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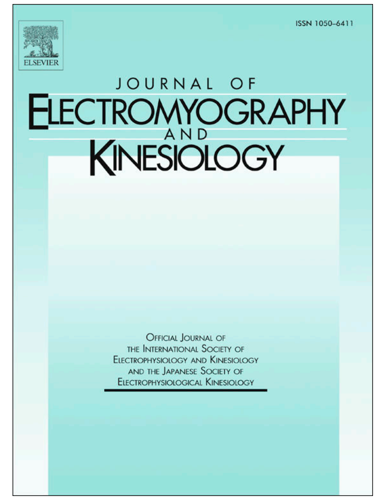

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Effect of Biomechanical Footwear on upper and lower leg muscle activity in comparison with knee brace and normal walking

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Effect of Biomechanical Footwear on upper and lower leg muscle activity in comparison with knee brace and normal walking

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Abstract

Aim: To evaluate the activity of knee stabilizing muscles while using custom-made biomechanical footwear (BF) and to compare it when walking barefoot and with a knee brace (Unloader®).

Methods: Seventeen healthy working-aged (mean age: 28 years; standard deviation: 8 years) individuals participated. The knee brace was worn on the right knee and BF in both legs. Surface electromyography (sEMG) data was recorded bilaterally from vastus medialis (VM), semitendinosus (ST), tibialis anterior (TA) and lateral gastrocnemius (LG) muscles during walking, and repeated-measures ANOVA with a post-hoc t-test was used to determine differences between the different walking modalities (barefoot, brace and BF).

Results: Averaged sEMG was significantly higher when walking with BF than barefoot or knee brace in ST and LG muscles. It was significantly lower when walking with the brace compared to barefoot in the right and left ST, LG and TA muscles. Analysis of the ensemble-averaged sEMG profiles showed earlier activation of TA muscles when walking with BF compared to other walking modalities.

Conclusion: BF produced greater activation in evaluated lower leg muscles compared to barefoot walking. Thus BF may have an exercise effect in rehabilitation and further studies about its effectiveness are warranted.

1. Introduction

Knee pain is a common reason for consulting the general practitioner. It occurs increasingly with advanced age and approximately 25% of the people aged over 55 years suffer from constant knee pain (Peat et al., 2001). Knee pain leads to avoidance of normal activities, which in turn cause atrophy in leg muscles and reduced walking speed (Noehren et al., 2018, White et al., 2013b). Knee osteoarthritis (KOA) and other musculoskeletal diseases are the most frequent causes of knee pain. Gait analysis studies suggest that individuals with KOA have impairments of gait parameters, for example reduced walking velocity, increased stance phase relative duration, decreased stride length, altered knee joint biomechanics and muscle activation (Astephen et al., 2008, Hortobágyi et al., 2005, Hubley-Kozey et al., 2006, Lynn et al., 2008, Mills et al., 2013, White et al., 2013a). Current clinical recommendations include exercise therapy as a first-line conservative management strategy for KOA (Fernandes et al., 2013, McAlindon et al., 2014). It is often difficult to motivate people to go to a gym and many people are not motivated to do home exercises either (Vuorenmaa et al., 2008). Thus conservative methods like biomechanical walking devices have received increasing interest, as one does not need to engage in separate exercise therapy, but can use the device in ordinary daily activities.

The use of knee braces has been recommended on the grounds that they have been shown to be effective in decreasing pain and improving physical function in KOA without severe adverse effects (Feehan et al., 2012, Moyer et al., 2015, Raja et al., 2011). Another promising treatment modality for KOA is a biomechanical footwear (BF). Such footwear is hypothesized to reduce pain and improve function by altering biomechanics during gait and thereby increasing loading and thus challenging and improving neuromuscular control (Bar-Ziv et al., 2010 and 2013, Drexler et al., 2012, Elbaz et al., 2014). Reinbach et al. (2020) found in controlled randomized study that biomechanical footwear resulted statistically significant improvement in pain at two-year follow-up, but of uncertain clinical importance. Further research is required to assess long-term efficacy and safety, before reaching conclusions about the clinical value of this device. The biomechanical footwear in previous studies were of different sizes and individually fitted for each participant. A novel adjustable biomechanical footwear can be attached underneath ordinary shoes with quick-release fasteners. It is produced only in two sizes and is easy to recycle among patients like knee orthoses and is less expensive than the biomechanical footwear reported above. Before starting a full-scale randomized controlled trial (RCT) in KOA, we evaluated if the new biomechanical footwear can produce greater leg muscle activity than walking without it.

A feasibility study was performed in healthy volunteers to test if the biomechanical footwear increased leg muscle activation and could be potentially useful in rehabilitation of knee pain. In this cross-sectional study lower limb muscle activity was measured with the surface electromyography (sEMG) during walking barefoot, with a knee brace or with biomechanical footwear.

2. Material & methods

2.1. Trial design

In this cross sectional comparative study a within-subject repeated-measurements design was used. Right and left sides were compared – the knee brace was only worn on the right side. Three trials were recorded per subject in each walking modality in the following order; barefoot, with knee brace, and with biomechanical footwear.

2.2. Subjects

The study participants were 17 healthy volunteers (4 males and 13 females). The participants were primarily physiotherapy and sport science undergraduate students from the local

Universities and employees of the local hospital. Healthy participants instead of patients with knee disease were enrolled in the study for the safety precaution to identify possible barriers and issues which could be related to using unstable biomechanical footwears. The inclusion criteria were age between 18 and 50, and a prospective participant was excluded if she or he had a prior neurological or orthopedic condition or knee-related symptoms as assessed using the Western Ontario and McMaster Universities Osteoarthritis Index (WOMAC) (Bellamy et al., 1988). The current study was within the scope of the RCT "Knee Brace and Biomechanical Footwear in the Treatment of Knee Osteoarthritis", which was approved by the Research Ethics Committee of the Central Finland Health Care District (ClinicalTrials.gov Identifier: NCT03684850). All participants provided written informed consent before participation.

2.3. Therapeutic devices

2.3.1. Biomechanical footwear

A new device called biomechanical footwear was developed in-house by the chief designer and first author (JY) of the present study and produced by local prosthesis manufacturer (Käpälämäen Proteesipalvelu Ltd., Jyväskylä, Finland). The biomechanical footwear has two adjustable straps and a shoe shaped platform with two semispherical elements under it; one under the heel and other under the ball of the foot (Fig. 1B and 2). The purpose of these elements is to render the biomechanical footwear unstable in inversion/eversion motions occurring in the frontal plane. To prevent excessive inversion/eversion, causing ankle strain, stoppers are located on the lateral and medial edge of the sole. The position of the posterior element is fixed, whereas the anterior element is adjustable and can be moved fore/aft i.e., towards the toe/heel according to the size of the foot. The center of the anterior element is situated approximately under the middle of the second metatarsal bone. It cannot be moved medial/laterally i.e., towards the first metatarsal/fifth metatarsal. The footwear was adjusted by the same experienced physiotherapist for all participants. Biomechanical footwear was worn under the subjects own regular running shoes. After fitting each therapeutic device, the participants were instructed to walk freely in the testing room until they felt comfortable using the device, which usually took a few minutes.

2.3.2. Knee brace

The biomechanical principle of the knee brace (Össur Unloader One®, Össur hf, Reykjavik, Iceland) is to unload the affected medial or lateral knee compartment by reducing the valgus or varus malalignment of the arthritic knee (Chew et al. 2007). In this study, the medial load reduction brace was worn on the right knee (Fig. 1A and 2). All participants were individually fitted with a brace by the same physiotherapist with several years of experience of fitting these braces.

2.4. Experimental procedure

2.4.1. Gait pattern

Participants were asked to walk at their usual, comfortable walking speed across the electronic GAITRite walkway (GAITRite® mat, CIR Systems Inc., Clifton NJ, USA). The GAITRite mat is a carpet 5.8 m long with an active sensor area of 427 cm long and 61 cm wide containing a total of 16128 pressure sensors. The walkaway is connected to a laptop computer with a USB cable. The sampling frequency of the system is 80 Hz. The walking speed was not standardized to avoid atypical walking patterns, which could influence muscle activation. Prior to the data collection, the participants practiced once by walking along the walkway. The temporo-spatial walking variables of gait velocity, step length and stance were measured with the software of GAITRite system. The gait velocity was obtained by dividing the distance traveled by the ambulation time (m/sec). The step length was from the heel center of the current footprint to the heel center of the previous footprint on the opposite foot (m). The stance phase is the weight bearing portion of each gait cycle and it is initiated by heel contact and ends with toe off of the same foot. Stance phase is presented as a percentage of the gait cycle time (% GC).

2.4.2. EMG recording

sEMG data were recorded with a portable measurement unit (A wireless biomonitor ME6000, Mega Electronics Ltd, Kuopio, Finland). The raw sEMG data were recorded at a sampling rate of 1000 Hz. The analog preamplifier in the sEMG measurement unit signal cables had a common mode rejection ratio of 110 dB and noise level of < 1.6 μV, and the signal was band-pass filtered at the measurement unit, with cutoff (half-power) frequencies at 8 and 500 Hz. Prior to electrode placement, the skin was shaved, cleaned with alcohol and treated with abrasive material. Disposable Ag/AgCl surface electrodes with a detection area of approximately 1.0 cm^2 each (BlueSensor M, Ambu A/S, Ballerup, Denmark) were placed in a bipolar arrangement, aligned parallel to the length of the muscle fibers over the mid-muscle belly, at an inter-electrode distance of 20 mm. The detecting electrodes were placed bilaterally on the vastus medialis (VM), semitendinosus (ST), tibialis anterior (TA) and lateral gastrocnemius (LG) muscles according to the guidelines of SENIAM (http://seniam.org). The reference electrode was placed beside the each pair of detective electrodes in accordance with instructions provided by the manufacturer (Mega Electronics Ltd) (Fig. 3).

In addition to sEMG recordings, all performances were recorded using a Canon MD235 Digital MiniDV Camcorder at 25 frames per second (Canon U.S.A., Inc. One Canon Plaza Lake Success, NY 11042). The camera was set up in the sagittal plane perpendicular to the walkway at a distance of 4.0 m and at a height of 0.8 meter on a tripod.

2.5. Data processing

The raw sEMG signal was synchronized with Mega software (Mega Electronics Ltd, Kuopio, Finland) to able analyze different phases of the walking cycle. The start and end point of the gait cycle was determined from the sEMG recording by identifying heel contact from the synchronized video. The EMG data was not synchronized with the data of GAITRite walkaway system. Only complete gait cycles with identifiable start and end heel contacts were used for analysis. The root mean square (RMS) value of the sEMG was used as a measure of average muscle activity. The RMS sEMG value was obtained for each muscle for barefoot, knee brace and footwear by calculating the sum

$$\text{RMS sEMG} = \sqrt{\frac{1}{N}\sum_{n=1}^{N} u_n^2}$$

over 6 to 12 complete recorded gait cycles depending on technical reasons and step length. The aim was to record as many gait cycles as possible. Here u_n is the measured sEMG voltage, N is the number of samples in the gait cycle(s).

These calculations were performed with MATLAB® using in-house code. Within-subject sEMG normalization was not needed because participants acted as their own controls and all measurements were performed in the same session, without altering electrode placement.

In addition to the RMS sEMG values, we also examined the temporal sEMG profiles for each muscle. One entire representative gait cycle with identifiable start of each heel contact was selected for each subject in each walking modality based on subjective visual inspection by the investigator. The entire period of the gait cycle was taken in account to avoid bias possibly caused by selecting only a certain phase. The raw sEMG curves were RMS-averaged with a 10 ms sliding window, and time-normalized to 0–100 % of the gait cycle. These temporal sEMG profiles were then averaged over all subjects to produce an ensemble-average temporal sEMG profile for each muscle in each walking modality. The 95 % confidence intervals (CIs) were calculated for the ensemble average profiles by assuming normal distribution of the individual sEMG profiles.

2.6. Statistical analysis

Repeated-measures analysis of variance (rANOVA) with a post-hoc Student's t-test was used to determine statistically significant differences in the RMS sEMG values between the different walking modalities in each muscle. The normality of the variables was evaluated using the Shapiro-Wilk test. A bootstrap-type F-test and post-hoc t-tests (10 000 replications) were used in case of violations of distribution assumptions (normality and/or sphericity). Bonferroni method was used to adjust the *p*-values for multiple comparisons. The results are reported as mean with standard deviation (SD) and with 95% CIs. Corrected p-values less than .05 were considered significant. In the temporal profiles interaction between the different walking modalities and gait cycle were analyzed using generalizing estimating equations (GEE) models with the exchangeable correlation structure. Generalized estimating equations were developed as an extension of the general linear model (eg. Ordinary Least Squares (OLS) regression analysis) to analyze longitudinal and other correlated data. All statistical analyses, except temporal profile interaction which was analyzed using Stata (15.1, StataCorp LP, College Station, TX, USA) were performed using R software.

3. Results

3.1. Descriptive

Participant mean age was 29 (SD 8; range 21–50) years; height 170 (7; 158–183) cm; body mass 66 (7; 52–79) kg; and body mass index 23 (2; 20–26) kg/m^2. According to the short-form of the International Physical Activity Questionnaire (IPAQ) (Craig et al. 2003), participants' physical activity varied from moderate (n = 3) to high (n = 14), i.e., mean 41 (SD 21) metabolic equivalent task hours (METh) per week, ranging from 6 to 80 METh/week. No statistically significant gender differences were observed in baseline variables, except that males were taller than females (179 SD 5 vs. 167 SD 5 cm, $p < 0.001$).

3.2. Gait pattern

The temporo-spatial gait variables differed significantly between the different walking modalities (Table 1). Habitual walking speed with biomechanical footwear was slower than walking with the knee brace, which in turn was slower than walking barefoot. Both feet showed significantly shorter step length when walking with biomechanical footwear than knee brace, and shorter step length when walking with the knee brace than barefoot. The proportion of the left and right stance phase to the whole gait cycle was smaller when walking with biomechanical footwear than knee brace or barefoot. When walking with the knee brace, the proportion of the stance phase to the gait cycle in the ipsilateral lower limb was greater than walking barefoot. However, no significant differences were observed between right and left step length or right and left proportional duration of stance phase when walking with the knee brace.

3.3. Muscle activity during gait

Differences in the RMS sEMG values between the different walking modalities are presented in Table 2. The rANOVA revealed that the mean values of the RMS sEMG differed significantly when walking with the biomechanical footwear compared to barefoot or knee brace in the right and left ST and LG muscles and in the left TA muscle. The same muscles showed significantly higher RMS sEMG values in the pairwise comparisons when walking with the biomechanical footwear than barefoot or with the knee brace, except that there was no difference in the RMS sEMG value in the left LG muscle when compared to the biomechanical footwear and barefoot. Compared to walking barefoot, the RMS sEMG value was lower when walking with the knee brace in the right ST and LG muscles and TA muscle in left leg (control leg).

The GEE analysis indicated from the ensemble-averaged temporal sEMG profiles that the timing of muscle activation differed in right TA muscle ($p < 0.001$) and left TA muscle when walking with different modalities ($p = 0.021$) (Table 3, Fig. 4). No differences were observed in the timing of other muscles during the gait cycle when walking with different modalities.

4. Discussion

4.1. Biomechanical footwear versus knee brace and barefoot

Based on the semispherical elements of the biomechanical footwear that promote a controlled perturbation during walking we hypothesized that the lower extremity muscle activity would be higher while walking with this than with knee brace or barefoot. The results showed that the activation measured by RMS sEMG was higher when walking with biomechanical footwear compared to walking barefoot or with the knee brace in the ST muscles, in the right LG muscle and in the left TA muscle. In addition, the left LG muscle exhibited higher RMS sEMG when walking with biomechanical footwear compared to walking with the knee brace. The ensemble average sEMG profile implies that there was increased activity in the ST muscles, with the biomechanical footwear during the stance phase, and in the LG muscles during terminal swing to mid stance.

It has been reported that varying the foot center of pressure (COP) by varying the positions of the semispherical elements in the sole of biomechanical device can induce significant changes in the lower limb sEMG activity during gait in KOA patients as well as healthy subjects (Goryachev et al., 2011a, 2011b). In their study Goryachev et al., (2011b) demonstrated significant changes in average sEMG activity in the LG muscles in healthy subjects during terminal stance and in the TA muscles during loading response and mid stance, when the COP of the foot was varied by adjusting the balance of biomechanical device (Goryachev et al., 2011b). Their device had specially designed sole that consists of two mounting rails enabling individual positioning of both elements under the forefoot and hindfoot regions. Our findings are similar to the results of Goryachev et al., (2011b), although our device has more limited options for semispherical element adjustments.

In our study the ensemble-averaged temporal sEMG profiles of walking barefoot resemble those in the literature for healthy subjects (Den Otter et al. 2004). The normal action of ST in gait is to decelerate knee extension in late swing, before heel strike, and to stabilize the knee together with the other hamstring muscles and knee extensors during loading response. In the present study, the RMS sEMG values in the ST muscles indicated greater muscle activity when walking with biomechanical footwer than barefoot or with the knee brace. The temporal sEMG graphs for the ST muscles seem to have increased sEMG activity when walking with biomechanical footwear, which may be attributed to increased demand for the muscles to stabilize the knee joint and maintain postural balance (Abulhasan and Grey 2017). The temporal sEMG profiles of the ST muscles seem to have an additional activity peak from terminal stance to initial swing when walking with biomechanical footwear, a result which may be associated with ST actively flexing the knee to assist the gastrocnemius in forward propulsion in toe-off. The possible differences of muscle activity findings in the phases of gait cycle from the temporal sEMG graphs should be interpreted with caution, because they are based on visual observation of the data instead of statistical testing. Nevertheless, a similar additional sEMG activity peak in ST has been reported previously in slower than normal walking speeds with regular footwear (Den Otter et al., 2004). The authors associated the additional activity peak with toe-off in the gait cycle. In the present study, only the heel strike, and not the toe-off, was marked to the sEMG data, but the timing of the additional activity peak in the ST muscles roughly corresponds to the period where toe-off occurs in normal gait.

In the LG muscles, the RMS sEMG values were higher when walking with biomechanical footwear than with the brace or barefoot in the right LG. In normal gait, the main function of the gastrocnemius muscle is to plantar flex the ankle during terminal stance to produce the forward propulsion of walking. A secondary function is to resist ankle dorsiflexion and knee extension during mid-stance, as the tibia/fibula moves anteriorly (Abulhasan and Grey 2017). When walking with biomechanical footwear, an additional period of LG muscle activity was observed, yet not statistically verified, in the temporal sEMG profiles from terminal swing to mid-stance, peaking at heel contact, compared to barefoot or the brace. This might result from increased activation of the LG to provide additional stability to the knee during heel strike and weight bearing or from alterations in joint moments by sagittal COP shift (Haim et al., 2008 and 2010). It has been reported previously that in healthy subjects the activation of gastrocnemius is increased when the COP of the foot is shifted anteriorly (Goryachev et al. 2011). However, this increased activation was observed in this study during terminal stance and pre-swing.

In the TA, the RMS sEMG value was greater when walking with biomechanical footwear than barefoot or with the brace, but the difference was only statistically significant on the left. Because the biomechanical footwear was worn on both feet, there is no apparent physiological reason for this and it may be purely technical or chance. A closer examination to the mean differences and their confidence intervals in the pairwise comparisons between footwear and barefoot as well as footwear and brace, however, reveals that the differences were of borderline significance, and with a larger number of participants it is possible that they could be statistically significant. The normal function of the TA is to resist ankle plantar flexion during heel contact, in order to prevent foot drop, and to dorsiflex the ankle during the initiation of the swing to pull the foot off the ground. TA muscles were activated in both legs earlier in the pre-swing when walking with biomechanical footwear compared to walking barefoot or with the brace. This may be due to the TA initiating ankle dorsiflexion earlier, to prevent the front of the footwear from dragging on the walking surface. In addition, the added mass of footwear below the ankle may contribute to earlier activation of the TA compared with the added mass of knee brace which is entirely above the ankle.

4.2. Methodological limitations and strengths

The differences in RMS sEMG values between the footwear and other modalities were not always clearly reflected in the temporal sEMG profiles. We attribute this to the fact that the temporal sEMG profiles were constructed from one gait cycle from each test subject, whereas the RMS sEMG values that were used in the statistical analyses were calculated from 6 to 12 gait cycles from all 17 subjects. This means that for the sEMG signal amplitude, the statistical power of the comparisons between the RMS sEMG values is greater than that of the temporal sEMG profiles. However, the number of samples/subjects used in the ensemble-averaged temporal profiles is sufficient to examine the differences in timing of muscle activity between walking with the footwear, with the brace and barefoot. The small sample size was considered in the selection of the statistical methods to be applied, and bootstrap sampling was used where necessary. However, the small sample size and the fact that age, gender or anthropometric measures of the participants were not considered in this investigation limits generalizing the results into the general population.

We hypothesized that the lower extremity muscle activity would be higher while walking with the biomechanical footwear than walking with knee brace or barefoot. The results of our study suggest that the hypothesis is confirmed. This information may be useful in planning and implementing therapeutic exercise interventions in patients with KOA. The next step is to investigate the effect of biomechanical footwear in the clinic with patients suffering from symptoms of KOA in a randomized study.

5. Conclusions

This study found that biomechanical footwear increases the neuromuscular activation of certain lower limb muscles around the knee joint during walking, whereas the knee brace decreased muscular activation. Based on these observations, we conclude that the use of biomechanical footwear increases muscle activation and influences the timing of muscle activation during walking and thus may improve neuromuscular control in healthy subjects. Biomechanical footwear may be useful in rehabilitation of knee pain and warrants further investigation.

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Declaration of competing interest

The authors have no conflicts of interest to declare.

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Declaration of Competing Interest

The authors declare that they have no competing interest or conflicts of interest. None of the authors have any financial or personal relationships that could inappropriately influence this work.

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Table 1. Temporo-spatial gait variables during walking barefoot, with biomechanical footwear (BF) or knee brace worn on the right knee.

Variable	Mean (SD)					Mean diff (95 % CI)		
	Barefoot	BF	Brace	*F*	*p*	BF vs barefoot	Brace vs barefoot	BF vs brace
Velocity (m/s)	1.70 (0.21)	1.47 (0.19)	1.65 (0.20)	43.1	< 0.001	-0.23 (-0.31, -0.15)*	-0.05 (-0.10, 0.00)*	-0.18 (-0.25, - 0.01)*
Step length R (m)	0.76 (0.059)	0.73 (0.060)	0.75 (0.054)	6.30	0.016	-0.03 (-0.07, -0.01)*	-0.01 [-0.03. 0.01]	-0.02 (-0.05, 0.00)
Step length L (m)	0.76 (0.056)	0.73 (0.057)	0.74 (0.054)	8.43	0.001	-0.04 (-0.06, -0.01)*	-0.02 (-0.04, -0.01)*	-0.01 (-0.04, 0.01)
Stance R (% GC)	60.6 (1.57)	58.3 (1.97)	61.0 (1.89)	67.0	< 0.001	-2.30 (-3.13, -1.44)*	0.40 (-1.16, 0.84)	-2.63 (-3.21, -2.04)*
Stance L (% GC)	60.7 (1.57)	58.2 (1.78)	60.9 (1.24)	52.0	< 0.001	-2.42 (-3.46, -1.50)*	0.20 (-0.28, 0.70)	-2.64 (-3.46, -1.84)*

Notes: SD = standard deviation; CI = confidence interval; R = right; L = left; % GC = percent of gait cycle; p = Bonferroni corrected p-value; $p < 0.05$ was considered statistically significant, *$p < 0.05$.

Table 2. Root mean square (RMS) sEMG in different lower limb muscles during walking barefoot, with biomechanical footwear (BF) or knee brace worn on the right knee.

Muscle	Mean (SD) [µV]					Mean diff (95 % CI[b]) [µV]		
	Barefoot	BF	Brace	*F*	*p*	BF vs barefoot	Brace vs barefoot	BF vs brace
VM R	62.8 (28.8)	67.1 (42.0)	59.0 (19.7)	0.503	0.627[b]	4.32 (-3.37, 14.7)	-3.76 (-21.0, 8.29)	8.08 (-8.94, 34.1)
VM L	59.3 (21.1)	54.9 (17.0)	57.4 (21.5)	2.40	0.107	-4.39 (-8.47, -0.744)	-1.88 (-5.61, 1.62)	-2.50 (-7.68, 1.81)
ST R	82.0 (32.5)	103 (36)	68.7 (30.7)	40.7	< 0.001[b]	20.8 (13.9, 27.4)***	-13.3 (-19.7, -7.51)**	34.1 (23.9, 43.6)***
ST L	85.1 (45.5)	108 (52)	79.1 (36.9)	16.2	< 0.001	22.5 (10.4, 35.0)**	-5.95 (-14.3 1.75)	28.5 (17.7, 40.1)**
TA R	141 (43)	155 (48)	140 (32)	3.08	0.060	14.2 (-0.06, 27.5)	-1.21 (-13.1, 9.48)	15.4 (-0.86, 33.3)
TA L	148 (49)	169 (57)	135 (40)	12.3	< 0.001[b]	21.0 (4.83, 36.6)*	-13.7 (-22.7, -6.17)*	34.7 (18.2, 54.0)**
LG R	104 (39)	137 (66)	85.8 (39.7)	26.6	< 0.001[b]	32.7 (16.7, 51.3)*	-18.2 (-24.6, -12.2)***	50.9 (36.0, 69.2)***
LG L	111 (54)	125 (48)	103 (43)	9.82	< 0.001	14.4 (2.35, 25.5)	-7.76 (-16.7, -0.03)	22.2 (11.2, 33.5)**

Notes: SD = standard deviation; µV = microvolt; VM = vastus medialis; ST = semitendinosus; TA = tibialis anterior; LG = lateral gastrocnemius; L = left; R = right. b = from bootstrapped F-distribution; CI = Bonferroni corrected confidence interval from bootstrapped distribution; *$p < .05$; **$p < .01$; ***$p < .001$; $p < 0.05$ was considered statistically significant; p-values of mean differences were Bonferroni corrected for multiple comparisons; 10 000 bootstrap samples were used.

Table 3. Interaction between different walking modalities (barefoot, knee brace and biomechanical footwear) and gait cycle.

Muscle	P-value
Vastus medialis, right	0.65
Vastus medialis, left	0.91
Semitendinosus, right	0.089
Semitendinosus, left	0.25
Tibialis anterior, right	<0.001
Tibialis anterior, left	0.021
Lateral gastrocnemius, right	0.051
Lateral gastrocnemius, left	0.15

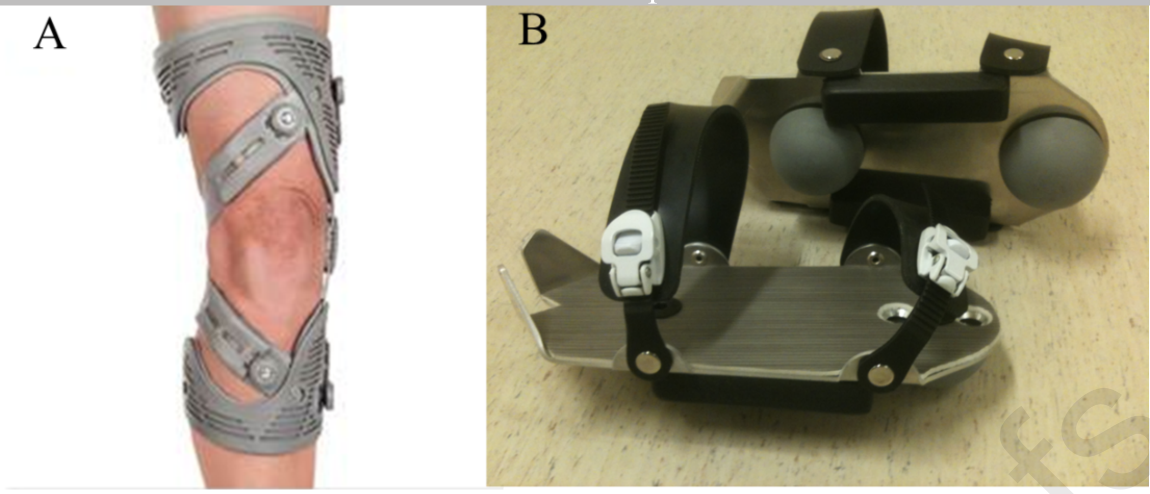

Fig. 1. Unloader One® knee brace (A) and biomechanical footwear device (B) used in this study.

Fig. 2. Biomechanical footwear (A) and Unloader® knee brace (B) shown in the experimental condition on the GAITRite® mat.

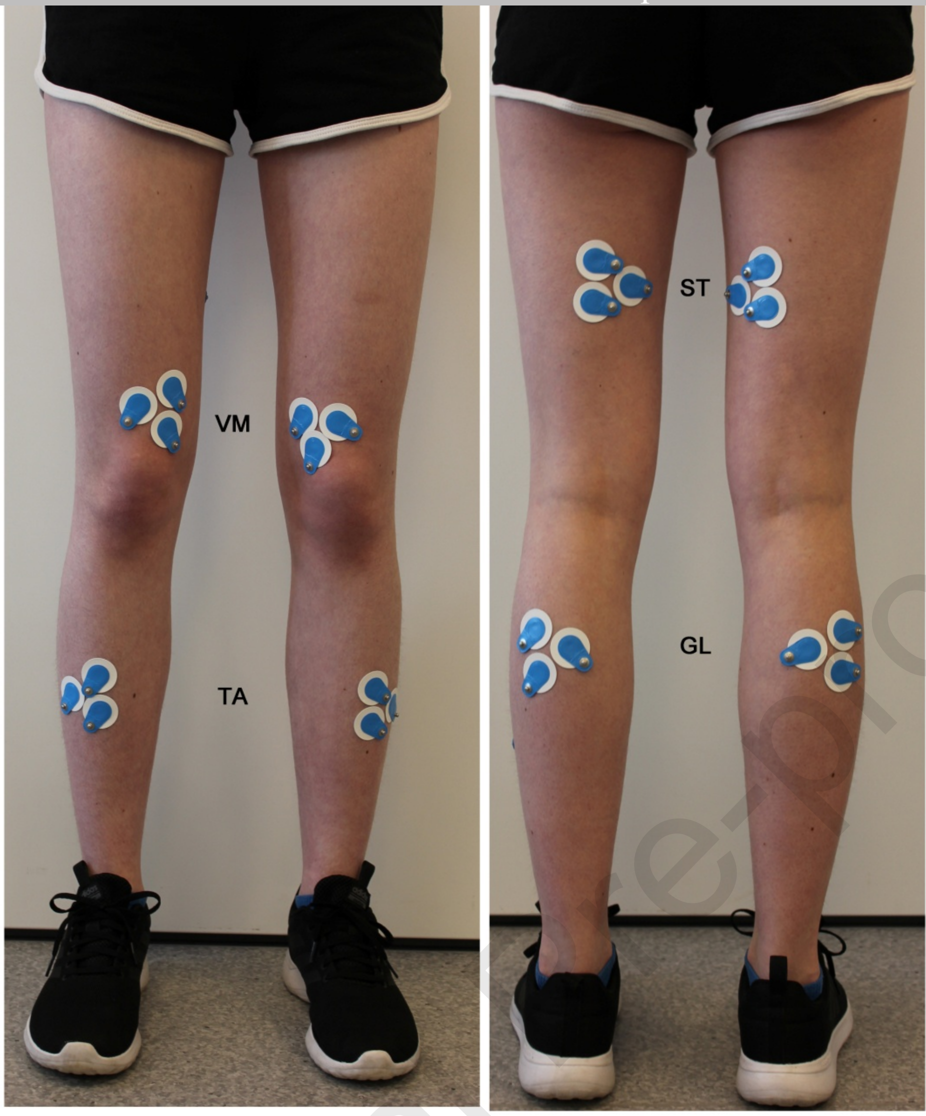

Fig. 3. Surface electromyography was evaluated on the following muscles bilaterally: VM = vastus medialis, ST = semitendinosus, TA = tibialis anterior, LG = lateral gastrocnemius.

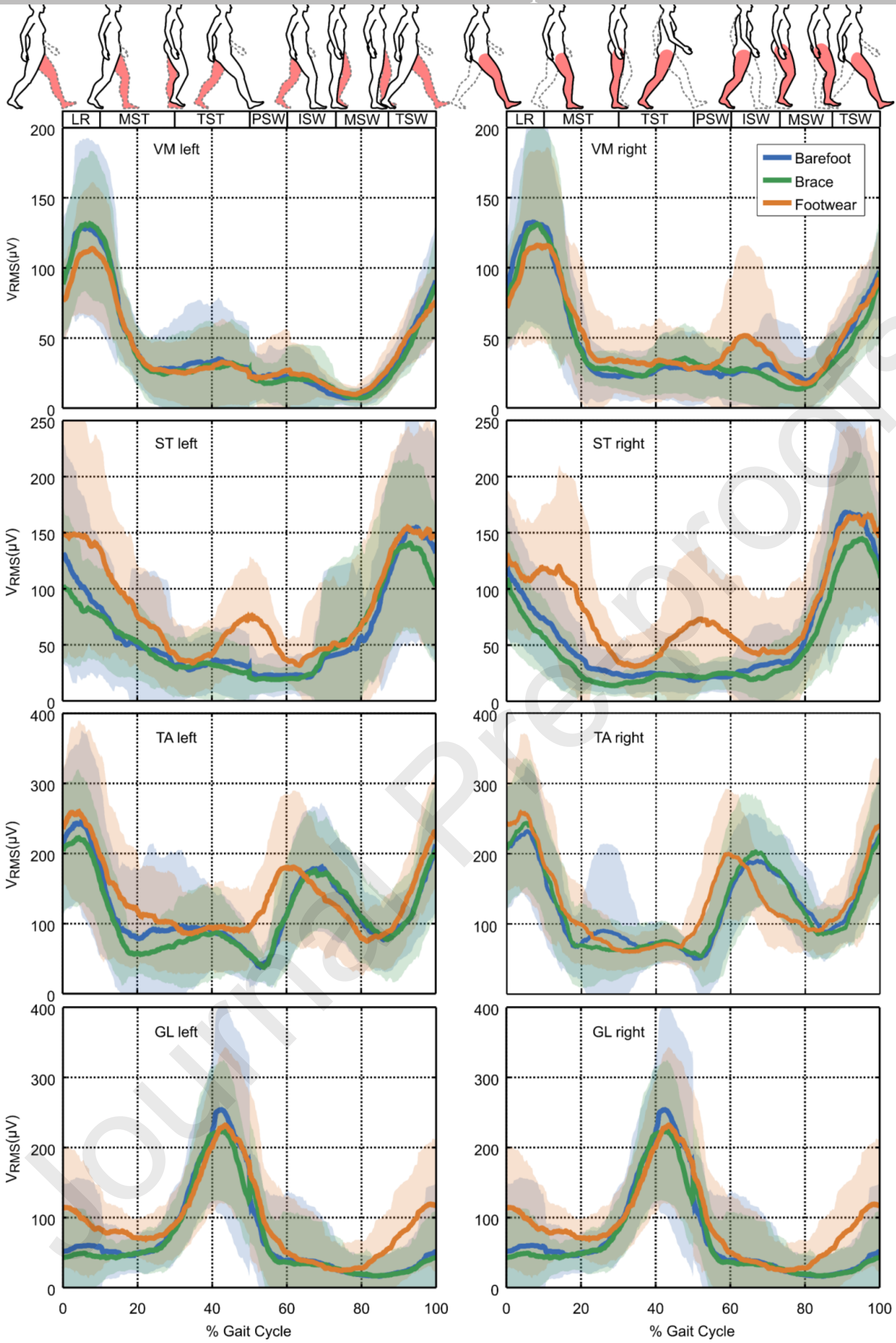

Fig. 4. Ensemble average sEMG profiles during walking with biomechanical footwear, unloader knee brace and barefooted. The lines represent the mean and the shadowed areas are the 95% confidence intervals. LR = loading response, MST = mid stance, TST = terminal stance, PSW = pre-swing, ISW = initial swing, MSW = mid swing, TSW = terminal swing; VM = vastus medialis; ST = semitendinosus; TA = tibialis anterior; LG = lateral gastrocnemius; V_{RMS} = root mean square averaged (RMS) sEMG voltage, with 10 ms sliding window.