prone to cross-talk when the electrodes are placed medially (e.g. Bogey et al., 2000), both surface and intramuscular EMG electrodes of SOL were placed laterally. FHL electrodes were inserted 5-10 cm proximally from lateral malleolus on the lateral side of the shank (Figure 2).

2.4.2. Force measurement

Half way along the 7-meter measurement area two force platforms (0.6 m x 0.4 m each; Kistler type 9281EA, Kistler AG, Winterthur, Switzerland; sampling frequency: 3000 Hz) were positioned in series to record three-dimensional ground reaction forces (GRF). Data from the stance phase of the right foot were collected with Qualisys Track Manager software (Qualisys AB, Sweden) and were used to define the time of heel contact (HC) and toe-off (TO), as well as the start of the push-off phase for each step. One successful stride per trial was recorded resulting in three strides per condition.

2.4.3. Plantar pressure

Plantar pressure under the right foot was measured with a Pedar-X insoles (99-sensor, Novel Inc., Munich, Germany; sampling frequency: 100 Hz) to define the timing of HC and TO for those steps that did not hit either of the force platforms but were still performed within the measurement area. Insoles were put inside the socks for barefoot walking. In flip-flops, insoles were not used but event timings were estimated based on EMG onset/offset and GRF data (see below). Plantar pressure data were sent to a personal computer via Bluetooth using the Pedar software. Before every trial an assistant pressed down the big toe to ensure that insoles did not move (Péter et al., 2015). To synchronize data collection a trigger signal was sent to Spike and

Qualisys Track Manager software at the start of the recordings. Synchronisation delays were defined by the laboratory technician and were accounted for before data analysis.

2.5. Analysis

2.5.1. EMG activity

Data analyses were performed in Matlab (MathWorks Inc., Natick, MA, US) and were limited to the stance phase, as plantar flexors are mainly activated in this phase. EMG signals were band-pass filtered (20-500 Hz, zero-lag 4th order Butterworth filter), and were rectified. The signals were then smoothed using a 10 Hz 4th order Butterworth zero-lag filter, and time-normalized (1-101 frames) for each stance phase using linear interpolation. Figure 3 shows a typical example of EMG activity in each step and walking condition recorded from one participant. The resulting curves of all steps per condition were averaged for each participant and muscle. These curves were then normalized to the peak value of the curve from the shod preferred condition (i.e. peak activity in shod represents 100% for each subject and muscle) (Cronin et al., 2015). The time between HC and peak activity (hereafter ‘time to peak activity’), and corresponding inter-individual coefficients of variation (CV%) were also calculated as a measure of inter-individual variability. Due to technical issues, MG EMG activity of one participant and LG EMG activity of another participant were not recorded.

2.5.2. Event timings

To define the timing of HC and TO for each step during shod and barefoot walking, GRF and plantar pressure data were used. For the steps that hit the force platforms (i.e. 3 steps per condition), HC and TO were defined based on a 10 N vertical GRF threshold (Osis et al., 2016). For other steps, vertical GRF was denoted by the sum of forces from the 99 insole sensors, and these force curves were up-sampled to 3000 Hz to line up force plate and insole data. Pressure insoles have inherent inaccuracy when defining timing and amplitude of vertical GRF, partly due to mismatch between foot and insole size. Therefore, HC and TO timings obtained from

force plate data were used to define thresholds for the vertical forces measured with the insole, which was possible after synchronising force plate and insole recordings. The start of the push-off phase was defined based on the horizontal component of the GRF.

In flip-flops, event timings were defined based on EMG offset and GRF, as follows. First, EMG onset and offset were defined for each muscle and step by amplifying the signal-to-noise ratio of the raw band-pass filtered EMG data using Teager-Kaiser Energy Operator (Solnik et al., 2010), and then applying approximated generalized likelihood ratio (AGLR) (Staudé and Wolf, 1999). The AGLR algorithm defines the onset and offset timings based on hypothesis testing as detailed elsewhere (Staudé and Wolf, 1999). Then, for each of the three steps on the force plates, EMG onset was defined relative to HC, and EMG offset was defined relative to TO for each muscle. The muscle with the lowest variability in EMG onset/offset (i.e. most reliable muscle) was chosen as a reference to define the HC and TO of the steps that did not hit the force plates. This analysis showed that EMG onset was less consistent than EMG offset, therefore the latter was used in all subjects. For this, the mean EMG offset of the three steps of the muscle with the lowest SD between steps, as well as step durations were used to estimate HC and TO timings for the steps that did not hit the force plates. The lowest variability between steps was in FHL, MG, and LG in two, four, and four subjects, respectively, with a standard deviation of 23.6 ± 15.9 ms (range = 6.6 – 64.0 ms) between steps. EMG offset for the selected muscle was 378.4 ± 108.6 ms (range = 249.7 – 610.7 ms) before TO, which was used in this analysis. Although this approach has some uncertainty, this also increased the number of analysed steps to 12 ± 1 per participant, likely resulting in more robust estimation of EMG activity as opposed to relying on three steps only.

2.6. Statistical analysis

For each of the barefoot and flip-flops conditions, statistical differences between preferred and matched conditions were tested for walking speed, stance phase duration, and push-off duration

with paired samples t-tests (SPSS, IBM New York, NY, US). Since there was no difference ($p>0.05$) between preferred and matched conditions in any of these variables, only the trials at preferred speed were included in further analysis.

2.6.1. Curve analysis

Statistical Parametric Mapping (SPM, (Friston, 2007)) was used to locate differences in intramuscular EMG amplitudes of each muscle between footwear conditions. With SPM we tested the EMG amplitudes at each point of the time-normalised stance phase. This analysis was performed in Matlab (open-source spm1d code, v.M0.1, www.spm1d.org). SPM one-way repeated-measures ANOVA was used as follows. First, to form a Statistical Parametric Map the scalar output statistic $SPM\{F\}$ was calculated for each time point. Then, to test the null hypothesis, the critical threshold was calculated beyond which only α % (set at 5%) of the smooth random curves would be expected to traverse. This critical threshold was calculated based on trajectory smoothness estimates via temporal gradients (Friston, 2007), and based on this smoothness, Random Field Theory expectations regarding the field-wide maximum (Adler and Taylor, 2007). If any values of $SPM\{F\}$ exceeded the critical threshold, EMG time-series were considered to be significantly different.

2.6.2. Time to peak activity

Normality of data distribution was tested by Shapiro Wilk's W test in SPSS. Statistical differences between conditions were tested for each muscle with one-way repeated measures ANOVA and the locations of the differences were tested after Bonferroni correction. Significance level was set at $P<0.05$.

3. Results

Walking speed, stance phase duration and push-off duration did not vary between the three footwear conditions (Table 1). At the group level, we found no differences in EMG activity between footwear conditions in any muscle (Figure 4). This might be due to large inter-

individual variability in the differences in EMG amplitudes between different footwear conditions (Figure 5), which encouraged to perform exploratory analysis at an individual level. This showed that in 66% of all conditions (25 out of 38 conditions [four muscles x ten individuals – 2 missing data]) the peak activity was the highest in shod walking and in 58% of the conditions (22 out of 38 conditions) it was the lowest in barefoot walking. For these subjects the differences between shod and barefoot peak activity were 65%, 32%, 16%, and 42% in FHL, SOL, MG, and LG, respectively. In comparison, at the group level, EMG activity in shod walking was higher than in barefoot by 26%, 19%, 7% and 29% in FHL, SOL, MG, and LG, respectively. CV% in the time to peak activity was the lowest in shod walking for all muscles (Figure 6).

4. Discussion

This study examined the acute effects of walking barefoot and in flip-flops on intramuscular EMG activity of plantar flexor muscles in habitually shod individuals. At the group level, no differences were detected between footwear types in any of the muscles when analysing across the time-normalised stance phase. The directions of changes in the magnitude of EMG activity were highly individual. Most participants showed the largest peak activity in all muscles in shod walking, and the lowest peak activity in barefoot walking. Inter-individual variability of the time to peak activity was lowest in shod walking in all muscles.

Besides examining the general effects of different footwear on ankle plantar flexor activity, defining the relative activity of FHL compared to triceps surae muscles was the primary interest in this study. Since participants in this study were habitually shod and showed no clear signs of foot abnormalities, we expected consistent effects of the different footwear conditions across individuals. However, individual differences were evident, which may partly explain why differences were not detected at the group level. In most cases, the highest peak EMG activity was found in shod walking and the lowest peak activity was found in barefoot walking in all

muscles, including FHL. This finding supports the idea that footwear may acutely increase leg compliance (Kelly et al., 2016), which requires increased activity from muscles that contribute to propulsion, which was recently observed in intrinsic foot muscles (Kelly et al., 2016). Conversely, the lower plantar flexor activity that we observed in barefoot walking may be associated with lower ankle stiffness compared to shod walking. Contrary to the triceps surae muscles, FHL spans the medial longitudinal arch and attaches on the distal phalanx of the big toe. It is assumed that this muscle supports the function of the medial longitudinal arch (Sarafian, 1993) together with passive elements such as the plantar fascia, while the contribution of intrinsic foot muscles seems to be negligible (Farris et al., 2019). We speculate that the absence of shoes allows the passive structures to better contribute to medial longitudinal arch function, decreasing the need for FHL activity. The anatomy of FHL further suggests a potential role in foot inversion, which is more pronounced in barefoot than in shod walking (Morio et al., 2009). However, although foot kinematics were not measured in this study, our results suggest that FHL is not the primary contributor to the acute increase in foot inversion, because FHL EMG activity was lower in barefoot than in shod walking. Future studies should examine the source of individual differences in muscle activity from a larger sample size, although this may be challenging, especially for FHL (Péter et al., 2019). Nonetheless, acute responses to walking barefoot instead of in regularly worn shoes may not reflect long-term adaptations to barefoot walking. For example, in long-term, walking in minimalistic shoes increased ankle plantar flexion and toe flexion strength (Brüggemann et al., 2005). These changes were associated with increased cross-sectional area of FHL and flexor digitorum longus muscles, which implies higher muscle activity in long-term.

We found the lowest inter-individual variability in time to peak activity in shod walking in all muscles. This is consistent with the notion that footwear restricts natural and individual-specific foot motion, imposing a fairly consistent motion pattern for each individual in the push-off

phase (Morio et al., 2009). Habitually shod individuals are used to the foot-supporting and stabilizing properties of their own footwear, which may also explain lower inter-individual variability. Furthermore, arch height of habitually shod individuals shows high inter-individual variability as compared to habitually barefoot individuals (D'Août et al., 2009). Considering that each habitually shod individual is used to different types of footwear and has a correspondingly different foot structure, each person may respond to the absence of shoes differently. This may be why acute or short-term footwear studies on habitually shod individuals seem to result in no, unexpected or inconsistent effects on foot and lower leg muscle function. This may also explain why we found no difference in EMG amplitudes at the group level between the different footwear conditions, and why we found relatively high inter-individual variability in the time to peak activity when walking barefoot and in flip-flops. To better understand the effects of habituation to a given type of footwear, long-term studies are needed. Indeed, intervention studies have shown consistent changes across individuals after walking or running barefoot or in minimal footwear for several weeks. These changes include increased strength and size of foot muscles, increased foot and leg stiffness, and greater use of the spring-like function of the medial longitudinal arch (Brüggemann et al., 2005; Chen et al., 2016; De Wit et al., 2000; Johnson et al., 2015; Miller et al., 2014; Perl et al., 2012; Ridge et al., 2018).

Due to the somewhat invasive nature of this study, the sample size was relatively low, potentially resulting in type II errors. However, group-level analyses may cover up individual responses to different footwear. Indeed, inter-participant differences in response to different footwear were evident. In gait, EMG activity patterns seem to be highly individual (Hug et al., 2019), which may be related to anatomical features such as muscle size proportions, moment arms, and calcaneus length (Ahn et al., 2011). Variable response to different footwear in this study implies the need for examining the effects of footwear on EMG patterns on a larger

sample, where clusters could potentially be identified and related to other parameters such as differences in walking kinematics in different footwear. As another potential source of inter-participant differences, intramuscular EMG electrodes have a small pick-up volume, so the measured EMG activity may not represent the activity of the whole muscle. The use of high-density surface EMG could minimise this error (e.g. dos Anjos et al., 2017; Staudenmann et al., 2009) in future studies, although this method cannot be used for deep muscles such as FHL. In this study, intramuscular EMG was chosen to minimise cross-talk, which may be especially important for FHL where the region for surface electrode placement is relatively small (Péter et al., 2015). Furthermore, intramuscular EMG activity was sampled from a larger proportion of FHL than the other muscles thus intramuscular EMG may be more representative of overall muscle activity for FHL than the other, larger examined muscles. In future studies the magnitude of the pick-up volume with respect to the target muscle volume should be taken into account. Furthermore, shoe properties (e.g. sole stiffness) might have been different between participants. Nonetheless, allowing participants to wear their own shoes was preferred since it eliminates the possibility of acute adaptation to new shoes. Wearing socks in barefoot walking could somewhat affect the sensation of the foot, but this was a necessity in order to secure the pressure insoles. In flip-flops, pressure insoles could not be used, which might have affected the definition of heel contact and toe-off timings. These events were estimated based on the EMG offset of the steps performed over the force plates in flip-flops for each participant. Although the standard deviation of EMG offset was relatively low, the mean of a small number of steps has relatively large uncertainty, which likely caused an estimation error when we defined the HC and TO timings in flip-flops for the steps which did not hit the force plates. Furthermore, we examined healthy individuals without clear foot deformities, so the application of our results may be restricted to this population.

In conclusion, this study did not detect differences in plantar flexor EMG activity between different footwear conditions in walking. Exploratory analysis of individual responses suggests that in most individuals, walking in shoes at preferred speed requires higher peak EMG activity for propulsion from the ankle plantar flexors compared to walking barefoot or in flip-flops. Furthermore, similar time to peak activity between individuals may be due to that shoes restrict the natural and individual-specific function of the foot and ankle, superimposing a particular motion pattern over the individual habitual motion pattern. Future studies should examine the long-term effects of walking barefoot and in flip-flops on ankle plantar flexor function to further understand the effects of footwear on the relative role of these muscles.

Declaration of competing interest

None of the authors has any financial or other conflicts of interest to declare.

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Table and figure legends

Table 1: Number of analysed steps and their characteristics are shown for the three walking conditions at preferred speed, and for barefoot and flip-flops walking conditions at the same speed as preferred shod walking (matched). Number of included steps is expressed as median \pm interquartile range, while other values are expressed as mean \pm standard deviation.

Figure 1: Standard thong-style flip-flops used in this study. Sole thickness was 10 and 15 mm at the front and back, respectively.

Figure 2: Intramuscular electromyography electrodes were inserted with real-time B-mode and Doppler ultrasonography guidance in flexor hallucis longus (FHL), soleus (SOL), medial (MG) and lateral gastrocnemius (LG) of the right leg. The recording tips of the wires were 3-5 cm distal to the insertion points.

Figure 3: Smoothed intramuscular electromyography (EMG) activity of one participant showing all steps for each muscle and walking condition at preferred walking speed before amplitude normalisation. Step cycles range from heel contact (0%) to toe-off (100%).

Figure 4: Electromyography (EMG) activity (mean \pm standard deviation, SD) for all conditions in each muscle from heel contact to toe-off. EMG activity was normalised to the peak activity of preferred shod walking (%peak). Statistical Parametric Maps (SPM) show the comparisons between the footwear conditions for each muscle, where SPM{F} trajectories reflect the magnitude of the test statistics. Thresholds (dashed, F*) were not exceeded implying no difference between conditions.

Figure 5: Intramuscular electromyography activity (normalized to peak activity of preferred shod walking, %peak) for each participant (#1-10) in all muscles and walking conditions. Step cycles start from heel contact and end at toe-off. Each trace represents the average of all steps per condition.

Figure 6: Individual (solid) and group mean (bold and dashed) values of time to peak activity relative to stance phase duration (%stance) for each muscle and footwear condition. Inter-individual coefficients of variation (CV%) are presented.

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Table 1: Characteristics of the analysed steps.

	Preferred			Matched	
	shod	barefoot	flip-flops	barefoot	flip-flops
Included steps (n)	12.00 ± 2.50	12.50 ± 4.00	12.00 ± 1.00	12.00 ± 2.00	12.00 ± 0.00
Speed (m/s)	1.43 ± 0.19	1.38 ± 0.22	1.41 ± 0.21	1.43 ± 0.20	1.43 ± 0.20
Stance phase duration (s)	0.67 ± 0.06	0.63 ± 0.08	0.68 ± 0.07	0.61 ± 0.06	0.67 ± 0.05
Push-off phase duration (s)	0.36 ± 0.04	0.36 ± 0.05	0.37 ± 0.07	0.34 ± 0.04	0.37 ± 0.05
Start of push-off (% of the step cycle)	53.89 ± 3.34	56.32 ± 4.29	53.86 ± 5.06	55.41 ± 3.44	54.95 ± 3.73

Figure 1



Figure 2

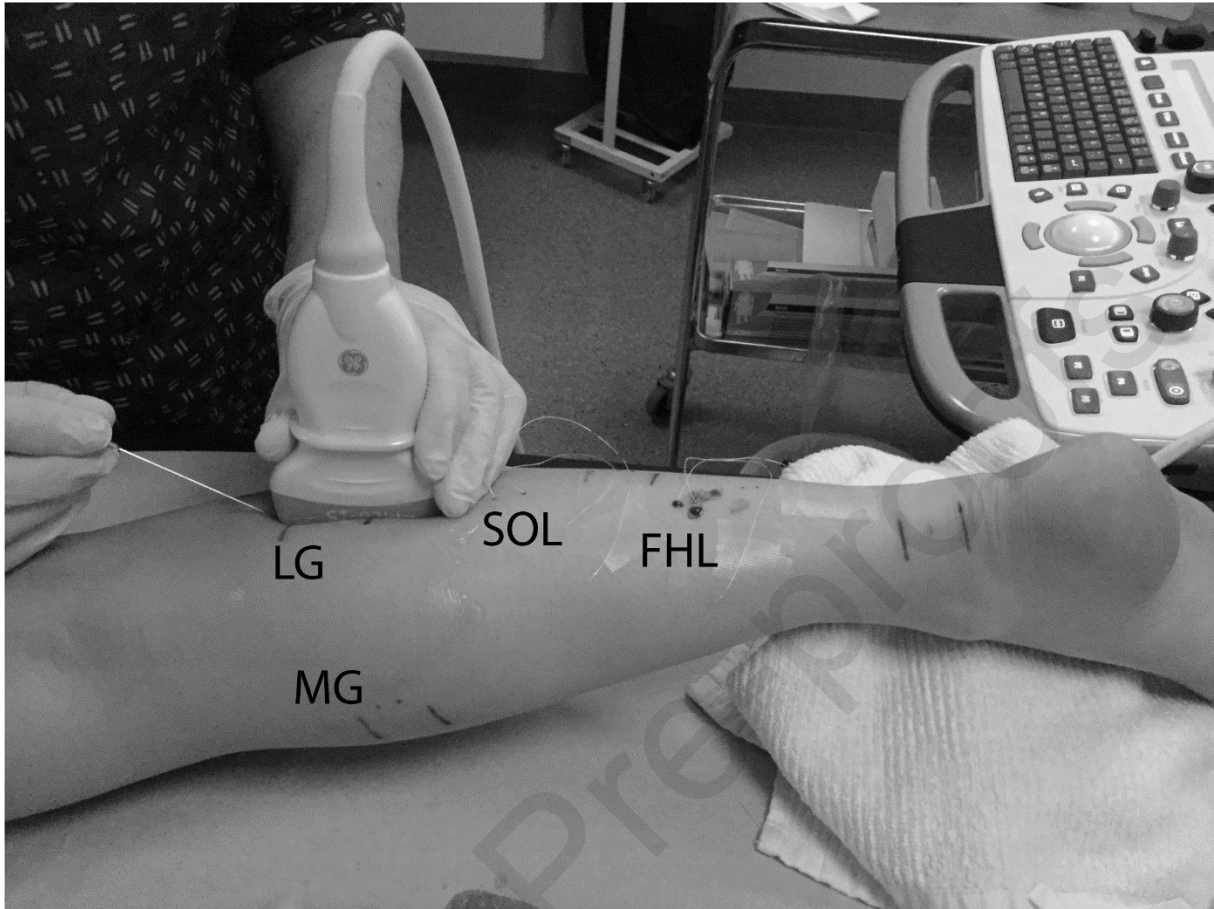


Figure 3

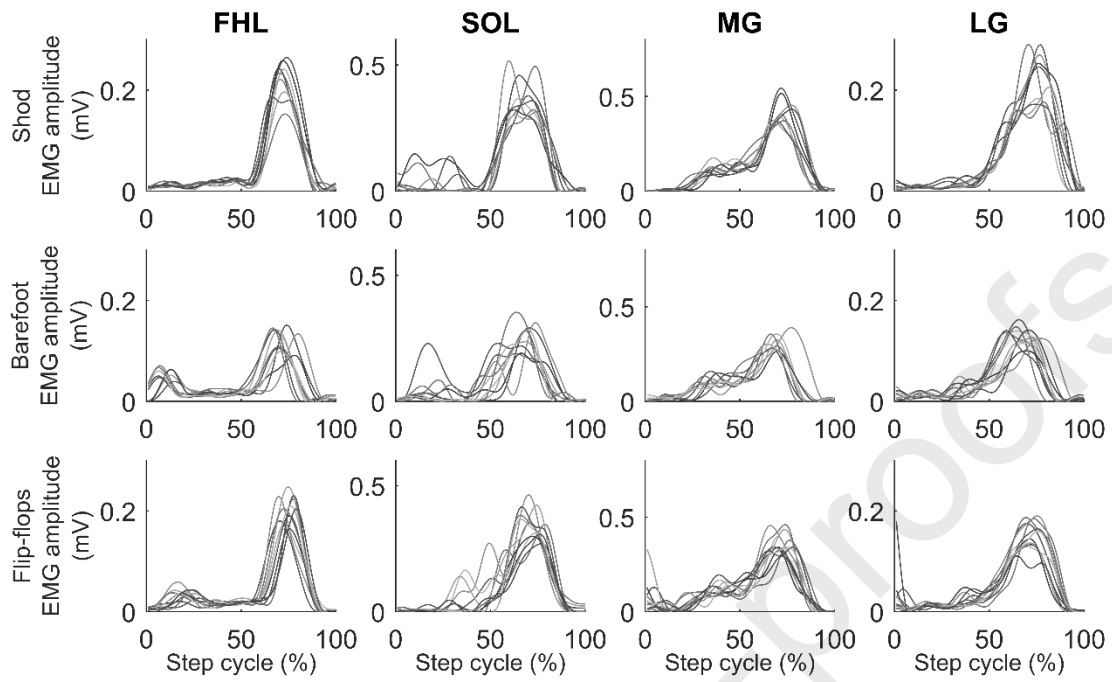


Figure 4

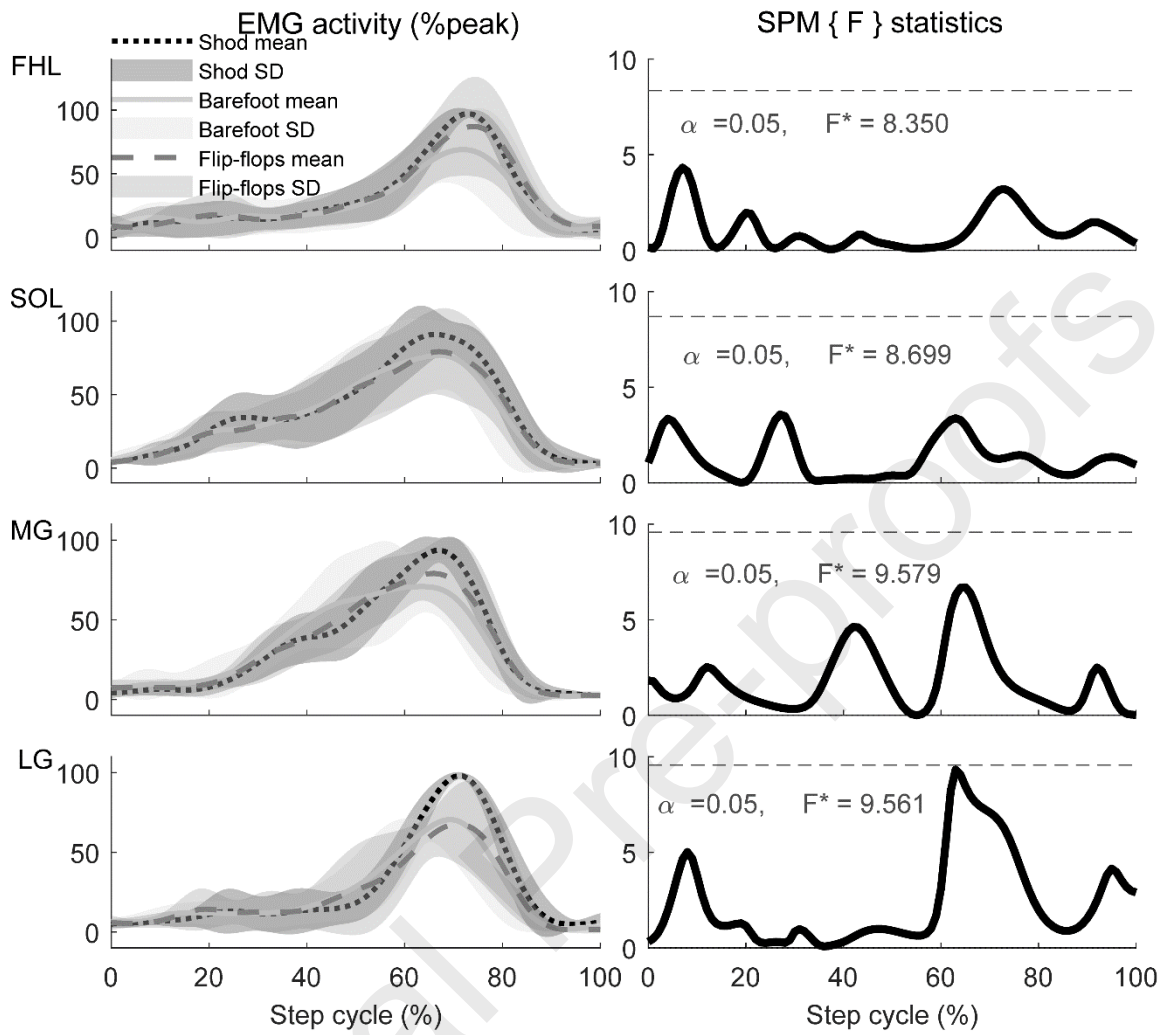


Figure 5

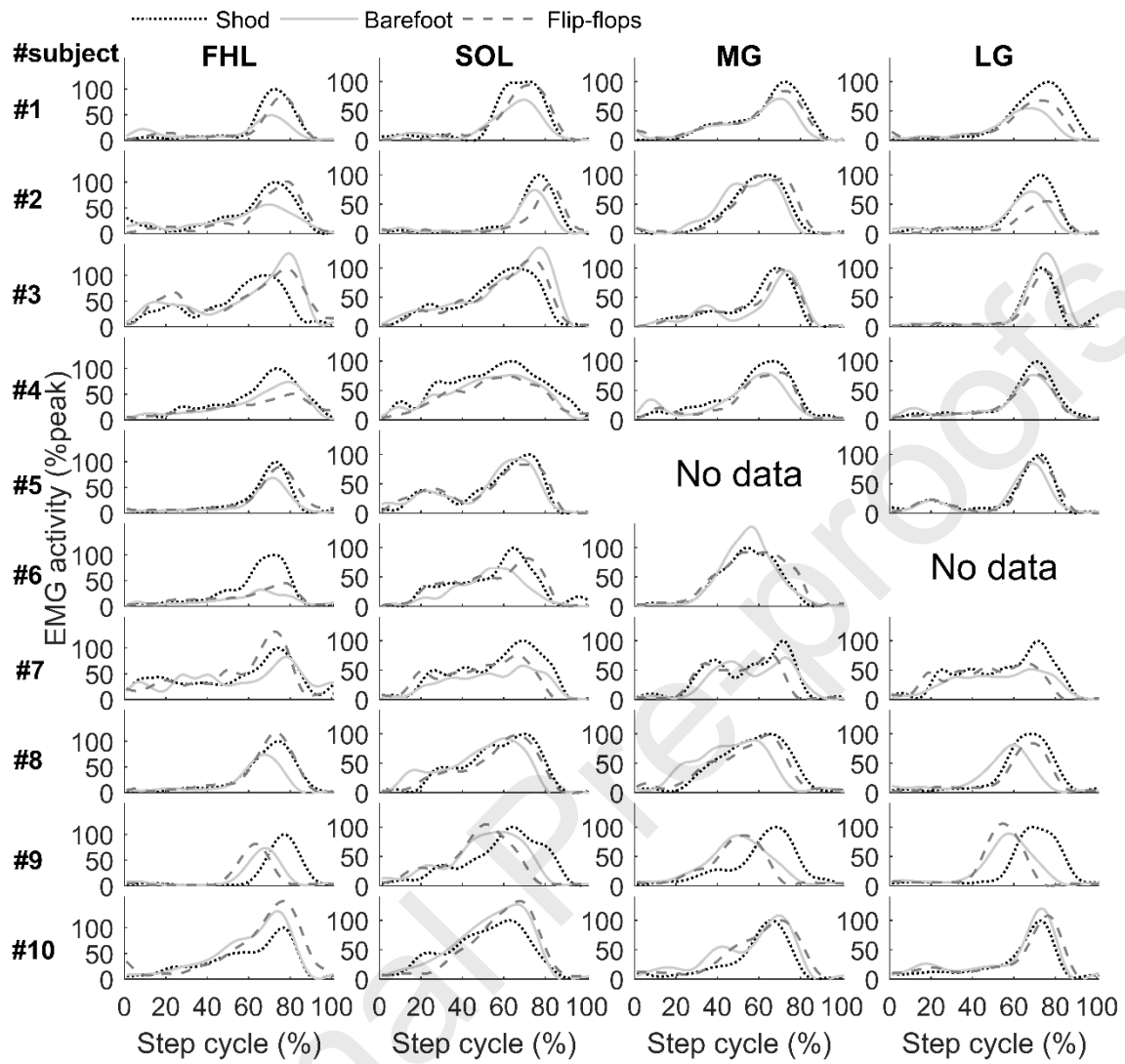
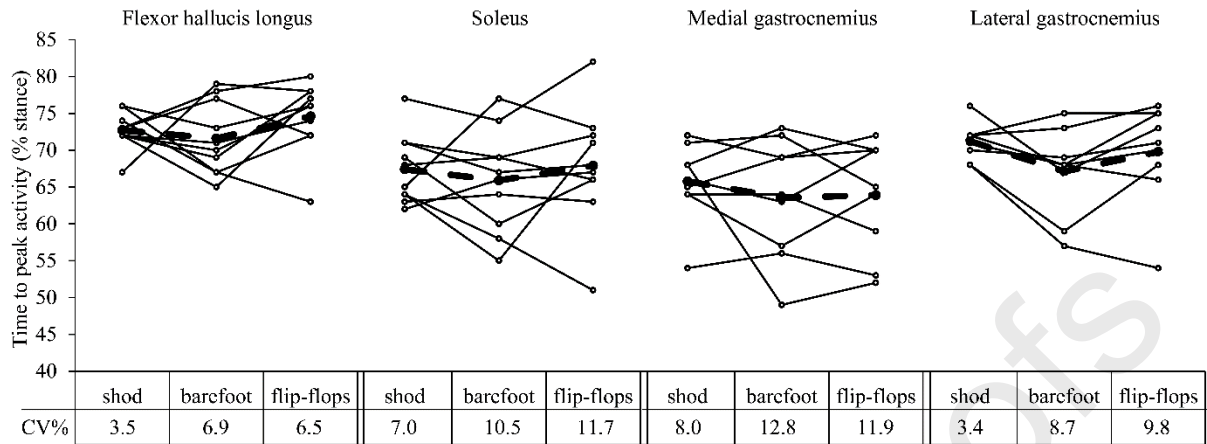


Figure 6



Author Biography



Annamária Péter is a post-doctoral researcher at the University of Jyväskylä, Finland. She received her MSc in kinesiology in Hungary in 2013 from the Semmelweis University then her PhD in 2019 from the University of Jyväskylä. Her research interests include gait analyses, lower leg muscle activation and muscle-tendon interactions in healthy and clinical conditions such as Achilles tendon rupture.



Anton Arndt performed his undergraduate studies in New Zealand and Australia in biology and Human Movement Sciences before receiving a scholarship for a PhD at the German Sport University, Köln. The PhD involved studies concerning asymmetrical loading of the Achilles tendon. This line of study continued at the Karolinska Institute in Sweden as a post-doc. At present Toni Arndt is a professor in biomechanics, specializing in muscle-tendon function and sports biomechanics, at The Swedish School of Sport and Health Sciences (GIH), where he is also Dean of the Research and Doctoral Education Board. He has published approximately 80 peer reviewed scientific articles and supervised eight PhD students to completion and at present supervises six. He is President of the International Society of Biomechanics.



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Taija Finni is a professor of kinesiology and vice dean at the Faculty of Sport and Health Sciences, University of Jyväskylä. She completed PhD in biomechanics at the University of Jyväskylä in 2001 and post-doctoral studies at the University of California. Prof Finni's research ranges from basic neuromuscular function to translational research related to physical activity and sedentary behavior (OpenKin, CHIPASE). Her ongoing projects relate to muscle-tendon function in healthy and injured people (UNRESAT) and training adaptations of children with cerebral palsy (EXECP), for example. She has over 100 publications and has supervised 11 PhD students to completion and has 6 PhD students currently under supervision. She serves as a senior section editor in *Scandinavian Journal of Medicine and Science in Sports* and is an active member of the International Society of Biomechanics and the European College of Sport Science. Homepage: <https://staff.jyu.fi/Members/finni>.



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None of the authors has any financial or other conflicts of interest to declare.