

Accepted Manuscript

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Authors: Simon Franklin, François-Xavier Li, Michael J. Grey



PII: S0966-6362(17)30983-9
DOI: <https://doi.org/10.1016/j.gaitpost.2017.10.027>
Reference: GAIPOS 5856

To appear in: *Gait & Posture*

Received date: 20-12-2016
Revised date: 19-10-2017
Accepted date: 27-10-2017

Please cite this article as: Franklin Simon, Li François-Xavier, Grey Michael J. Modifications in lower leg muscle activation when walking barefoot or in minimalist shoes across different age-groups. *Gait and Posture* <https://doi.org/10.1016/j.gaitpost.2017.10.027>

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Modifications in lower leg muscle activation when walking barefoot or in minimalist shoes across different age-groups.

Simon Franklin ^a, François-Xavier Li ^{a*}, Michael J. Grey ^b

^a School of Sport, Exercise and Rehabilitation Sciences, University of Birmingham, Edgbaston, Birmingham, B152TT, United Kingdom

^b Acquired Brain Injury Rehabilitation Alliance (ABIRA), School of Health Sciences, University of East Anglia, Norwich Research Park, NR4 7TJ, United Kingdom

*Corresponding author at: School of Sport, Exercise and Rehabilitation Sciences, University of Birmingham, Edgbaston, Birmingham, B15 2TT, United Kingdom. Tel: +44 (0)121 414 4114. E-mail address: f.x.li@bham.ac.uk

Highlights

- Minimalist shoes are intermediate to barefoot and conventional footwear
- Stance phase gastrocnemius medialis activity is decreased in supportive shoes
- Initial stance tibialis anterior activity reduced when barefoot/in minimalist shoes
- Reduced peroneus longus activity at initial stance when barefoot in younger ages

Abstract

Ageing is associated with a decline in muscle strength and impaired sensory mechanisms which contribute to an increased risk of falls. Walking barefooted has been suggested to promote increased muscle strength and improved proprioceptive sensibility through better activation of foot and ankle musculature. Minimalist footwear has been marketed as a method of reaping the suggested benefits of barefoot walking whilst still providing a protective surface. The aim of this study was

to investigate if walking barefoot or in minimalist footwear provokes increased muscle activation compared to walking in conventional footwear. Seventy healthy adults (age range 20-87) volunteered for this study. All participants walked along a 7m walking lane five times in four different footwear conditions (barefoot (BF), minimalist shoes (MSH), their own shoes (SH) and control shoes (CON)). Muscle activity of their tibialis anterior (TA), gastrocnemius medialis (GCM) and peroneus longus (PL) were recorded simultaneously and normalised to the BF condition. MSH are intermediate in terms of ankle kinematics and muscle activation patterns. Walking BF or in MSH results in a decrease in TA activity at initial stance due to a flatter foot at contact in comparison to conventional footwear. Walking BF reduces PL activity at initial stance in the young and middle age but not the old. Walking in supportive footwear appears to reduce the balance modulation role of the GCM in the young and middle age but not the old, possibly as a result of slower walking speed when BF.

Keywords: barefoot, footwear, electromyography, gait, ageing

Introduction

As proponents of bipedal locomotion, humans possess an inherently unstable system requiring constant modulation by balance mechanisms in order to prevent falling [1]. For millions of years humans walked barefoot (BF) and the feet have evolved to cope with the demands of bipedal locomotion. The human foot comprises 104 cutaneous mechanoreceptors responsible for sensing changes in pressure, vibration and skin stretch and their distribution highlight their role in balance and

movement control [2]. These plantar mechanoreceptors contribute to the reflex modulation of the phases of gait via the detection of pressure during foot-ground contact [3] and along with proprioceptive afferents assist in the planning and correction of movement [4]. This information is essential for controlling static and dynamic stability.

Footwear habits have since changed and its suggested modern day footwear may be impairing the capability of these afferent receptors. Highly structured and supportive shoes could limit the input as the foot is not as susceptible to changes in shape, pressure and touch due to the confines placed upon it. This idea has been furthered by Nigg (2015), who hypothesised walking BF activates the smaller muscles within the feet and the larger muscles crossing the ankle joint. He suggested these smaller muscles might provide greater stability as they can more quickly sense changes in different directions and with smaller amounts of force being required [5]. Whilst this position paper primarily focussed on running performance and injuries, the premise of improved stability by activating the smaller muscles, could have implications for the older population in terms of fall prevention.

Wearing footwear may also lead to foot muscle weakening due to the reduction in the stresses put upon the foot by means of supportive features [6]. Ageing causes a decline in muscle strength along with sensory impairments and these factors contribute to the increased susceptibility to falls. Research has shown wearing minimalist shoes (MSH) for athletic training resulted in a significant increase in toe flexor strength [7]. This suggests changing the footwear worn to less supportive, more 'barefoot-like' footwear, may better activate the foot muscles. However, research is required to determine if purely walking in MSH better activates afferent and efferent mechanisms and if this can have a positive influence on stability.

A recent systematic review investigated the effect of footwear, or the lack of footwear, on walking [8]. Aside from outlining the overall kinematic differences between shod and barefoot walking, the review highlighted the paucity of research on BF and MSH use in older age populations and the distinct lack of study on muscle activity differences between shod, minimally shod and unshod conditions.

Consequently, the aims of this study were to investigate if walking BF or in MSH share the same lower leg muscle activation patterns and to determine if greater muscle activation is provoked compared to conventional footwear. We also aimed to determine if there were any differences with respect to age and years spent wearing structured footwear. We hypothesised muscle activation patterns between walking BF and in MSH would be similar and there would be greater activation of the lower leg muscles during the stance phase in these conditions. We also hypothesised the old age group would show a greater increase in muscle activity when walking BF compared to wearing structured footwear.

Methods

70 healthy adults (27 males, age range 20-87years) participated and were split into 3 age groups (YOUNG <40 years (n=20), MID >40 years and <70 years (n=30) and OLD >70 years (n=20) (Table 1). All participants were able to ambulate independently and had no known gait disorders or abnormalities. All participants completed a general health questionnaire and signed an informed consent prior to testing as approved by the University ethics committee (ERN_14-0560).

Kinematic markers were placed bilaterally at the lateral epicondyle (R/LKNE), base of the calcaneus (R/LHEE), medial malleolus (R/LANK) and first metatarsophalangeal joint (R/LTOE). When wearing footwear, markers were

attached to the shoes in the same positions as determined by palpation. Surface EMG electrodes (Wave Wireless EMG, Cometa Systems, Milan) were placed on the right leg over the belly of the tibialis anterior (TA), peroneus longus (PL) and gastrocnemius medialis (GCM) muscles in the positions outlined by the SENIAM guidelines [9]. At each site the skin was shaved, abraded and cleaned with an alcohol swab before attaching two disposable pre-hypoallergenic gelled (1cm diameter) self-adhesive Ag/AgCl snap electrodes with an inter-electrode distance of 1.5cm. The EMG signals were collected at a rate of 2000Hz, amplified with a gain of 1000 (input impedance 20M Ω , CMRR >100dB), and bandpass filtered from 10–1000Hz.

Thirteen Vicon MX cameras (Vicon, Oxford, UK) recording at a sampling rate of 250Hz collected three dimensional kinematic data. Gait cycle phases were computed using the R/LHEE and R/LTOE markers and absolute ankle angle was determined using the foot vector (RANK and RTOE markers) with respect to a vertical vector from the ankle.

Participants walked at a self-selected speed through a 7m walking lane from a mark based on 3 practice trials such that 3 steps were taken prior to data collection commencing. Participants completed 5 trials in each of the four randomly assigned footwear conditions. The footwear were BF, a MSH (Product ID: 2169, Two Barefeet Boarding Co.), a control shoe (CON) (Style Code: 10001, Hobos Womens, Style Code: 50109, Hobos Mens) and the participants own footwear (SH). EMG and kinematics were recorded in synchrony.

[Insert-Figure-1-approximately-here]

Post-processing of the data was completed using custom-written scripts in Matlab (MATLAB, The MathWorks, Natick, MA, USA). Kinematic data were low pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 12Hz. Muscle activity data were zero offset, before being full wave rectified and then low pass filtered using a fourth-order Butterworth filter with a cut-off frequency of 10Hz. Once frequency matched to the synchronised kinematic data, the linear envelope for each participant's trials were cut from right heel strike to right heel strike and normalised to the gait cycle (0-100%). Each trial comprised 1-4 recorded cycles. Maximal voluntary contractions were not completed due to previous reports of poor reliability in achieving a maximum for the PL [10, 11]; therefore all cycles for each participant were collated and normalised to the average of all the cycles when BF. The normalised cycles within each respective trial were ensemble averaged to provide an average muscle activity trace for each trial. Each trial was then divided into stance and swing phases and the stance phase sub-divided into Initial Double Support (IDS), Single Support (SS) and Late Double Support (LDS). The mean activity was then computed for each muscle within each gait cycle phase. Due to recording errors in certain trials resulting in missing data the number of trials available for comparison was limited to 4.

Mixed design repeated measures Analysis of Variance (ANOVA) were completed for each variable to determine the differences across footwear (BF vs MSH vs CON vs SH), trial (1:4) and age group (YOUNG vs MID vs OLD). Mauchly's test of sphericity was completed to ensure validity and in the case where this test was violated a Greenhouse-Geisser correction was applied. Statistical analyses were performed using SPSS V.22 for Windows (IBM Corporation, Somers, NY) with levels of significance set to $p < 0.05$.

Results

[Insert-Table-1-approximately-here]

Tibialis Anterior

In the stance phase there was a significant footwear effect

($F(2.657,177.989)=23.920$, $p<0.001$, $\eta^2_p=0.263$). Walking BF exhibited lower TA activation compared to MSH, CON and SH by $0.096\text{mV}\pm 0.032$, $0.249\text{mV}\pm 0.038$ and $0.242\text{mV}\pm 0.039$ respectively. Walking in the MSH showed lower TA activation than the CON and SH conditions by $0.153\text{mV}\pm 0.035$ and $0.146\text{mV}\pm 0.036$ respectively. After stance phase subdivision there was a significant effect of footwear in the IDS phase ($F(2.190,146.759) = 37.416$, $p<0.001$, $\eta^2_p=0.358$) and SS phase ($F(3, 201) = 20.145$, $p<0.001$, $\eta^2_p=0.231$) but not in the LDS phase. In the IDS phase walking BF lead to lower TA activation in the IDS phase compared to the MSH, CON and SH conditions by $0.238\text{mV}\pm 0.037$, $0.547\text{mV}\pm 0.061$ and $0.489\text{mV}\pm 0.071$ respectively whilst walking in MSH showed lower TA activation than the CON and SH conditions by $0.309\text{mV}\pm 0.056$ and $0.251\text{mV}\pm 0.064$ respectively. In the SS phase walking BF resulted in reduced TA activation during the SS phase compared to the MSH, CON and SH conditions by $0.290\text{mV}\pm 0.059$, $0.338\text{mV}\pm 0.054$ and $0.351\text{mV}\pm 0.053$ respectively.

Gastrocnemius Medialis

In the stance phase there was a significant interaction effect of footwear*age

($F(4.674,156.570)=3.175$, $p=0.011$, $\eta^2_p=0.087$). The YOUNG showed lower GCM activation when wearing CON compared to BF, the MSH and SH conditions by

0.161mV \pm 0.029, 0.194mV \pm 0.039 and 0.108mV \pm 0.031 respectively; the MID had a lower GCM activation when wearing the CON compared to MSH and SH conditions by 0.130mV \pm 0.032 and 0.079mV \pm 0.026 respectively whereas the OLD showed no differences across footwear. Stance phase subdivision displayed a significant footwear*age interaction effect in the SS phase ($F(4.814, 161.253) = 3.085, p=0.012, \eta_p^2=0.084$) and a significant main effect of footwear in the LDS phase ($F(2.198, 147.276) = 14.169, p<0.001, \eta_p^2=0.175$) but no significant differences in the IDS phase. In the SS phase the YOUNG exhibited lower GCM activation when wearing the CON compared to BF, the MSH and SH conditions by 0.210mV \pm 0.031, 0.141mV \pm 0.037 and 0.113mV \pm 0.027 respectively; the MID showed lower GCM activation in the CON compared to the MSH by 0.099mV \pm 0.027 whereas the OLD showed no differences across footwear. Conversely in the LDS phase, walking BF lead to lower GCM activation during the LDS phase compared to the MSH, CON and SH conditions by 0.626mV \pm 0.145, 0.975mV \pm 0.177 and 1.260mV \pm 0.257 respectively.

Peroneus Longus

In the stance phase there was a significant main effect of footwear ($F(2.328, 155.946)=5.335, p=0.004, \eta_p^2=0.074$). Walking BF lead to reduced PL activation compared to the CON and SH conditions by 0.067mV \pm 0.023 and 0.124mV \pm 0.034 respectively. With stance phase subdivision there was a significant interaction effect between footwear*age ($F(5.045, 2.805) = 2.805, p=0.018, \eta_p^2=0.077$) in the IDS phase, a significant main effect of footwear in the LDS ($F(3, 201) = 5.414, p=0.001, \eta_p^2=0.075$) but no significant differences in the SS phase. In the IDS phase the YOUNG had a reduced PL activity when BF compared to the CON and SH

conditions by $0.368\text{mV}\pm 0.086$ and $0.313\text{mV}\pm 0.088$ respectively and also when wearing the MSH compared to CON and SH conditions by $0.278\text{mV}\pm 0.082$ and $0.223\text{mV}\pm 0.075$ respectively. The MID displayed reduced PL activity when BF compared to the MSH, CON and SH conditions by $0.153\text{mV}\pm 0.051$, $0.366\text{mV}\pm 0.070$ and $0.390\text{mV}\pm 0.072$ respectively and when wearing the MSH compared to CON and SH conditions by $0.213\text{mV}\pm 0.067$ and $0.237\text{mV}\pm 0.061$ respectively whilst the OLD showed no differences between footwear. In the LDS phase walking BF lead to lower PL activation compared to the CON and SH conditions by $0.222\text{mV}\pm 0.077$ and $0.238\text{mV}\pm 0.085$ respectively.

[Insert-Figure-2-approximately-here]

Ankle Angle Heel Strike

There was a significant effect of footwear ($F(2,484,166.422)=64.094$, $p<0.001$, $\eta^2_p=0.489$). Walking BF resulted in greater plantar flexion compared to the MSH, CON and SH conditions by $3.118^\circ\pm 0.385$, $5.597^\circ\pm 0.487$ and $5.866^\circ\pm 0.599$ respectively. Walking in the MSH resulted in greater plantar flexion compared to the CON and SH conditions by $2.480^\circ\pm 0.405$ and $2.748^\circ\pm 0.502$ respectively.

Gait Speed

There was a significant interaction effect between footwear*age ($F(6,201)=4.322$, $p=0.002$, $\eta^2_p=0.114$). The YOUNG walked slower BF compared to when wearing the CON and SH conditions by $0.032\text{m/sec}\pm 0.011$ and $0.034\text{m/sec}\pm 0.013$ respectively. The MID walked slower BF than the MSH, CON and SH conditions by $0.038\text{m/sec}\pm 0.008$, $0.067\text{m/sec}\pm 0.009$ and $0.065\text{m/sec}\pm 0.010$ respectively and walked slower in the MSH than the CON conditions by $0.029\text{m/sec}\pm 0.010$. Similarly

the OLD walked slower BF than the MSH, CON and SH conditions by $0.064\text{m/sec}\pm 0.010$, $0.108\text{m/sec}\pm 0.011$ and 0.101 ± 0.013 respectively and walked slower in the MSH than the CON and SH conditions by $0.045\text{m/sec}\pm 0.012$ and 0.038 ± 0.013 respectively. It should be noted, although significant, the differences in gait speed between conditions were less than 5%.

Discussion

This study was designed to determine if there are lower leg muscle activity differences between walking barefoot, in minimalist shoes or conventional footwear (CON and SH). The results illustrate that the first hypothesis is to be rejected as the degree of muscle activation differed between BF and MSH conditions.

Contrary to our second hypothesis, walking BF or in MSH was not observed to lead to increases in muscle activity during stance and in the TA and PL was seen to be lower than in conventional footwear. Furthermore the third hypothesis is also to be rejected as the OLD age group showed the least amount of differences across footwear conditions.

There was no increase in stance phase lower leg muscle activity when walking BF or in the MSH condition. The GCM, has been attributed a role in balance control during gait due to its ability to modulate the vertical displacement of the centre of mass (CoM) in relation to the centre of pressure (CoP) thus acting to prevent falling [12].

During the SS phase the body pivots over the ankle and approaches the LDS phase. The CoM trajectory follows an arc shape whereby the top of the arc is the point where the CoM is directly above the ankle and after this point it begins to lower due to the separation between the CoM-CoP and influence of gravity. The GCM's role is to increase its activity in order to maintain vertical support and prevent the CoM trajectory dropping too low by increasing the anterior progression of the CoP [13].

This has an indirect effect on step length and gait velocity [12]. In this study, when walking in the CON shoe there was a decrease in GCM activity compared to all other footwear conditions in the YOUNG; a decrease compared to the MSH and SH conditions in the MID but no difference across footwear in the OLD. The CON provides in-built support and greater overall anterior-posterior length due to a large sole size and therefore it was hypothesised that less emphasis would be placed on the GCM to control the CoM vertical displacement. This was only witnessed in the YOUNG with the MID showing no difference between the CON and BF conditions and the OLD showing no difference across all footwear conditions. A confounding factor which could partially explain these results could be the effect of walking speed. Consistent with previous findings [8], all ages walked slower BF however the amount of discrepancy grew with increasing age such that the difference in speed between BF and the CON condition in the OLD was over 3 times greater than it was in the YOUNG. Walking slower decreases the balance modulation role of the GCM therefore this may offset the increase in muscle activity due to the removal of supportive shoe structures potentially explaining the lack of difference witnessed in the OLD.

PL activity was reduced when participants walked BF compared with conventional footwear. As the PL plays a role in the maintenance of lateral stability around the foot during walking [14], our data suggest that we are prone to greater lateral instability during the initial loading phase when wearing conventional footwear. This could be a result of reduced foot position awareness. This was witnessed in the YOUNG and MID age groups however the OLD showed no differences across footwear. It is possible the small reduction in walking speed observed in the BF condition in the OLD group reduced the reliance on the PL activity but it could also hint at age-

related detriments in proprioceptive acuity. It's been previously shown older adults have an increased threshold to touch, pressure and vibration whilst joint position acuity is also negatively affected [15, 16]. This insensitivity could be prominent in the smaller intrinsic foot muscles and hence the increased afferent information available to the YOUNG and MID when minimally/unshod may not have the same benefit to the OLD. What is clear however is removing supportive footwear in the OLD age group does not worsen their lateral stability as implied by the lack of difference in PL activity. Proprioceptive acuity has however been shown to be receptive to improvements with training in elderly women [17]. As this was an acute exposure to walking barefoot it is not known whether consistent exposure to minimally/unshod conditions could promote proprioceptive improvements of the foot muscles leading to similar results to the younger age groups. Further study is required to investigate the activation patterns of these smaller muscles within the foot to explore this theory.

A decrease in TA activity during initial stance was observed when walking BF and in MSH compared to conventional footwear. Whilst the TA's primary role is to provide toe clearance during the swing-phase [18], it also assists in stability control during weight-acceptance by eccentrically contracting to lower the foot to the ground. This reduction in activity when minimally/unshod corresponds to the increase in plantar flexion at heel strike witnessed in these footwear conditions supporting previous research [8]. This change in foot position at initial contact, likely as a result of the decline in walking speed when minimally/unshod to reduce the forces associated with a heel strike contact, requires reduced input from the TA to control the load and dissipate the force [19].

It should be stated by maintaining shoe integrity and affixing markers to the shoe surface rather than through cut-outs, small discrepancies in marker position between

BF and shod may be present. This could explain a small part of the strike angle differences. Furthermore, the small (<5%) gait speed differences across footwear conditions may be linked with the reduced TA and PL muscle activity. It is possible the slower walking speed when BF or in MSH, may have occurred in order to minimise forces associated with a heel strike.

In summary, this study investigated the muscle activity differences between walking BF, walking in MSH and conventional footwear. MSH are intermediate in terms of ankle kinematics and muscle activation patterns. Walking BF and in MSH results in a decrease in TA activity at initial stance due to a flatter foot at contact. Walking BF also leads to a reduction in PL activity at initial stance in the young and middle age but not the old. Walking in supportive footwear leads to a reduction in GCM activation in the young and middle age but not the old, possibly as a result of slower walking speed when BF.

Conflict of Interest Statement

Conflicts of interest: none.

Acknowledgements: The research is funded by the Medical Research Council and Arthritis Research UK as part of the MRC-ARUK Centre for Musculoskeletal Ageing Research (Grant Reference Number: MR/ K00414X/1).

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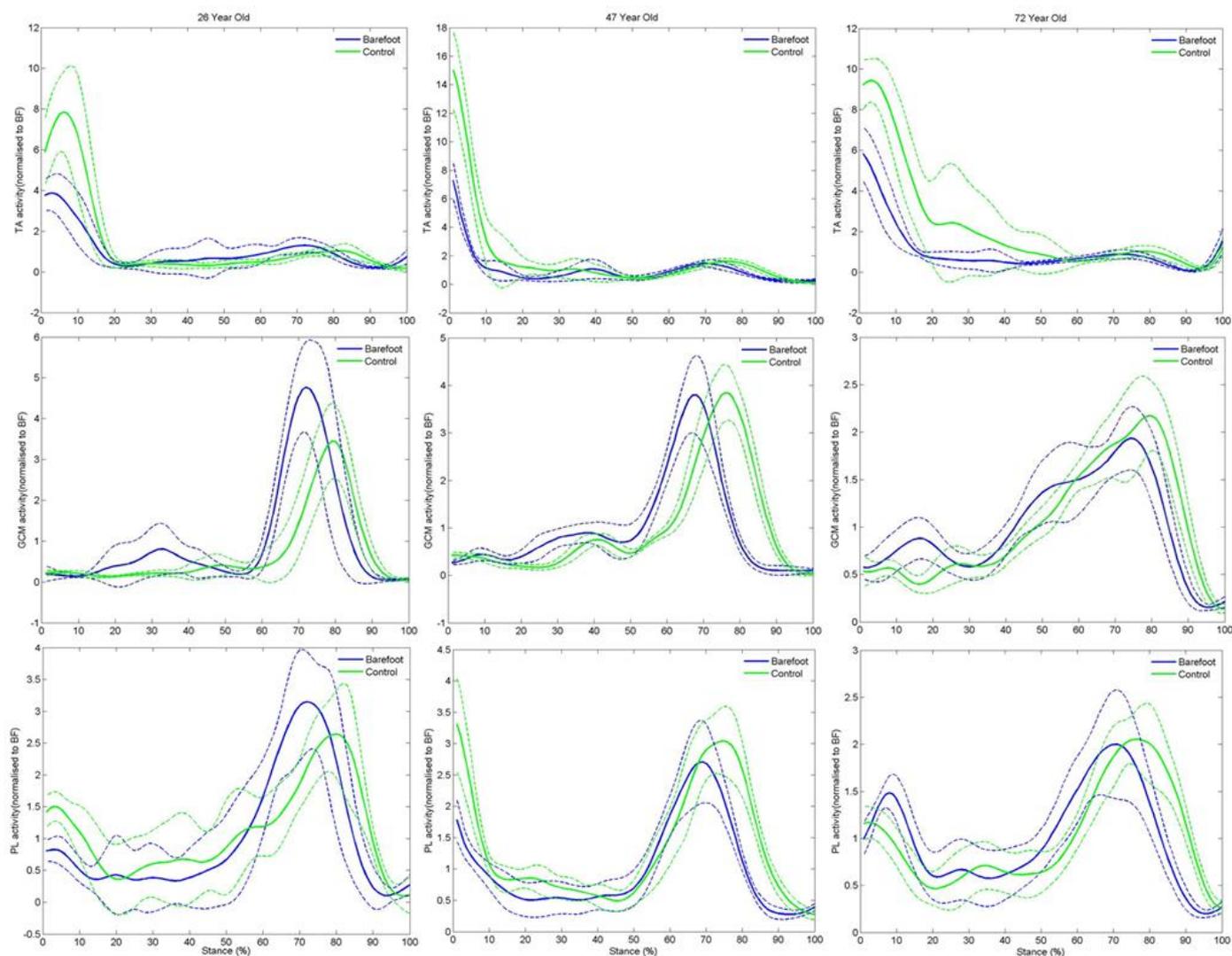
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Figures

Figure 1: **A)** The control shoe worn by males (Style Code: 50109, Hobos Mens). **B)** The control shoe worn by females (Style Code: 10001, Hobos Womens). **C)** and **D)** The unisex minimalist shoe worn by all participants (Product ID: 2169, Two Barefeet Boarding Co.).

Figure 2: Graphs to illustrate the average activity for the 3 muscles (tibialis anterior (TA): top, gastrocnemius medialis (GCM): middle, peroneus longus (PL): bottom) in the barefoot (blue lines) and control shoe (green lines) conditions after each cycle was normalised to the average activity of all cycles within the stance phase during the barefoot (BF) condition for each participant. Dotted lines indicate the standard deviation across all the cycles within each respective footwear condition. The left column of graphs is of a representative participant from the young age group (26years old), the middle column displays a representative participant from the middle age group (47years old) and the right column displays a representative participant from the old age group (72years old).





Age Group	No. In group	Age (years)	BMI (kg/m ²)	Own Shoe weight (g)
Young	20	27.85 (4.83)	23.25 (3.46)	293 (70.29)
Middle	30	54.85 (9.85)	25.04 (3.48)	319 (114.38)
Old	20	77.55 (4.39)	25.21 (4.39)	304.25 (71.6)

Table 1 – A summary of the participant's statistics. Data is displayed as means (S.D). BMI = Body Mass Index; S.D = Standard Deviation; g = grams