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Trunk kinematic, kinetic, and neuro-muscular response to foot center of pressure translation along the medio-lateral foot axis during gait

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ABSTRACT

Footwear devices that shift foot center of pressure (COP), thereby impacting lower-limb biomechanics to produce clinical benefit, have been studied regarding degenerative diseases of knee and hip joints, exhibiting evidence of clinical success. Ability to purposefully affect trunk biomechanics has not been investigated for this type of footwear. Fifteen healthy young male subjects underwent gait and electromyography analysis using a biomechanical device that shifts COP via moveable convex elements attached to the shoe sole. Analyses were performed in three COP configurations for pairwise comparison: (1) neutral (control) (2) laterally deviated, and (3) medially deviated. Sagittal and frontal-plane pelvis and spine kinematics, external oblique activity, and frontal and transverse-plane lumbar moments were affected by medio-lateral COP shift. Transverse-plane trunk kinematics, activity of the lumbar longissimus, latissimus dorsi, rectus abdominus, and quadratus lumborum, and sagittal-plane lumbar moment, were not significantly impacted. Two linear mixed effects models assessed predictive impact of (I) COP location, and (II) trunk kinematics and neuromuscular activity, on the significant lumbar moment parameters. The COP was a significant predictor of all modeled frontal and transverse-plane lumbar moment parameters, while pelvic and spine rotation, and lumbar longissimus activity were significant predictors of one frontal-plane lumbar moment parameter. Model results suggest that, although trunk biomechanics and muscle activity were altered by COP shift, COP offset influences lumbar kinetics directly, or via lower-limb changes not assessed in this study, but not by means of alteration of trunk kinematics or muscle activity. Further study may reveal implications in treatment of low back pain.

1. Introduction

Low back pain (LBP) is the leading cause of disability globally (Hoy et al., 2014), with a lifetime prevalence of up to 84% (Walker, 2000). It severely disrupts daily activities and work and, outside of the tremendous personal burden, has enormous economic detriment, costing tens of billions of dollars a year in productivity loss in the US alone (Rizzo et al., 1998).

Various causes of LBP have been proposed, including high impact forces during gait and abnormal foot biomechanics, with the latter associated with lumbopelvic muscular dysfunction (Papuga and Cambron, 2016). A connection has been observed between foot biomechanics and lumbopelvic function, as well as lumbopelvic biomechanics and back pathologies (Barwick et al., 2012). It is thus not surprising that footwear-based treatment options are popular for LBP, and aim to minimize biomechanical contributors to LBP, particularly those associated with abnormal foot posture (Barwick et al., 2012; Papuga and Cambron, 2016). Significant improvement in LBP has been reported with shoe insoles and foot orthotics (Cambron et al., 2017; Shabat et al., 2005). However, several systematic reviews concluded that, although they might be effective, there was insufficient evidence to support them as a treatment for LBP (Chuter et al., 2014; Sahar and Cohen, 2009).

A specialized subcategory of therapeutic footwear composed of foot-worn devices which shift foot center of pressure (COP), such as wedge insoles (Leitch et al., 2011; Van Gheluwe and Dananberg, 2004), variable stiffness shoes (Boyer et al., 2012; Jenkyn et al., 2011), and a unique device with movable convex

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elements built into a shoe sole (Khoury et al., 2013), has been shown to impact lower-limb kinematics, kinetics, and neuromuscular activity, and reduce joint loads (Crenshaw et al., 2000; Erhart et al., 2008; Goryachev et al., 2011b; Haim et al., 2010; Haim et al., 2008; Solomonow-Avnon et al., 2016a; Solomonow-Avnon et al., 2016b; Solomonow-Avnon et al., 2015). As such, clinical benefit was demonstrated for degenerative diseases of the knee (Barrios et al., 2009; Erhart-Hledik et al., 2012; Erhart et al., 2010; Haim et al., 2012; Rodrigues et al., 2008) and hip (Solomonow-Avnon et al., 2017). This type of footwear is suggested to affect biomechanics on the basis of the link segment model, which models the lower limbs as a system of rigid links, forming a functional kinetic unit (Zajac et al., 2002). According to this assumption, external perturbations applied to the foot segment (i.e., COP manipulation), therefore, would impact biomechanics of all other joints in the linked system (i.e., foot → ankle → knee → hip). To date, our research group exhibited correlation between specific COP shifts, and changes in biomechanics and neuromuscular activity, in both healthy and osteoarthritis patients, at the ankle (Haim et al., 2010), knee (Goryachev et al., 2011a; Goryachev et al., 2011b; Haim et al., 2011; Haim et al., 2010; Haim et al., 2008), and hip (Haim et al., 2010; Solomonow-Avnon et al., 2016a; Solomonow-Avnon et al., 2016b; Solomonow-Avnon et al., 2015), supporting the assumption, although links between biomechanical changes in each successive joint (i.e., ankle-knee, knee-hip, etc.) remain to be investigated explicitly. Apart from this, our recent study showed that COP shift directly affects the moment arm of the knee joint, thus impacting knee joint kinetics (Solomonow-Avnon et al., 2019). Similar mechanisms could additionally explain biomechanical changes in the other joints. Although clinical studies of one such device indicated improvement in spatiotemporal parameters, pain, function, and quality of life in non-specific LBP patients (Barzilay et al., 2016; Elbaz et al., 2009), to date, this type of footwear has not been investigated in regards to lumbopelvic biomechanics or trunk muscle activity.

The objective of the present study, therefore, is to quantify the effect of COP translation on kinematics, kinetics, and neuromuscular response of the lumbopelvic region, and demonstrate the potential to model or predict lumbar moments, based on COP, and trunk kinematics and muscle activity. Lumbar moments may be directly indicative of loads incurred on the lumbar spine (Andersson et al., 1980; National Institute for Occupational Safety and Health, 1981; Schultz et al., 1982), and thus represent a parameter of interest to control via COP translation. The current study is part of a larger study aiming to quantify and model effects of COP translation along various axes (i.e., antero-posterior and medio-lateral foot axes, COP offset along both axes simultaneously, and dorsi-flexion and plantar-flexion along the antero-posterior foot axis) constituting all possible implementable configurations of the study device. The current study focuses on COP translations along the medio-lateral foot axis. We hypothesized that COP translation along this axis would significantly impact kinematics, kinetics, and muscle activity of the lumbopelvic region, and that lumbar moments would change predictably in response to COP deviation.

2. Methods

2.1. Participants

The study group consisted of 15 healthy young male adults (Age = 26.3 ± 2.6 years) with similar anthropometric measures (Shoe size = French 43, Body mass = 71.5 ± 9.3 kg, Height = 174.9 ± 3.6 cm). Subject exclusion criteria were previous surgery, musculoskeletal injuries, or pathologies of the lower limbs or back, inability to cooperate with or understand instructions, problems with balance, or any neurological, orthopedic, or cardiovascular pathology that would cause pain, impact function, or prevent physical exertion. The Ethics Sub-Committee approved the study, and subjects gave informed consent after the purpose and methods were explained.

2.2. The biomechanical system

The APOS biomechanical device (APOS System, APOS Medical and Sports Technologies Ltd. Herzliya, Israel) was used. A detailed description of the device was previously reported (Haim et al., 2008). In brief, COP manipulation is accomplished using a shoe in which two adjustable convex-shaped biomechanical elements are attached to the feet by means of the sole, designed with two mounting rails (Fig. 1(a)). A medial or lateral shift of the elements causes a corresponding directional shift of COP (Haim et al., 2008; Khoury et al., 2013).

2.3. Experimental protocol

The neutral configuration was custom-positioned by a single trained physiotherapist, and defined as the position of the device elements in which there was the least varus, valgus, dorsiflexion, or plantar flexion torque about the ankle, as determined by observational gait analysis (Fig. 1(b)). The lateral and medial COP configurations (Fig. 1(c) and (d), respectively) were defined as the most

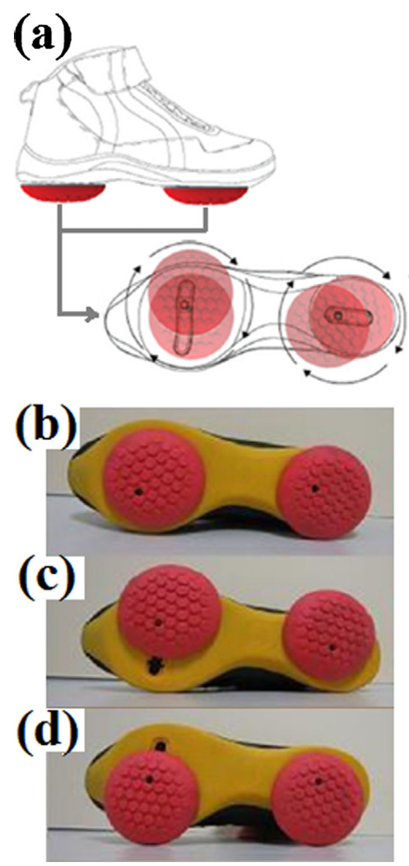

Fig. 1. (a) Biomechanical device with adjustable elements in a particular subject's (b) neutral, (c) lateral, and (d) medial configurations. Neutral was set such there was minimum observable varus, valgus, dorsiflexion, or plantar flexion ankle torque. Lateral and medial were set at the most extreme positions along the medio-lateral foot axis, within the constraints of the device, such that there was no observable change in ankle kinematics.

extreme deviations of the biomechanical elements from the neutral sagittal axis within the constraints of the device, such that there were no observable alterations in frontal-plane ankle kinematics. This was done so that the positionings would not induce abnormal biomechanics, or an awkward gait, and would therefore be clinically implementable.

Subjects were given a several-minute period prior to data acquisition to walk at a comfortable self-selected speed to become accustomed to the shoes. Subjects were asked to maintain this self-selected speed for all subsequent gait trials. Next, gait analyses were performed in the three COP configurations, neutral, lateral, and medial (Fig. 1(b–d), respectively), in random order. Before study initiation, a preliminary analysis confirmed that gait speed remained relatively constant across gait trials and device configurations, and thus was not controlled. This methodology was adopted to ensure the most natural gait possible.

2.4. Data acquisition and processing

Three-dimensional motion analysis was performed using an 8-camera Vicon motion analysis system (Oxford Metrics Ltd., Oxford, UK) for kinematic data capture, at a 100-Hz sampling frequency. Ground reaction forces (GRFs) were recorded by two 3-dimensional AMTI OR6–7-1000 force plates (Advanced Mechanical Technology, Inc., Watertown, MA) placed in tandem in the center of a 10-m walkway, at a 1000-Hz sampling frequency. Six to 8 successful walking trials were performed in each COP configuration, and study parameters were averaged for each set of trials. A successful trial was defined as one in which the dominant foot landed exclusively and entirely within the confines of one of the force plates. A standard marker set was used (Kadaba et al., 1990). Joint angles were calculated based on marker locations using 'PlugIn-Gait' (Oxford Metrics Ltd., Oxford, UK), and moments, normalized by body mass, were calculated via 'PlugInGait' using inverse dynamic analyses.

A surface electromyography (EMG) ZeroWire system (Aurion Ltd., Italy) was used to measure muscle activity of: lumbar longissimus (LL), latissimus dorsi (LD), rectus abdominis (RA), external obliques (EO), and quadratus lumborum (QL). After cleansing the skin with isopropyl alcohol, two disposable Ag/AgCl Skintact W-60 60-mm EMG recording electrodes (10–1000 Hz, 16-bit resolution) were adhered properly to each muscle on the dominant side with a 2-cm inter-electrode distance according to the Surface EMG for Non-Invasive Assessment of Muscles (SENIAM) recommendations for surface EMG placement (Freriks and Hermens, 1999). Electromyography data were recorded at 1000 Hz, simultaneously with motion analysis data, for each COP configuration, and stored on a computer. The EMG signals were bandpass filtered (30–500 Hz), and subsequently full-wave rectified. The 30-Hz lower limit of the filter was chosen to remove heartbeat artifacts (Drake and Callaghan, 2006).

The following gait parameters were calculated at GRF peaks 1 & 2: medio-lateral COP shift, sagittal-plane (S-P), frontal-plane (F-P), and transverse-plane (T-P) pelvic angles, spine angles, and lumbar moment (moment between pelvis and thorax). Medio-lateral COP shift was calculated as the perpendicular distance between the GRF origin, recorded as coordinates on the force plates, and the line connecting the toe and heel reflective markers (Haim et al., 2008; Solomonow-Avnon et al., 2016a). Cadence and walking speed were also recorded. Root mean square EMG (RMS-EMG; square root of the mean square EMG value) of the 5 muscles at GRF peaks 1 & 2 were calculated as RMS-EMG of the 5% of stance phase centered at the GRF peaks. Within subject normalization of EMG data was not necessary since measurements were compared in each COP configuration, and electrodes were not repositioned (Edwards et al., 2008). All analyses were performed for the dominant side.

2.5. Statistical analysis

Statistical analysis was performed in Matlab (MathWorks, Inc., Natick, MA). The Wilcoxon signed rank test was used for pairwise comparisons between each of the three device configurations. Subsequently, two separate linear mixed effects (LME) models, using restricted maximum likelihood, were tested for each significant lumbar moment parameter as the response variable: models as a function of (I) COP only, and (II) kinematic and neuromuscular components, excluding COP. For both models, the lumbar moment parameter was the response variable and subject (1–15) was a random effect. For Model I, COP was a fixed effect, while in Model II pelvic and spine kinematics, and RMS-EMG of all tested muscles, were potential fixed effects prior to a model selection procedure. The design of Model II involved a stepwise linear regression with Akaike Information Criterion (AIC) for model selection (Akaike, 1974), where AIC is provided as an option in Matlab for the criterion by which terms are added or removed from the model in the stepwise regression procedure. Fixed effects chosen by the procedure were used to create an LME model with subject as a random effect. Models I and II were not combined into one model, as COP was the only parameter that was directly and intentionally manipulated, and likely has a collinear effect on all biomechanical parameters.

3. Results

3.1. Gait parameters

All gait parameter results at GRF peaks 1 and 2 described below, and the associated statistical analysis, are presented in Table 1. Cadence and walking speed did not differ between COP configurations. At GRF peaks 1 and 2, COP was significantly more medial, on average, by 3.6 ± 2.8 mm and 3.5 ± 2.1 mm, respectively, in the medial COP configuration, while it was significantly more lateral by 4.4 ± 1.6 mm and 4.0 ± 3.0 mm, respectively, in the lateral COP configuration. Consequently, at peaks 1 and 2, there was an average significant difference in COP location of 8.0 ± 2.0 mm and 7.4 ± 3.1 mm, respectively, between medial and lateral configurations.

At GRF peak 1, S-P pelvic tilt was 5.4% (effect size = 0.5 ± 1.0°) significantly decreased (less forward tilt), on average, in the medial configuration (Fig. 2(a)). Additionally, F-P pelvic obliquity was significantly increased (the contralateral side of the pelvis is higher) by 19.6% (effect size = 0.7 ± 0.7°) and 11.9% (effect size = 0.4 ± 0.6°) for the medial (Fig. 2(a)) and lateral (Fig. 2(c)) configurations, respectively. At GRF peaks 1 and 2, S-P spine flexion was 11.7% (effect size = 0.8 ± 1.4°) and 19.5% (effect size = 1.0 ± 1.7°) increased (net extension angle was decreased), respectively, in the medial configuration (Fig. 2(a) and (b), respectively). At GRF peak 1, spine lateral flexion was increased by 11.5% (effect size = 0.6 ± 0.9°) in the medial configuration (Fig. 2(a)), and 9.5% (effect size = 0.5 ± 0.7°) in the medial compared to lateral configuration. No significant changes occurred for T-P pelvic or spine kinematics.

At GRF peak 2, F-P lumbar moment was significantly increased (ipsilateral flexion is positive), on average, by 287.9% (effect size = 62.1 ± 106.0 N-mm/kg) in the lateral configuration (Fig. 2(d)). At peaks 1 and 2, the F-P lumbar moment was significantly decreased by 151.1% (effect size = 82.5 ± 81.2) and 278.9% (effect size = 113.0 ± 109.6 N-mm/kg), respectively, in the medial compared to lateral configuration. At peak 2, the T-P lumbar moment was significantly increased (internal rotation is positive) by 14.1% (effect size = 19.1 ± 33.9 N-mm/kg) in the medial configuration (Fig. 2(b)), and decreased by 19.4%

Table 1
Comparison of gait and temporal parameters in the medial, neutral, and lateral foot center of pressure configurations.

Parameter	GRF Peak	M	N	L	p-values		
					M-N	L-N	M-L
COP [mm]							
ML-COP	Pk1	3.85 (5.55)	7.48 (4.57)	11.89 (4.42)	0.002*	0.000*	0.000*
	Pk2	7.61 (3.81)	11.10 (3.39)	15.06 (1.89)	0.000*	0.000*	0.000*
Pelvic Angles [°]							
Tilt	Pk1	9.45 (6.15)	9.99 (6.30)	9.61 (6.17)	0.048*	0.095	0.303
	Pk2	10.80 (6.56)	11.22 (6.60)	10.92 (6.65)	0.135	0.330	0.421
Obliquity	Pk1	4.03 (2.26)	3.37 (2.28)	3.77 (2.40)	0.003*	0.018*	0.121
	Pk2	−1.65 (1.91)	−1.49 (1.99)	−1.53 (1.87)	0.489	0.762	0.489
Rotation	Pk1	2.59 (2.52)	2.10 (2.46)	2.67 (2.18)	0.107	0.359	0.639
	Pk2	−3.02 (3.55)	−3.01 (2.84)	−2.13 (3.36)	1.000	0.083	0.073
Spine Angles [°]							
Flexion/Extension	Pk1	−5.87 (8.31)	−6.65 (7.99)	−6.21 (7.73)	0.018*	0.188	0.083
	Pk2	−4.12 (9.05)	−5.12 (8.40)	−4.60 (8.32)	0.041*	0.083	0.252
Lateral Flexion	Pk1	6.11 (2.60)	5.48 (2.82)	5.58 (2.72)	0.015*	0.277	0.022*
	Pk2	−2.53 (2.70)	−2.16 (2.79)	−2.54 (2.95)	0.208	0.188	0.890
Rotation	Pk1	3.32 (3.29)	2.85 (2.86)	3.19 (2.79)	0.107	0.095	0.421
	Pk2	−6.00 (4.31)	−5.85 (3.99)	−6.08 (4.57)	0.762	0.525	0.720
Lumbar Moments [N-mm/kg]							
Sagittal	Pk1	−236.67 (290.10)	−180.35 (232.36)	−213.33 (230.05)	0.389	0.188	0.389
	Pk2	−494.42 (291.75)	−452.61 (284.70)	−509.45 (293.25)	0.277	0.121	0.762
Frontal	Pk1	−137.13 (104.85)	−104.99 (101.84)	−54.62 (72.82)	0.169	0.064	0.004*
	Pk2	−72.49 (169.82)	−21.57 (159.87)	40.53 (109.47)	0.121	0.035*	0.003*
Transverse	Pk1	3.03 (56.06)	−0.97 (44.96)	−14.31 (45.42)	0.720	0.121	0.135
	Pk2	154.52 (59.40)	135.44 (50.40)	109.18 (61.27)	0.041*	0.010*	0.000*
Temporal Parameters							
Speed [m/s]		1.36 (0.12)	1.35 (0.10)	1.35 (0.12)	0.804	1.000	0.890
Cadence [steps/min]		105.91 (7.06)	105.35 (5.50)	105.80 (6.78)	0.847	0.890	0.934

Values are mean (standard deviation).
* indicates statistically significant result. Positive directions indicate lateral COP, forward pelvic tilt, ipsilateral side of pelvis lower than the contralateral (obliquity), internal pelvic rotation, spine flexion, spine lateral flexion (toward ipsilateral side), spine internal rotation (rotation of thorax toward contralateral side), lumbar flexion moment, lateral flexion moment (toward ipsilateral side), internal rotation moment. M = medial configuration; N = neutral configuration; L = lateral configuration; Pk1 = peak 1 of ground reaction force; Pk2 = peak 2 of ground reaction force; ML-COP = medio-lateral foot center of pressure.

(effect size = 26.3 ± 35.4 N-mm/kg) in the lateral configuration (Fig. 2(d)). Consequently, it was 41.5% (effect size = 45.3 ± 34.6 N-mm/kg) greater in the medial compared to lateral configuration. S-P lumbar moment was not significantly affected by medio-lateral COP translation.

3.2. EMG parameters

All EMG parameter results at GRF peaks 1 and 2 described below, and the associated statistical analysis, are presented in Table 2. At GRF peaks 1 and 2, EO RMS-EMG was significantly increased, on average, by 72.0% (effect size = 0.3 ± 0.4 mV) and 84.0% (effect size = 0.4 ± 0.4 mV), respectively, in the lateral configuration (Fig. 2(c) and (d), respectively). At peak 2, it was also increased by 62.9% (effect size = 0.3 ± 0.6 mV) in the medial configuration (Fig. 2(b)). The RMS-EMG of the 4 remaining muscles (LL, LD, RA, QL) was not significantly affected by medio-lateral COP translation.

3.3. Linear mixed effects models

Linear mixed effects Models I and II are presented in Tables 3 and 4, respectively, for the lumbar kinetic parameters which were significant (i.e., F-P lumbar moment at peaks 1 and 2 of the GRF, and T-P lumbar moment at peak 2).

With respect to LME Model I (Eq. (1)) for "Kinetic Parameter" representing F-P lumbar moment at peaks 1 and 2 of the GRF, and T-P lumbar moment at peak 2, COP was a significant predictor (response variable and significant fixed effect of COP shown in bold).

$$\boldsymbol{Kinetic\,Parameter} = 1 + \mathbf{COP} + (1|Subject) \quad (1)$$

According to the model, a 1-mm shift in the COP in the lateral direction increases peak 1 and 2 F-P lumbar moment by 7.82 and 10.28 N-mm/kg, respectively, and decreases peak 2 T-P lumbar moment by 4.54 N-mm/kg, or the opposite for a 1-mm medial shift.

The LME Model II for F-P lumbar moment at peaks 1 and 2 of the GRF, and T-P lumbar moment at peak 2, following model selection, is represented by Eqs. (2)–(4), respectively, where response variables and significant fixed effects are shown in bold.

$$\boldsymbol{F\text{-}Plumbar\ moment\ Pk1} = 1 + PT + PO + SFE + SLF + LL + RA + EO + (1|Subject) \quad (2)$$

$$\boldsymbol{F\text{-}Plumbar\ moment\ Pk}2 = 1 + PT + PO + \boldsymbol{PR} + SFE + \boldsymbol{SR} + \boldsymbol{LL} + LD + RA + (1|Subject) \quad (3)$$

$$\boldsymbol{T\text{-}Plumbar\ moment\ Pk}2 = 1 + PO + PR + SFE + SLF + SR + LL + LD + QL + (1|Subject) \quad (4)$$

where PT = Pelvic Tilt, PO = Pelvic Obliquity, PR = Pelvic Rotation, SFE = Spine Flexion/Extension, SLF = Spine Lateral Flexion, SR = Spine Rotation, LL = Lumbar Longissimus RMS-EMG, LD = Latissimus Dorsi RMS-EMG, RA = Rectus Abdominis RMS-EMG, EO = External Oblique RMS-EMG, and QL = Quadratus Lumborum RMS-EMG.

With respect to F-P lumbar moment at peak 1 of the GRF (Eq. (2)) and T-P lumbar moment at peak 2 (Eq. (4)), none of the fixed effects terms chosen by the model selection procedure were significant in the LME model. For F-P lumbar moment at peak 2 (Eq. (3)), pelvic rotation, spine rotation, and LL RMS-EMG were significant

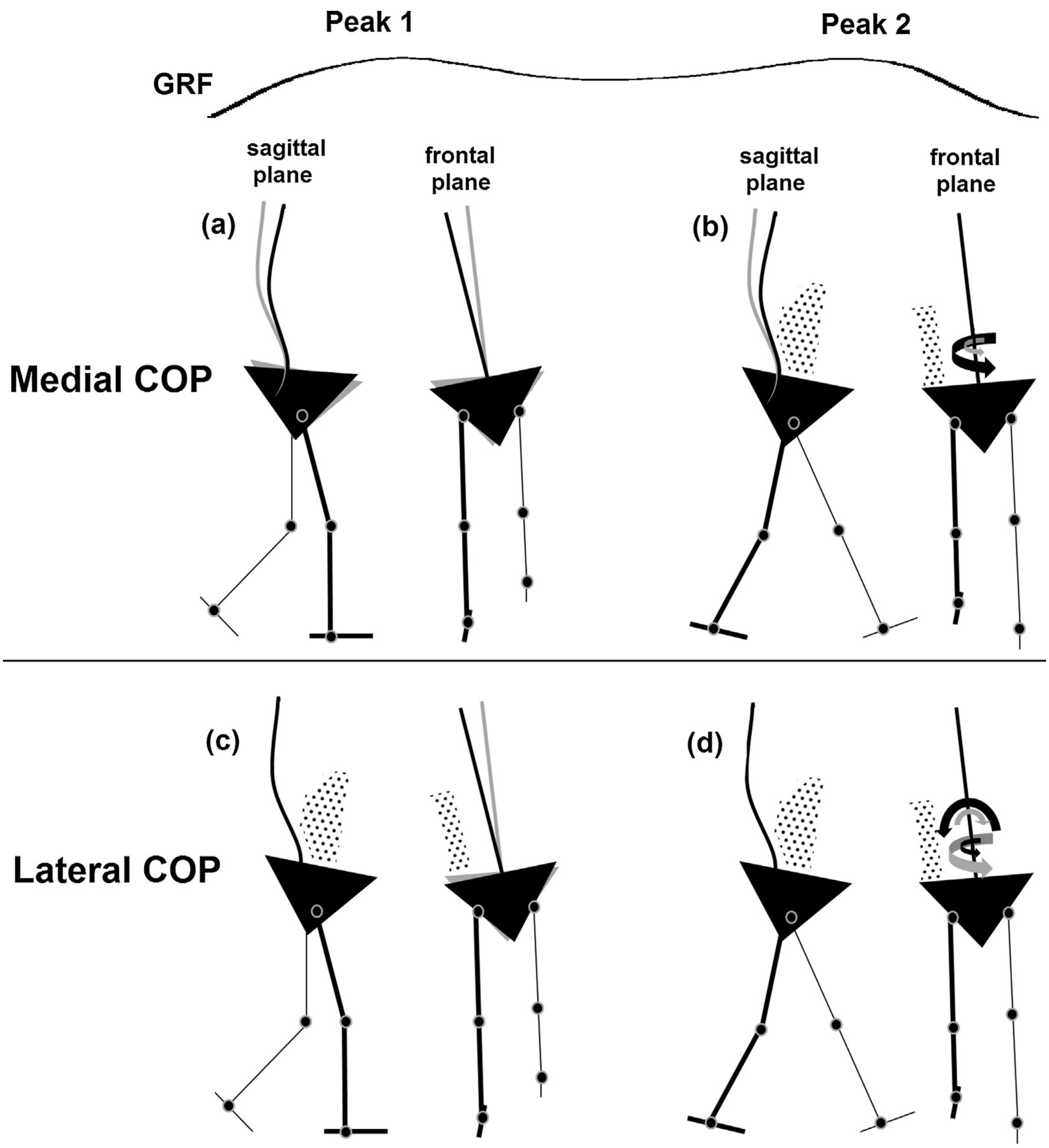

Fig. 2. Illustration of significant changes in pelvic and spine kinematics, external oblique activity (black dots), and lumbar moments (arrows) observed at peaks 1 and 2 of the ground reaction force (GRF) with a medial (a and b, respectively) or lateral (c and d, respectively) shift in foot center of pressure (black overlay) compared to neutral configuration (gray). Measurements are for the right side (frontal plane is with face forward). Black dots indicate increased external oblique activity compared to neutral. Lumbar moments are in the direction of obliquity towards the ipsilateral side if arrow is counterclockwise, and toward the contralateral side if clockwise. All depicted transverse-plane lumbar moments are in the direction of internal rotation. A larger arrow indicates a larger moment. Spine curvature and lower limbs are shown for illustration purposes only with the right limb in bold. Illustrations are not drawn to scale, and are not anatomically accurate.

predictors. According to the model, internal pelvic rotation, external spine rotation, and a decrease in LL activity, act to increase lumbar moment at peak 2, or the opposite direction for a decrease in lumbar moment.

It must be noted that the increase or decrease predicted for the lumbar moment by the fixed effects terms does not necessarily refer to the magnitude of the moment, but rather to increase toward the positive or negative directions, respectively, where a net positive value indicates ipsilateral flexion moment, and a net negative value, contralateral flexion moment.

4. Discussion

In accordance with the hypothesis, COP shift along the medio-lateral foot axis significantly impacted spine and pelvic kinematics and kinetics, and trunk muscle activity. Specifically, all sagittal and frontal-plane pelvis and spine kinematics (Table 1), as well as external oblique activity (Table 2), were affected by either a medial or lateral COP shift from the neutral, at either GRF peak 1 or 2, or both. Correspondingly, frontal-plane lumbar moment was impacted at peaks 1 and 2, and transverse-plane lumbar moment at peak 2 (Table 2). Further, also in accordance with the hypothesis, COP offset was a significant predictor of lower-back kinetics, as seen by a linear mixed effects model (Eq. (1), Table 3).

Although significant differences in pelvic and spine kinematics existed, and percent changes may seem substantial, absolute differences are very small (i.e., one degree or less, on average). This is consistent with very small differences observed in hip kinematics in previous similar studies (Solomonow-Avnon et al., 2016a; Solomonow-Avnon et al., 2015). Although the differences in hip kinematics were small, they were associated with changes in hip kinetics and neuromuscular activity that had potential clinical benefit (Solomonow-Avnon et al., 2016a; Solomonow-Avnon et al.,

Table 2
Comparison of RMS-EMG in the medial, neutral, and lateral foot center of pressure configurations.

Muscle	GRF Peak	M	N	L	p-values		
					M-N	L-N	M-L
RMS-EMG [mV]							
LL	Pk1	0.699 (0.667)	0.402 (0.384)	0.560 (0.512)	0.083	0.095	0.525
	Pk2	0.723 (0.650)	0.472 (0.494)	0.666 (0.636)	0.208	0.151	0.890
LD	Pk1	0.021 (0.009)	0.024 (0.015)	0.024 (0.013)	0.330	0.689	0.600
	Pk2	0.032 (0.015)	0.037 (0.020)	0.036 (0.020)	0.489	0.489	0.330
RA	Pk1	0.197 (0.681)	0.034 (0.043)	0.026 (0.025)	0.639	0.762	0.639
	Pk2	0.205 (0.713)	0.026 (0.024)	0.026 (0.023)	0.978	0.489	0.847
EO	Pk1	0.700 (0.528)	0.483 (0.548)	0.831 (0.611)	0.208	0.008*	0.600
	Pk2	0.702 (0.530)	0.431 (0.464)	0.793 (0.642)	0.030*	0.001*	0.978
QL	Pk1	0.042 (0.061)	0.018 (0.008)	0.027 (0.023)	0.561	0.252	0.720
	Pk2	0.043 (0.044)	0.029 (0.012)	0.032 (0.016)	0.489	0.840	0.847

Values are mean (standard deviation).
* indicates statistically significant result. M = medial configuration; N = neutral configuration; L = lateral configuration; Pk1 = peak 1 of ground reaction force; Pk2 = peak 2 of ground reaction force; LL = lumbar longissimus; LD = latissimus dorsi; RA = rectus abdominus; EO = external obliques; QL = quadratus lumborum; RMS-EMG = root mean square EMG.

Table 3
Linear mixed effects Model I parameters.

Fixed effects	Coefficient (standard error)	*p*-value
Response variable: Frontal-plane lumbar moment Pk1 (R^2 = 0.721 adjusted)		
Intercept	−159.40 (31.10)	0.000*
ML-COP offset [mm]	7.82 (2.19)	0.001*
Response variable: Frontal-plane lumbar moment Pk2 (R^2 = 0.760 adjusted)		
Intercept	−133.60 (52.99)	0.015*
ML-COP offset [mm]	10.28 (3.43)	0.005*
Response variable: Transverse-plane lumbar moment Pk2 (R^2 = 0.837 adjusted)		
Intercept	184.17 (19.07)	0.000*
ML-COP offset [mm]	−4.54 (1.14)	0.000*

* indicates statistically significant result. ML-COP = medio-lateral COP; Pk1 = peak 1 of the ground reaction force; Pk2 = peak 2 of the ground reaction force.

Table 4
Linear mixed effects Model II parameters.

Fixed effects	Coefficient (standard error)	*p*-value
Response variable: Frontal-plane lumbar moment Pk1 (R^2 = 0.527 adjusted)		
Intercept	35.53 (66.23)	0.593
Pelvic tilt [°]	−4.93 (8.37)	0.557
Pelvic obliquity [°]	4.02 (11.59)	0.729
Spine flexion/extension [°]	1.37 (6.54)	0.834
Spine lateral flexion [°]	−16.11 (9.17)	0.081
RMS-EMG LL [mV]	−17.96 (19.59)	0.361
RMS-EMG RA [mV]	−4.56 (19.47)	0.815
RMS-EMG EO [mV]	14.03 (17.31)	0.419
Response variable: Frontal-plane lumbar moment Pk2 (R^2 = 0.728 adjusted)		
Intercept	−167.96 (75.67)	0.028*
Pelvic tilt [°]	6.41 (7.85)	0.416
Pelvic obliquity [°]	−14.38 (10.32)	0.166
Pelvic rotation [°]	15.48 (5.55)	0.006*
Spine flexion/extension [°]	−1.41 (6.03)	0.815
Spine rotation [°]	−20.96 (5.59)	0.000*
RMS-EMG LL [mV]	−35.18 (16.15)	0.031*
RMS-EMG LD [mV]	−134.96 (521.7)	0.796
RMS-EMG RA [mV]	−8.30 (18.51)	0.655
Response variable: Transverse-plane lumbar moment Pk2 (R^2 = 0.772 adjusted)		
Intercept	140.23 (22.75)	0.000*
Pelvic obliquity [°]	2.23 (6.99)	0.751
Pelvic rotation [°]	−4.38 (2.56)	0.089
Spine flexion/extension [°]	2.05 (1.49)	0.171
Spine lateral flexion [°]	−6.69 (5.09)	0.191
Spine rotation [°]	1.51 (3.47)	0.664
RMS-EMG LL [mV]	−4.82 (6.29)	0.445
RMS-EMG LD [mV]	−370.22 (207.28)	0.076
RMS-EMG QL [mV]	14.28 (107.4)	0.894

* indicates statistically significant result. Pk1 = peak 1 of the ground reaction force; Pk2 = peak 2 of the ground reaction force; RMS-EMG = root mean square EMG; LL = lumbar longissimus; LD = latissimus dorsi; RA = rectus abdominus; EO = external obliques; QL = quadratus lumborum; RMS-EMG = root mean square EMG.

2016b; Solomonow-Avnon et al., 2015). Thus, further study is required to assess clinical impact of the differences observed in the present study.

Additionally, more complex multi-factorial relationships may exist between trunk kinetics, kinematics, and neuromuscular response, however, for simplicity and as a preliminary foundation for further study, a simple linear model was chosen. Additionally, for Model II (Eqs. (2)–(4), Table 4), it is possible that interactions between fixed effects had a predictive effect on response variables; however, for simplicity these were not represented. Regardless, both Models I and II (Tables 3 and 4, respectively) showed a good fit for the model for all of the lumbar moment planar components, with the exception of Model II for the sagittal plane. The r-squared values for Model I were slightly higher than those for the corresponding lumbar moment components of Model II, suggesting that COP may explain slightly more of the variance in lumbar moment than trunk neuromuscular activity and pelvic and spine kinematics. Furthermore, COP was a significant predictor of all three lumbar moment parameters that changed significantly with medio-lateral COP shift (i.e., frontal-plane lumbar moment at peaks 1 and 2 of the GRF, and transverse-plane lumbar moment at peak 2), whereas kinematic and neuromuscular predictors were significant for frontal-plane lumbar moment at peak 2 of the GRF only. These results may suggest that COP shift primarily affects lumbar moment by impacting the lever arm between the external force acting upon the lumbopelvic-hip complex, and the lumbar segment, rather than by a secondary consequence of altering kinematics and neuromuscular activity. This is supported by the results showing that a lateral deviation in COP, in which the lever arm would intuitively increase as a result of the origin of the ground reaction force vector moving farther away from the body, causes an increase in lumbar moment (Table 1), perhaps by means of altering frontal-plane lumbopelvic moment (i.e., the moment that would act to rotate the pelvis about the sagittal axis). Likewise, a medial deviation, in which the lever arm would intuitively decrease, causes a decrease in lumbar moment. This, however, is mere speculation, as the effect of COP shift on the magnitude and orientation of the GRF was not analyzed in the present study, and COP shift may or may not impact the lever arm with respect to the spine. In addition, COP may have impacted lumbar moment via lower-limb biomechanical changes that were not assessed in this study. Further study may reveal a more precise mechanism to refine the model associating COP shift with lower-back kinetic changes.

Interestingly, transverse-plane pelvic and spine kinematics, as well as lumbar longissimus activity, were significant predictors of frontal-plane lumbar moment at peak 2 of the GRF (Eq. (3), Table 4). Although this was the case, none of these parameters were significantly influenced by the medio-lateral COP manipulation of the present study (Tables 1 and 2), further suggesting that the COP shift itself may be the key parameter to alter lumbar moment.

For simplicity, we used the standard 'PlugInGait' kinematic model, which uses inverse dynamics to calculate joint kinetics. In the present study, we took the pelvis-thorax complex, referred to as the lumbar segment, to be a gross representation of the lower back. The Vicon 'PlugInGait' documentation refers to this segment as the "waist". Future studies may wish to use a more complex model of the spine to assess vertebral kinematics and intervertebral loads, taking into consideration spinal curvature and multiple vertebral segments, which are not represented in the 'PlugInGait' model. Additionally, any attempt to propose a mechanism for the link between COP shift and the specific changes in pelvic and spine biomechanics observed in this study would be highly speculative without a complete description involving the lower limbs and GRF. Previous studies have described changes in ankle (Haim et al., 2010), knee (Goryachev et al., 2011b; Haim et al., 2010; Haim et al., 2008) and hip (Haim et al., 2010; Solomonow-Avnon et al., 2016b; Solomonow-Avnon et al., 2015) biomechanics with medio-lateral COP manipulation using the device of this study. However, not all of the kinematic and kinetic variables in the three planes were reported, and no study exists on the ankle joint. Therefore, future study should aim to reveal a mechanism for the biomechanical link between COP shift and pelvic and spine kinematics, kinetics, and neuromuscular response, involving all lower-limb segments. Nevertheless, the ability to impact trunk kinematics, kinetics, and neuromuscular activity, as well as to predict kinetics based on COP offset, was successfully demonstrated.

Several limitations of the present study must be acknowledged. The neutral configuration was used so as not to upset the kinematic model by moving reflective markers. Thus, the only change in the experimental setup was repositioning of the device elements (COP shift). A regular shoe with a flat bottom presents a different walking condition than that with the instability induced by the convexity of the device elements, and thus has less neuromuscular demand. Therefore, using the neutral configuration provides an adequate reference to assess medial or lateral COP offsets, and provides three points along the medio-lateral foot axis. Additionally, results pertain only to subjects with similar characteristics to the cohort of the present study (i.e., healthy young males), and further study is required to confirm the results in other populations. Further, the models of the present study assume a homogeneous response of lower back kinetics to the predictors defined in the study; however, a subject-specific model may be more appropriate for determining model parameters. Nevertheless, the models of the present study demonstrate the relevant association between the predictors and the kinetic variables. Finally, we used surface electromyography to measure activity from a selection of relatively superficial muscles. Medio-lateral COP translation may have a substantial impact on other muscles we did not measure, including those that are more deeply situated.

5. Summary

The present study exhibited the ability to predictably manipulate the lumbar moment via translation of COP along the medio-lateral foot axis. Manipulation of medio-lateral axis COP has a significant impact on sagittal and frontal-plane pelvis and spine kinematics, external oblique activity, and frontal and transverse-plane lumbar moments. While pelvic and spine rotation, and lumbar longissimus activity were significant predictors of one out of three lumbar moment parameters significantly affected by medio-lateral COP translation (i.e., F-P lumbar moment at GRF Peak 2) in a linear mixed effects model, COP by itself was a predictor of all lumbar moment parameters significantly affected by medio-lateral COP translation (i.e., F-P lumbar moment at GRF Peaks 1 & 2, and T-P lumbar moment at GRF Peak 2) in a separate linear mixed effects model, suggesting that COP is an independent predictor of lumbar kinetics. Results of the present study may have clinical implications. Specifically, ability to impact and predictably manipulate lower-back kinetics by externally shifting the COP may have implications for treatment of low-back pain or low-back injury by potentially altering inter-vertebral loads. Additionally, results of the study bring us one step closer to full characterization of the trunk biomechanical response to externally applied foot perturbations, and hence add to the general base of knowledge regarding spine biomechanics.

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Conflict of interest statement

Avi Elbaz and Amit Mor hold shares in APOS Medical and Sports Technologies Ltd. The remaining authors have no conflict of interest to declare.

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