Journal of Electromyography and Kinesiology 21 (2011) 333–339

Contents lists available at ScienceDirect

Journal of Electromyography and Kinesiology

journal homepage: www.elsevier.com/locate/jelekin

The effect of manipulation of the center of pressure of the foot during gait on the activation patterns of the lower limb musculature

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ARTICLE INFO

Article history:
Received 10 August 2010
Received in revised form 25 November 2010
Accepted 29 November 2010

Keywords:
Electromyogram
Gait
Center of pressure
Lower limb

ABSTRACT

Background: Therapeutic devices that manipulate the center of pressure (COP) of the foot can induce kinetic and kinematic changes in gait. Appropriate changes in joint moments and muscle activation during gait have been proven to be beneficial for patients with neuromuscular and orthopedic disorders. The purpose of this study was to investigate the effect of different COP positions during gait on the activity of the lower limb musculature of healthy subjects.
Methods: A novel foot-worn biomechanical device that allows controlled manipulation of the COP during gait was used. Twelve healthy males underwent EMG analyses of the key muscles of the leg while wearing the device. The trials were carried out at six COP positions relative to neutral configuration: anterior, posterior, medial, lateral, dorsi flexion and plantar flexion.
Results: The EMG activity of the lateral gastrocnemius varied significantly with COP during terminal stance ($p = 0.023$) and preswing ($p = 0.020$), the tibialis anterior during load response ($p = 0.019$) and mid-stance ($p = 0.004$), the biceps femoris during terminal stance ($p = 0.009$) and the vastus lateralis during initial contact ($p = 0.010$).
Conclusion: There are significant changes in the muscle activity of the lower limb in response to manipulation of the COP of the foot during gait.

1. Introduction

Pathologies such as patellofemoral pain syndrome (PFPS) and osteoarthritis (OA) are common musculoskeletal conditions (Dixit et al., 2007; Hogenmiller and Lozada, 2006). Therapists have turned to new devices that can manipulate a patient's foot center of pressure (COP) in order to decrease pain and improve function. The most prominent of these have been footwear-derived biomechanical devices. Investigators have defined the effects of some of these COP manipulations on kinetic patterns. Wedged insoles, for example, have been suggested to shift the location of the COP in the coronal plane, thereby altering resulting torques from the foot proximally (Kakihana et al., 2005; Maly et al., 2002; Xu et al., 1999). Application of wedge insoles has been reported to decrease the load and the magnitude of the adduction moment at the medial compartment of the knee joint in healthy and arthritic subjects (Kakihana et al., 2005; Crenshaw et al., 2000; Ogata et al., 1997; Yasuda and Sasaki, 1987). In two previous studies we analyzed the kinetic outcomes of a novel biomechanical apparatus that allows for controlled manipulation of the COP during gait. Adjusting the COP in the coronal plane (i.e., from medial to lateral) correlated with significant changes in the knee adduction moment during stance (Haim et al., 2008). Likewise, manipulation of the COP in the sagittal plane (i.e., from posterior to anterior) significantly related with ankle dorsiflexion torque and knee extension torque during stance (Haim et al., 2010).

The current study is an extension of the studies reported in (Haim et al., 2008, 2010). In the current study we analyze the electromyography (EMG) data that was collected during the study described above (Haim et al., 2008, 2010), that is the kinematic, kinetic and electromyography data were collected simultaneously. Moreover, for consistency with the results reported in our previous works, we used the same methods at the same timeframes during the gait cycle so that the kinematic, kinetic and EMG phenomena could be correlated.

There have been several studies on the effects of COP changes on lower limb musculature activation. In a study by Mulavara et al. (1994), patients leaning forward moved their COP anteriorly and increased activation of their gastrocnemius (GC) muscle. Patients leaning backward moved their COP posteriorly and increased activation of their tibialis anterior (TA) muscle. In a study by Krishnamoorthy et al. (2004), patients who were asked to release a load from extended arms showed a forward shift in the COP.

To date, investigators have not examined the activation of the lower limb muscles in response to precisely controlled manipulation of the COP in multiple planes. This can be done using the novel

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doi:10.1016/j.jelekin.2010.11.009

biomechanical device we applied in previous studies (Haim et al., 2008, 2010). Moreover, the biomechanical device manipulation is patient specific and the COP can be maintained throughout gait; thus, the changes in muscle activation can be measured dynamically. The purpose of this work is to determine the specific changes in activation of key muscles of the lower limb in response to manipulation of the COP in the coronal and sagittal planes in healthy individuals.

We hypothesized that coronal shifts of the COP would correspond with activation changes of coronal muscles in a way that compensates for the perturbation (i.e., a medial COP shift will cause an increase in the activation of laterally positioned muscles) and that a sagittal COP shift will correspond with activation changes of sagittal muscles, in order to maintain force and torque equilibrium within the lower kinematic chains.

2. Methods

2.1. Participants

The study population was comprised of 12 healthy male volunteers with equivalent shoe size (French 43) and a similar anthropometric profile (i.e., weight, height and dominant leg). The patients' mean ± SD age was 25.95 ± 2.48 years, height was 177.35 ± 3.52 cm and weight was 74.04 ± 4.12 kg. Exclusion criteria were any orthopedic musculoskeletal or neurological pathology or any orthopedic injury or trauma in the 12 months preceding the study. Approval of the Ethics Sub-Committee was obtained and all participants gave informed consent.

2.2. Biomechanical system

A biomechanical device (APOS System, APOS–Medical and Sports Technologies Ltd., Herzliya, Israel) that allows controlled manipulation of the COP (Haim et al., 2008, 2010) was used in the study (Fig. 1). The device consists of two convex-shaped biomechanical elements attached to each foot. One is located under the hindfoot region and the other is located under the forefoot region. The elements are attached to the foot using a platform in the form of a shoe. The platform is equipped with a specially designed sole, consisting of two mounting rails. These enable flexible and continuous positioning of each element in multiple planes under each region. A shift of the elements in coronal and sagittal planes shifts the COP in the coronal and sagittal planes, respectively (Haim et al., 2008, 2010). The system was generously donated by the manufacturer prior to the study. A pilot study was conducted to assess the stability of the apparatus. It determined that, for healthy adults, satisfactory walking stability can be kept within the range of 1.8 cm posterior and 1.8 cm anterior, 1.5 cm medial and 2 cm lateral deviation of the biomechanical elements from the neutral sagittal axis.

2.3. Experimental protocol

Prior to the study, one physician performed a functional assessment of all subjects. A qualified physiotherapist calibrated the biomechanical device. First, positioning of the elements for the "functional neutral configuration" (FNC) was determined and documented. The FNC was defined as the position in which the apparatus exerted the least valgus, varus, dorsal or plantar torque about the ankle in the individual being examined. The FNC was determined from observation during the subject's walk and from subject feedback regarding comfort of use while walking. The neutral configuration was used as a baseline to calibrate the other positions for the study, i.e., the other positions were determined with respect to it so that minor errors in determining the FNC were consistent throughout the measurements and therefore could be ignored. The configurations checked during the experiment were anterior, posterior, medial, lateral, dorsi flexion and plantar flexion. Anterior and posterior configurations were defined as 1.5 cm anterior and 1.5 cm posterior deviation of the biomechanical elements along the neutral sagittal axis. Medial and lateral configurations were defined as 0.8 cm medial and 1.5 cm lateral deviation of the biomechanical elements from the neutral sagittal axis. Dorsi flexion and plantar flexion were defined as 3 cm elevation of the front and back of the shoe platform, respectively (Fig. 2). In order to become accustomed to the effect of the shoe and the experimental procedure, subjects were asked to walk at a self-selected velocity

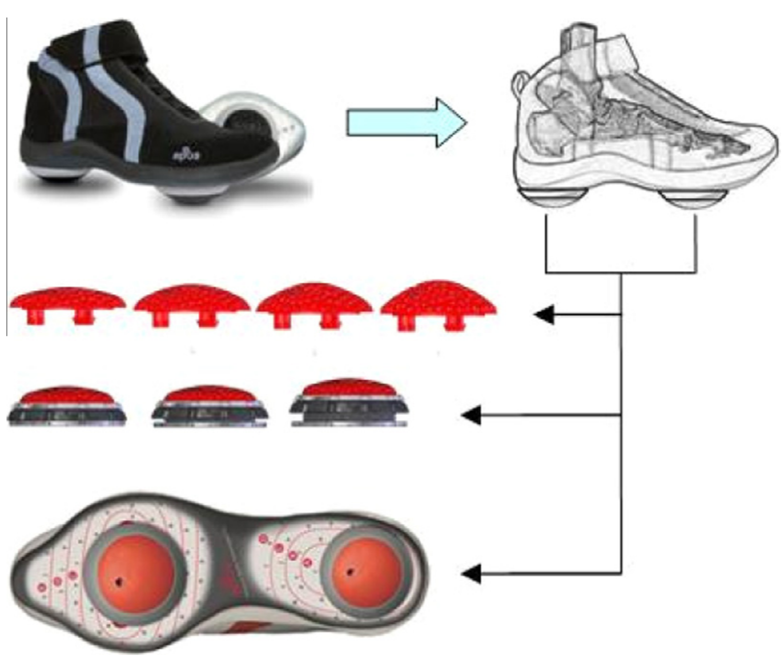

Fig. 1. Biomechanical system consisting of a foot-worn platform with mounting rails and moveable elements with height-adjustable bases.

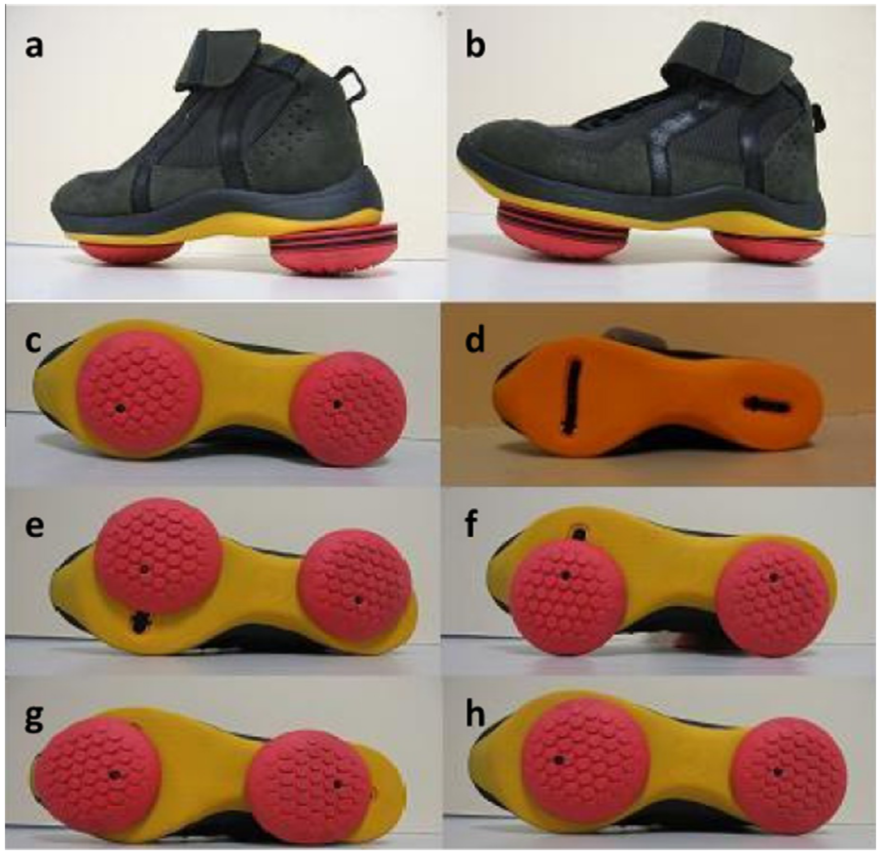

Fig. 2. Calibration of the elements on the base of the biomechanical device: (a) Plantar flexion configuration (hindfoot element elevated); (b) dorsal flexion configuration (forefoot element elevated); (c) neutral sagittal configuration; (d) elements removed; (e) lateral sagittal axis configuration; (f) medial sagittal axis configuration; (g) anterior configuration; (h) posterior configuration. This study presents muscle activity in lateral, medial, anterior, posterior, dorsi and plantar flex configurations only.

for several minutes prior to data collection. During this "training" period, a metronome was tuned to correspond to the subject's self-selected velocity. The metronome was then used throughout the data collection to ensure consistent cadence.

The biomechanical device was calibrated only once for each position. Subjects were asked to walk over a 10-m walkway in their pre self-selected speed. Six trials of each condition were collected per subject for averaging and consistency. All conditions were tested in random order on the same day.

2.4. Data acquisition and processing

A Surface EMG ZeroWire system (Aurion Ltd., Italy) was used to measure the activity of all the following muscles: lateral gastrocnemius (LG), medial gastrocnemius (MG), vastus lateralis (VL), vastus medialis (VM), tibialis anterior (TA), semitendinosus (ST) and biceps femoris (BF). The system includes a set of wireless surface EMGs that can be attached comfortably to the participant's skin. After cleaning the skin with isopropyl alcohol, one EMG recording electrode (10–1000 Hz, 16 bit resolution on all measurements) was fixed properly to each muscle by an experienced physician according to the "surface EMG for non-invasive assessment of muscles" (SENIAM) recommendations for surface EMG placement (Hermens et al., 1999). EMG data were recorded simultaneously for all muscles continuously throughout the six trials for each COP position. The kinematic data was recorded by an eight camera Vicon motion tracking system and the ground reaction forces were recorded by two 3-dimensional AMTI OR6-7-1000 force plates. Similar to Croce (1986) and Haim et al. (2008, 2010) all analyses were performed for the dominant leg, which was the right leg for all study participants.

All data were acquired simultaneously by Vicon Nexus software (Oxford Metrics Ltd., Oxford UK) and exported to MATLAB™ software for analysis. The EMG signals were sampled at 120 Hz. The EMG signal envelope was calculated. The EMG envelope is usually calculated according to a standard protocol (Langzam et al., 2006; Nigg et al., 2005). Data were analyzed in a similar fashion to the protocol of Langzam et al. First, a high pass filter (HPF) of 10 Hz (Butterworth, 4th order) was carried out to remove motion artifacts. The signal was then rectified and the peak envelope was extracted. A low pass filter (LPF) of 15 Hz (Butterworth, 4th order) was then carried out.

Each step of the gait cycle was divided into stance and swing, as measured by the force plates. All calculations were performed for stance only. The time scale of the stance was normalized from 0% to 100%. The stance period was divided into several phases with respect to time: initial contact (IC; 0–2%), load response (LR; 0–10%), midstance (MS; 10–30%), terminal stance (TS; 30–50%), preswing (PS; 50–60%) and terminal contact (TC; 60–100%) (Haim et al., 2008, 2010).

Using a protocol similar to Edwards et al. (2008), an average rectified value (ARV) for the EMG of each muscle during each phase of the stance period was obtained by calculating the integral of the graph pertaining to a specific phase. As suggested by Edwards et al. (2008), within subject normalization was not needed for the data because participants acted as their own control and all procedures were performed in the same session, without the electrode positions being altered. For all stance phases, ARVs of six trials under each condition were averaged for each subject to get a representative value.

Table 1
Chi-square and significance values for lower limb muscles over the 6 COP positions in each phase of stance.

	Initial contact (IC)	Load response (LR)	Midstance (MS)	Terminal stance (TS)	Preswing (PS)	Terminal contact (TC)
ST Chi-square	0.600	0.720	4.200	7.320	2.520	4.320
p value	0.896	0.868	0.241	0.062	0.472	0.229
BF Chi-square	3.960	1.582	6.164	11.509	6.600	2.236
p value	0.266	0.664	0.104	0.009*	0.086	0.525
VM Chi-square	3.866	6.114	5.657	10.514	4.686	8.4
p value	0.569	0.295	0.341	0.062	0.455	0.136
VL Chi-square	15.191	10.743	8.971	9.314	9.657	7.543
p value	0.010*	0.057	0.110	0.097	0.086	0.183
LG Chi-square	6.720	2.514	6.229	13.086	13.429	1.657
p value	0.242	0.774	0.285	0.023*	0.020*	0.894
MG Chi-square	6.175	7.779	5.753	7.156	1.338	5.130
p value	0.29	0.169	0.331	0.209	0.931	0.400
TA Chi-square	4.600	9.960	13.080	4.320	3.240	0.600
p value	0.204	0.019*	0.004*	0.229	0.356	0.896

Chi-square values were calculated using Friedman's test.
The LG varied significantly with COP position in TS and PS phases of stance. The TA varied significantly in LR and MS. The BF varied significantly in TS. The VL varied significantly in IC.
* $p \leqslant 0.05$ was considered statistically significant.

Table 2
Mean ± SD ARV values for LG over 6 COP positions in Terminal Stance and Preswing.

	Anterior ARV	Posterior ARV	Dorsi flex ARV	Plantar flex ARV	Lateral ARV	Medial ARV
Lateral gastrocnemius Terminal stance	0.503 ± 0.331	0.465 ± 0.281	0.505 ± 0.335	0.399 ± 0.250	0.418 ± 0.288	0.554 ± 0.361
Lateral gastrocnemius Preswing	0.659 ± 0.468	0.583 ± 0.417	0.686 ± 0.395	0.551 ± 0.396	0.563 ± 0.351	0.795 ± 0.569

Mean ± SD ARV values are in mV.

Table 3
Mean ± SD ARV values for TA over 4 COP positions in Load Response and Midstance.

	Anterior ARV	Posterior ARV	Dorsi flex ARV	Plantar flex ARV
Tibialis anterior Load response	3.052 ± 1.042	3.779 ± 1.371	3.588 ± 1.816	3.812 ± 1.459
Tibialis anterior Midstance	1.651 ± 0.901	2.115 ± 1.016	1.747 ± 1.026	2.452 ± 1.338

Mean ± SD ARV values are in mV.

Then parameter values of different COP conditions were compared for every phase of stance.

2.5. Statistical analysis

Friedman tests were used to compare the EMG activity of MG, LG, VM and VL muscles across lateral, medial, anterior, posterior, dorsi flexion and plantar flexion positions and ST, BF, TA muscles across anterior, posterior, dorsi flexion and plantar flexion positions in each specific phase of stance. A *p* value of less than 0.05 was considered as statistically significant. All analyses were performed using SPSS (version 17.0).

3. Results[1]

The primary muscles which average rectified activity varied with COP were the distal muscles – LG, MG and TA. The LG and TA varied significantly with COP, the MG did not. LG showed significant change with COP in the TS ($X^2 = 13.086$, $p = 0.023$) and PS ($X^2 = 13.429$, $p = 0.020$) phases of stance, the TA varied significantly with COP in the LR ($X^2 = 9.960$, $p = 0.019$) and MS ($X^2 = 13.080$, $p = 0.004$) phases of stance. The proximal muscles showed similar tendencies to the distal ones but the influence of the COP on the muscle activity was not as evident. The BF varied significantly with COP during TS ($X^2 = 11.509$, $p = 0.009$), and the VL varied significantly with COP during IC ($X^2 = 15.191$, $p = 0.010$). The activation changes in all the other muscles in all phases of stance did not differ significantly with COP variation. The results of statistical analysis are presented in Table 1; ARVs of the LG and TA in the stance phases when significant changes were found are presented in Tables 2 and 3.

The LG and TA showed unique changes in specific COP positions. The LG showed significantly more activation in the medial COP position. More specifically, in the PS phase, the average rectified EMG activity of the LG showed a 41% increase (from 563 to 795 μV) in activity when the COP was shifted from the lateral to the medial position. Fig. 3a illustrates the ARV of the LG in the different COP positions during the PS phase of stance in box plots. The activity of the muscle was significantly lower in the lateral, posterior and plantar flex positions in comparison to its activity in the medial, anterior and dorsi flex positions, respectively, in TS and PS phases

[1] Gait phases and events reported here correspond to the events reported in Haim et al. (2008, 2010).

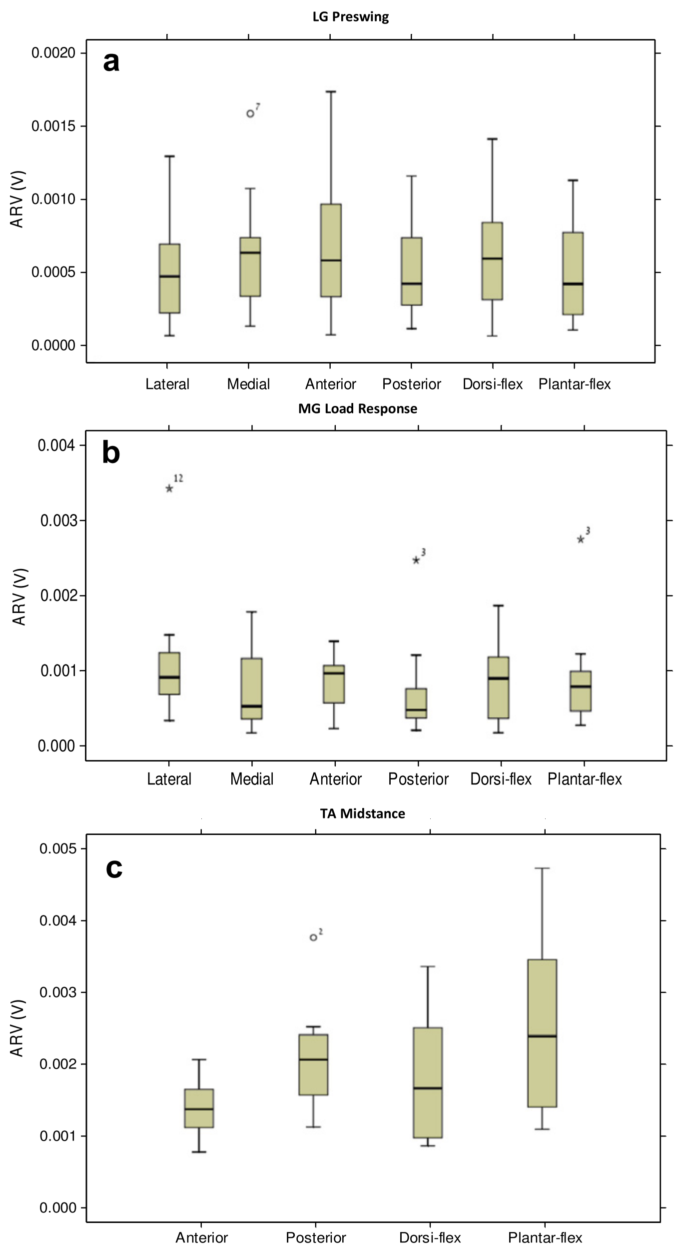

Fig. 3. Average rectified values (ARV) of the EMG activity of lower limb muscles in specific phases of gait in response to center of pressure (COP) manipulations during walking are presented in a box plot. The line in the center of the box represents the median value, the box represents the inter-quartile range and the whiskers represent the range. The circle indicates an outlier observation. The number of the observation is stated. (a) LG muscle activity over all COP positions in the preswing phase of the stance period of gait. (b) MG muscle activity over all COP positions in the load response phase of the stance. (c) TA muscle activity over four COP positions in the midstance phase of the stance period of gait.

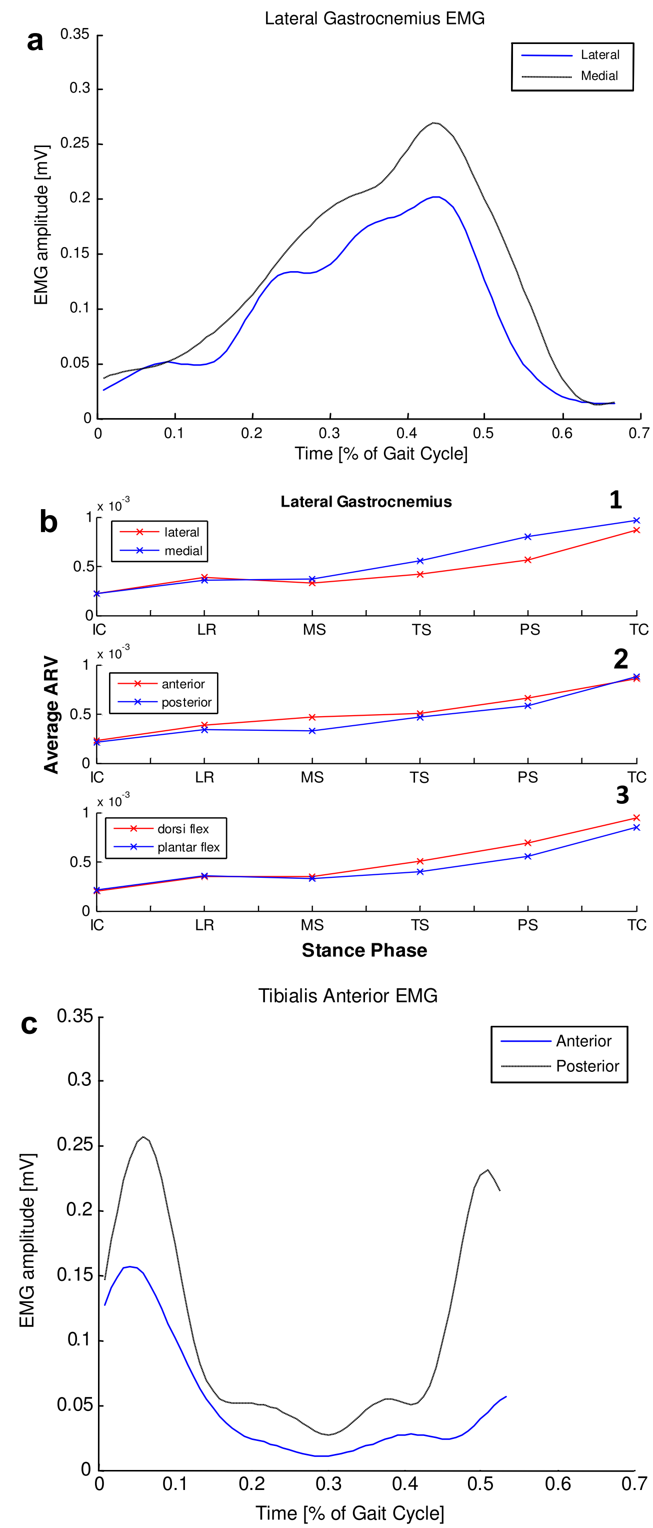

Fig. 4. Changes in activity of lower limb muscles over the entire stance period of gait in response to center of pressure (COP) manipulations during walking (a) EMG activity of the lateral gastrocnemius muscle of one subject over the entire stance period of gait during ambulation in the lateral vs. medial COP positions. (b) Averaged ARV of the lateral gastrocnemius muscle over the entire stance period of gait. (1) Ambulation in the lateral vs. medial COP positions; (2) ambulation in the anterior vs. posterior COP positions; (3) ambulation in the dorsi flexion vs. plantar flexion COP positions. (c) EMG activity of the tibialis anterior muscle of one subject over the entire stance period of gait during ambulation in the anterior vs. posterior COP positions.

of stance. Fig. 4a illustrates the EMG activity of the LG muscle for one person averaged over six steps in the lateral and medial COP positions. The activity of the LG was lower in the lateral COP position throughout the entire stance period. It can be observed in Fig. 4a that the activity of the LG was maximal in the middle of the stance phase; therefore, the significant changes of activation were observed during the terminal stance and the preswing.

Fig. 4b illustrates the mean ARV of the LG in all stance phases. Fig. 4b(1) shows that LG activity was similar in the medial and lateral COP positions for the beginning of the stance period but increased in the medial COP position in all the following phases. Fig. 4b(2) shows that LG activity was similar in the anterior and posterior COP positions for the initial and terminal phases of stance but increased in the anterior COP position in all other phases. Fig. 4b(3) illustrates that the LG activity was similar in the dorsi flexion and plantar flexion COP positions for the initial phase of stance but increased in the dorsi flexion COP position in all following phases.

The TA showed significantly more activation in the posterior COP position. More specifically, in the in the MS phase, the TA showed a 28% increase (from 1650 to 2115 μV) in activity when the COP was shifted from the anterior to the posterior position. Fig. 3c shows that the TA was significantly less active in the posterior and dorsi flexion positions than in the anterior and plantar flexion positions. Fig. 4c illustrates the EMG activity of the TA for one subject averaged over six steps. The TA activity was higher in the posterior COP position compared to the anterior position throughout the entire stance phase. It can be seen in Fig. 4c that the activity of the TA was higher in the initial and terminal stance phases; therefore, the significant changes of activation were observed in load response and midstance.

Lastly, the MG showed a consistent non-significant increase in activation in the lateral compared to the medial COP position, in the anterior compared to the posterior COP position and in the dorsi flexion compared to the plantar flexion COP position (Fig. 3b).

4. Discussion

The study examined the dynamic effects of COP manipulation during walking on the lower limb musculature. Variation of COP during walking was found to be associated with significant changes, mostly in distal muscles, specifically in the LG and TA. The changes in COP were made at the distal end of the limb using a shoe platform and, as such, the majority of the changes in muscle activation were seen distally.

As we initially hypothesized, a coronal COP shift was found to correlate with an activity change of the lateral and medial muscles, and a sagittal COP change was found to correlate with the anterior and posterior muscles. These results can be explained from mechanical (i.e., joints moments Haim et al., 2008, 2010) and anatomical perspectives. The GC is located on the posterior part of the calf. An anterior COP, therefore, should cause an increase in activation of the GC in an attempt to balance the leg. Similarly, the activation of TA, which is located in the anterior calf, should increase in a posterior COP. This cause and effect was demonstrated by the present study and by Mulavara et al. (1994). In that study, the authors reported that an anterior COP shift increased the activation of the GC muscle, while a posterior shift increased activation of the TA muscle. These same changes were observed in the present study during walking. The results of the present study expand on the findings of Mulavara et al. and demonstrate that these associations between muscle activation and COP position can be maintained during gait. Moreover, literature on the natural dynamic changes in the lower limb during gait without shoes has documented very similar changes in lower limb electromyograms in response to natural changes in COP during gait (Whittle, 2003).

In the present study, the GC was further split into LG and MG. Results showed that LG decreased in activation with a lateral shift in COP. This can also be explained mechanically and anatomically since, in a lateral COP, the LG should decrease in activation in an attempt to balance the leg. Similarly, the MG decreased in activation with a medial shift in COP. This latter effect, however, was not statistically significant. The reason for differences in the intensity of change between the LG and MG has yet to be determined. We suspect that the reason for this difference comes from the fact that the mechanical axis acting as the foot passes medial to the knee joint center during gait (Haim et al., 2008). This natural lack of symmetry between the medial and lateral compartments might be the reason for the significant changes in the LG activity with lateral and medial COP shifts as opposed to the evident, but non-significant, changes in MG activity with the same COP shifts.

The results of the present study may offer clinically relevant implications to several musculoskeletal pathologies. A COP manipulation can be applied, for example, to patients who suffer from foot drop, since these individuals suffer from weakness in the anterior leg muscles responsible for dorsiflexion of the foot (Gefen, 2001). In this case, these patients may benefit from a posterior manipulation of the COP during gait in order to increase the activity of anterior compartment muscles. A device that can induce increased GC activation, like the one described in our study, may also be a useful tool for training for OA patients, since it has been shown that ankle muscle strength plays an important role in maintaining dynamic balance in subjects with knee OA (Hsieh et al., 2008). However, further studies should be made to verify these suggestions.

Several limitations to the current study should be noted. Firstly, the changes in muscle activation during different COP shifts were not compared to a control configuration. This may include the device without any elements or with elements configured to a neutral COP. In the neutral COP position, the leg is more balanced than in the shifted COP positions; thus, the muscle work aimed to maintain balance is reduced. The same is true when the elements are removed (Haim et al., 2008). In addition, the subjects in the present study were young healthy males. As such, the results of the study are valid only for subjects with characteristics similar to this group. Further studies are needed before these findings can be extended to other populations, especially patients with neuromuscular disorders. We currently have studies underway with this goal in mind.

Conflict of interest statement

The authors have no conflict of interest to declare.

Acknowledgements

The authors thank APOS–Medical and Sports Technologies Ltd. for their generosity in contributing the devices used in the study and Nira Koren-Morag, PhD, for statistical assistance.

References

Crenshaw SJ, Pollo FE, Calton EF. Effects of lateral-wedged insoles on kinetics at the knee. Clin Orthop Relat Res 2000;375:185–92.

Croce RV. The effects of EMG biofeedback on strength acquisition. Biofeedback Self Regul 1986;11:299–310.

Dixit S, DiFiori JP, Burton M, Mines B. Management of patellofemoral pain syndrome. Am Fam Physician 2007;75(2):194–202.

Edwards L, Dixon J, Kent JR, Hodgson D, Whittaker VJ. Effect of shoe heel height on vastus medialis and vastus lateralis electromyographic activity during sit to stand. J Orthop Surg 2008;3:2.

Gefen A. Simulations of foot stability during gait characteristic of ankle dorsiflexor weakness in the elderly. IEEE Trans Neural Syst Rehabil Eng 2001;9(4):333–7.

Haim A, Rozen N, Wolf A. Control of knee coronal plane moment via modulation of center of pressure: a prospective gait analysis study. J Biomech 2008;41(14):3010–6.

Haim A, Rozen N, Wolf A. The influence of sagittal center of pressure offset on gait kinematics and kinetics. J Biomech 2010;43(5):969–77.

Hermens HJ, Freriks B, Merletti R, Stegeman D, Blok J, Rau G, et al. European Recommendations for Surface Electromyography (SENIAM) 1999. CD-rom. Roessingh Research and Development b.v., Enschede, the Netherlands, ISBN: 90-75452-14-4.

Hogenmiller MS, Lozada CJ. An update on osteoarthritis therapeutics. Curr Opin Rheumatol 2006;18(3):256–60.

Hsieh CJ, Yang Sai Wei, Hsieh LF. Role of muscle strength in dynamic balance for subjects with knee osteoarthritis. IFMBE Proc 2008;21(3):798–801.

Kakihana W, Akai M, Nakazawa K, Takashima T, Naito K, Torii S. Effects of laterally wedged insoles on knee and subtalar joint moments. Arch Phys Med Rehabil 2005;86:1465–71.

Krishnamoorthy V, Latash ML, Scholz JP, Zatsiorsky VM. Muscle modes during shifts of the center of pressure by standing persons: effect of instability and additional support. Exp Brain Res 2004;157:18–31.

Langzam E, Isakov E, Mizrahi J. Evaluation of methods for extraction of the volitional EMG in dynamic hybrid muscle activation. J Neuroengineering Rehabil 2006;3:27.
Maly MR, Culham EG, Costigan PA. Static and dynamic biomechanics of foot orthoses in people with medial compartment knee osteoarthritis. Clin Biomech (Bristol, Avon) 2002;17:603–10.
Mulavara AP, Verstraete MC, Simon BN. Correlation between muscle activity and the center of pressure. Eng Med Biol Soc 1994;1:384–5.
Nigg B, Hintzen S, Ferber R. Effect of an unstable shoe construction on lower extremity gait characteristics. Clin Biomech 2005;21:82–8.
Ogata K, Yasunaga M, Nomiyama H. The effect of wedged insoles on the thrust of osteoarthritic knee. Int Orthop 1997;21:308–12.
Whittle MW. Gait analysis: an introduction. 3rd ed. Philadelphia: Elsevier Ltd.; 2003. p. 67–8.
Xu H, Akai M, Kakurai S, Yokota K, Kaneko H. Effect of shoe modifications on center of pressure and in-shoe plantar pressures. Am J Phys Med Rehabil 1999;6:516–24.
Yasuda K, Sasaki T. The mechanics of treatment of the osteoarthritic knee with a wedged insole. Clin Orthop Relat Res 1987;215:162–72.

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Alon Wolf, PhD, earned all his academic degrees from the Faculty of Mechanical Engineering at Technion-I.I.T. In 2002 he joined the Robotics Institute of Carnegie Mellon University and the Institute for Computer Assisted Orthopaedic Surgery as a member of the research faculty. He was also an adjunct Assistant Professor in the School of Medicine of the University of Pittsburgh. In 2006 Dr. Wolf joined the Faculty of Mechanical Engineering at Technion, where he founded the Biorobotics and Biomechanics Lab (BRML). The scope of work done in the BRML provides the framework for fundamental theories in kinematics, biomechanics and mechanism design, with applications in medical robotics, rehabilitation robotics, and biorobotics, such as snake robots.