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Foot Center of Pressure Trajectory Alteration by Biomechanical Manipulation of Shoe Design

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Abstract

Background: Footwear-generated biomechanical manipulations have been shown to alter lower limb kinetics. It has been suggested that this is due to altered trajectory of the foot's center of pressure (COP), conveying a shift in location of the ground reaction force and modifying moments and forces acting on proximal body segments. However, past studies have focused on qualitative association between footwear design and the COP locus. Moreover, this association was calculated via indirect analysis. The purpose of the present study was to directly examine and quantify the correlation between measured footwear biomechanical manipulation and the location of the COP trajectory during gait.
Methods: A novel biomechanical device allowing flexible positioning of 2 convex-shaped elements attached to its sole was utilized. A total of 20 healthy male adults underwent direct in-shoe pressure measurements while walking with the device set at 7 mediolateral configurations. COP data were collected during gait and analyzed with respect to different stance subphases.
Results: COP location significantly correlated with a shift of the elements medially or laterally. The linear model describing this correlation was found to be statistically significant.
Conclusion: There was significant correlation between the plantar orientation of the shoe device configuration and the COP.
Clinical Relevance: Changes in COP trajectory may be valuable in patients suffering from multiple foot disorders elevating pressure on the foot. Accurate COP control could aid in the manipulation of the forces acting on the proximal joints during gait. In addition, these findings may have implications in the field of biomechanical apparatus design and practice.

Keywords: foot center of pressure, gait, plantar pressures, wedge position

The foot center of pressure (COP) is a theoretical locus about the foot, which is the average location of all the forces acting between the plantar surface of the foot and the ground at any given time during the stance phase.[3,15] Typically, the COP trajectory propagates from the medial aspect of the hindfoot, curving laterally at midstance and progressing medially at terminal stance.[4,8,15] This reflects both the anatomical properties of the foot and the relationship of the location of the body's center of mass to the location of the foot.[1,2,6,8,11,18] By definition, the projection on the ground plane of the ground reaction force (GRF), which is the average vector exerted by the ground on the body, and the trajectory of the COP during single limb support overlap.[3,19]

Footwear-generated biomechanical manipulations have been proposed to alter the trajectory of the COP, thereby altering the location of the GRF and modifying moments and forces acting on proximal body segments.[12] This principle has been the focus of past studies and has been widely implemented in clinical practice.[10,12] The application of a laterally wedged shoe insole was first introduced in the 1980s.[21] Maly et al hypothesized that the mechanism of reduction of knee adduction moment (KAM) with the use of insoles is a lateral shift in the COP.[13] Xu et al confirmed that insole conditions caused a change in the location of the COP during gait.[20] Using computer modeling simulation, Shelburne et al reported that a 1 mm displacement of the COP can decrease the KAM by 2%.[16]

The association between the COP and the footwear design is not trivial. On one hand, if the interface between the ground and the plantar surface of the footwear were

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Figure 1. Biomechanical platform and mobile elements.

infinitesimal, then the COP and the location of the contact point would be congruent. However, the ground contact point expands on a wider area. On the other hand, assuming force and moment equilibrium, the location of the COP must overlap with the vertical projection of the body's center of mass. Yet, in a dynamic situation, this may not be so;[17] the GRF vector is a reflection of the total mass-times-acceleration product of all body segments and therefore represents the total of all net muscle and gravitational forces acting at each instant of time over the stance period.[19]

In a previous study,[7] we utilized a novel foot-worn biomechanical device composed of 2 adjustable convex biomechanical elements placed under the sole of a shoe, allowing controlled manipulation of the contact point of the device and the ground. We examined the outcome of foot manipulation on the location of the COP and KAM in healthy subjects and found that the location of the COP was significantly associated with the mediolateral orientation of the elements of the biomechanical device. However, the correlation between the extent of the biomechanical manipulation (ie, the distance of element shift in the coronal plane from the neutral axis) and the location of the COP during gait was not established. Therefore, we devised the current study to establish this relationship. We hypothesized that manipulation of the biomechanical apparatus via translation of elements in the coronal plane (ie, from medial to lateral) would result in a direct shift of the COP trajectory.

Methods

Participants

A total of 20 healthy male volunteers without any known history of injury or any postural or skeletal disorder that could affect normal posture or gait composed the study cohort. All had the same shoe size (French 43), a right dominant leg, and a similar anthropometric profile. The participants' mean ± SD age was 24 years ± 11 months, height was 176.1 ± 3.29 cm, and weight was 68.8 ± 6.67 kg.

The study was conducted according to the Helsinki Declaration and was approved by the Ethics Committee of Ha'emek Medical Center in Afula, Israel. The purpose and methods of the study were explained to the subjects, and all participants gave written informed consent prior to their inclusion.

The Biomechanical System

A novel biomechanical shoe device (APOS-Medical and Sports Technologies Ltd., Herzliya, Israel) that consists of 2 convex-shaped biomechanical elements attached to each of the feet (Figure 1) was utilized. One was located under the hindfoot region and the other was located under the forefoot region. The elements were available at different levels of convexity and were attached to the subject's foot using a platform in the form of a shoe. The platform was equipped with a specially designed sole that consisted of 2 mounting rails enabling flexible positioning of each element under each region. Each element position could be calibrated individually to convey specific biomechanical challenges in multiple planes.

Experimental Protocol

Each participant was given time to become familiar with the laboratory environment and was allowed a number of walking trials with the biomechanical system prior to its calibration. The calibration was performed by a single trained physiotherapist; the first position of the elements for the "functional neutral sagittal axis" was determined and documented. The functional neutral axis (N 0.0) was defined as the position at which the apparatus conveyed the least valgus or varus torque at the ankle of the individual being examined. Six medial and lateral axes were then defined as 0.4, 0.8, and 1.2 cm medial and lateral deviation of the biomechanical elements from the neutral sagittal axis: M 0.4, M 08, M 1.2, L 0.4, L 0.8, and L 1.2, respectively (Figure 2).

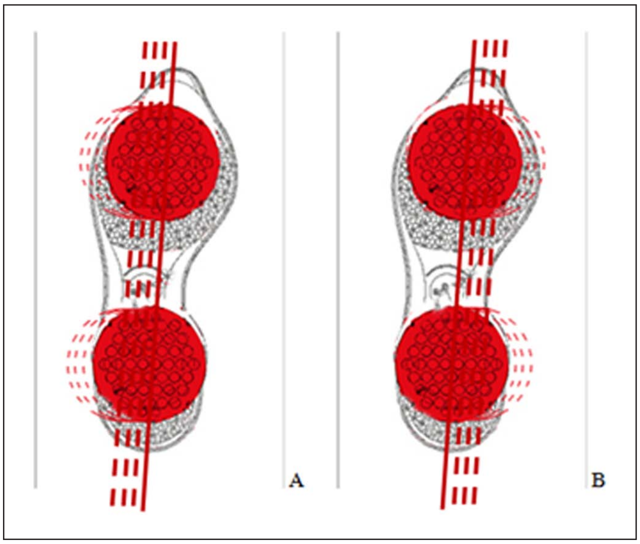

Figure 2. Different configurations of the biomechanical shoe device: Six medial (M 0.4, M 0.8, M 1.2) and lateral (L 0.4, L 0.8, L 1.2) axes were defined as 0.4, 0.8, and 1.2 cm medial (dashed lines in B) and lateral (dashed lines in A) deviation of the biomechanical elements from the neutral (N 0.0) sagittal axis (solid line), respectively.

Successive plantar pressure testing was performed with the biomechanical elements placed in 7 configurations: neutral configuration, 3 medial displacements, and 3 lateral displacements.

All measurements were applied while the participant walked on a 10 m flat surface at a constant velocity indicated by a metronome to ensure consistent cadence throughout the trials. All conditions were tested in a random order on the same day. To ensure uniformity of testing conditions, all subjects were provided with the same biomechanical system, and all calibrations of the biomechanical device were performed by the same physiotherapist.

Data Acquisition and Processing

Plantar pressure testing was applied using the Pedar-x pressure-measurement system (Novel Electronics, St. Paul, MN), which consisted of insoles 2.5 mm in width incorporating a matrix of 99 pressure-sensitive capacitive insoles that could be placed into the subject's footwear to measure pressure during gait. The sensors were sampled at a rate of 100 Hz to create a compound pressure-time data series that provided the centroid of loading at each point in time. The centroid of loading measured during a stance phase provided the COP trajectory during that time.

To examine the relationship between the coronal shift of the elements and the COP trajectories, the centroid of loading data was extracted and analyzed using Matlab. Stance phase data was identified as the data in which the resultant load exceeded a threshold of 5% of the maximum load measured through the trial. First and last steps were excluded. Stance phase was subdivided into 4 subphases: loading response: 0% to 10%; midstance: 10% to 30%; terminal stance: 30% to 50%; and preswing: 50% to 60%. A mean medial-lateral COP coordinate was calculated in association with a specific stage of the stance and averaged over all steps. Finally, the relative COP offset was assessed at each subphase by examining the difference in the mean COP coordinate of each configuration with respect to the neutral configurations. Lateral and medial offset of the COP were defined as positive and negative values, respectively, and reported in mm.

Statistical Analysis

Data were analyzed by R© version 2.11.1 (CRAN, Vienna, Austria). The element changes from medial to lateral were defined as negative and positive shifts from the neutral position; for example, a medial shift of 1.2 equals −1.2, and a lateral shift of 1.2 equals 1.2. To determine if the mean COP was affected by all the shifts of the elements and if the effect was dependent on the phase of stance, a linear model was fit to the data by multivariate regression. In the regression model, the COP was the dependent variable and gait phase, element shift, and their interaction were included as independent covariates. Element shift was considered a continuous variable and gait phase was considered a categorical variable. An analysis of variance (ANOVA) table of the linear model was used to determine the significance of each variable and interaction. R-squared was used to determine the model's goodness of fit. A statistically significant difference was considered as a P value less or equal to .05. All reported P values are 2-sided.

Results

The COP trajectory throughout the stance shifted in accordance with the offset of the biomechanical elements. Figure 3 demonstrates a representative subject's plots of 7 COP trajectories measured during the stance phase of gait with the biomechanical elements placed in 7 conditions: neutral configuration, 3 medial displacements (negative values), and 3 lateral displacements (positive values). Mean values and standard deviations of the mean COP at each subphase with 3 medial (M 0.4, M 0.8, M 1.2), 3 lateral (L 0.4, L 0.8, L 1.2), and neutral (N 0.0) positioning of the biomechanical elements are presented in Table 1.

The ANOVA table of the linear model showed that all 3 covariates in the model were statistically significant. In other words, mean COP varied significantly with the shift of the elements medially and laterally ($P < .0001$) and with different subphases of the stance ($P < .0001$), as well as with the interaction between both factors ($P < .0001$).

The linear model showed a high goodness of fit (R-squared of .839). The linear model was found to be

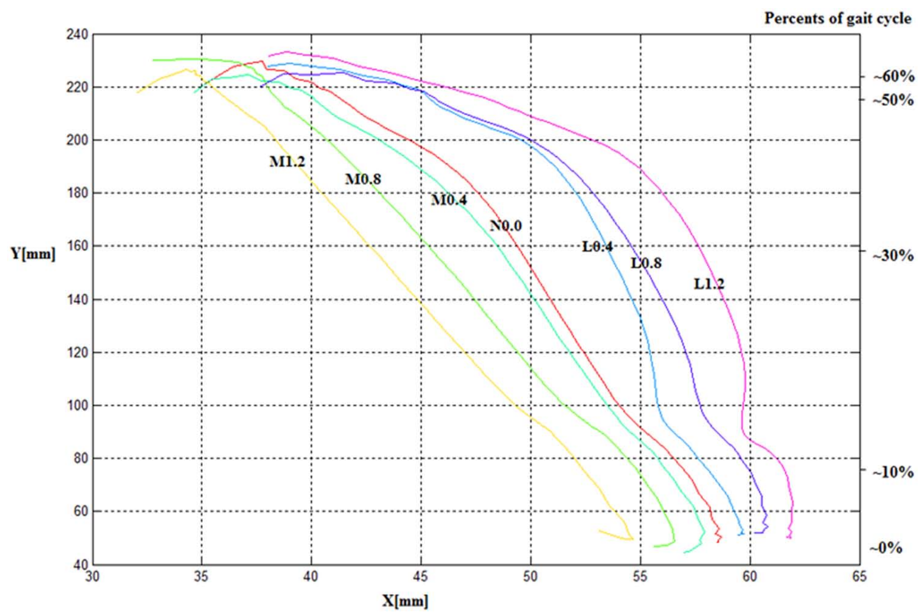

Figure 3. A representative subject's plot of the COP trajectories at 3 medial (M 1.2, M 0.8, M 0.4), 3 lateral (L 1.2, L 0.8, L 0.4), and 1 neutral (N 0.0) configurations. The left Y-axis represents the sagittal axis (anterio-posterior) and the X represents the coronal axis (medial-lateral). Increased X coordinate indicates a shift of the COP toward the lateral side and increased Y coordinate indicates a shift of the COP toward the anterior side. The right Y-axis corresponds to the percentage of gait cycle. The axis is divided according to the 4 subphases of the stance phase of the gait cycle.

Table 1. Mean ± SD Values of the Mean COP Data Calculated at Different Subphases of the Stance

	Stance Subphase							
	Loading Response		Midstance		Terminal Stance		Preswing	
Medial −1.2	51.25	± 1.83	45.01	± 2.94	36.8	± 3.14	33.33	± 3.01
Medial −0.8	52.16	± 2.06	46.68	± 3.11	38.62	± 3.39	34.62	± 3.19
Medial −0.4	53.25	± 2.15	48.65	± 3.02	40.66	± 3.05	35.79	± 3.27
Neutral 0.0	54	± 2.15	50.75	± 2.71	43.84	± 3.06	38.7	± 3.61
Lateral 0.4	54.16	± 2.5	50.75	± 3.07	44.41	± 3.54	38.71	± 4.04
Lateral 0.8	55.35	± 2.39	52.11	± 2.8	45.44	± 3.26	39.18	± 4.01
Lateral 1.2	56.63	± 2.35	53.97	± 2.85	47.94	± 3.78	40.73	± 4.14

The element changes from medial to lateral were defined as negative and positive shifts from the neutral position, respectively.

statistically significant ($P < .001$). The multivariate linear regression is presented in Table 2 as 4 simple linear models according to gait subphase. The observed means for each element shift according to gait subphase are presented as points in Figure 4. The linear models for each gait subphase are presented as linear lines in Figure 4. The goodness of fit of the linear model can be observed from the figure itself.

Discussion

The relationship between COP trajectory and footwear-generated biomechanical manipulations has been widely investigated; however, past studies offered a qualitative assessment. The current study tested the linear association between the plantar orientation of the biomechanical device configuration and the COP trajectory. This was significant throughout the whole stance phase (at all 4 subphases—Figure 4). Different augmentations of this association, however, were observed at different subphases of the stance. The observed linear association between the plantar orientation of the biomechanical device configuration and the COP trajectory was highest (ie, greater slope value) at the midstance and terminal stance. We speculate that the higher associations at the midstance and terminal stance

Table 2. Linear Models According to Gait Phase

Gait Subphase	Linear Equation	Intercept *P* Value	Slope *P* Value
Loading response	COP = 53.8 + 2.1 × ES	< .0001	.0014
Midstance	COP = 49.7 + 3.5 × ES	< .0001	< .0001
Terminal stance	COP = 42.6 + 4.4 × ES	< .0001	.0327
Preswing	COP = 37.3 + 3.0 × ES	< .0001	.2757

COP = center of pressure; ES = element shift.

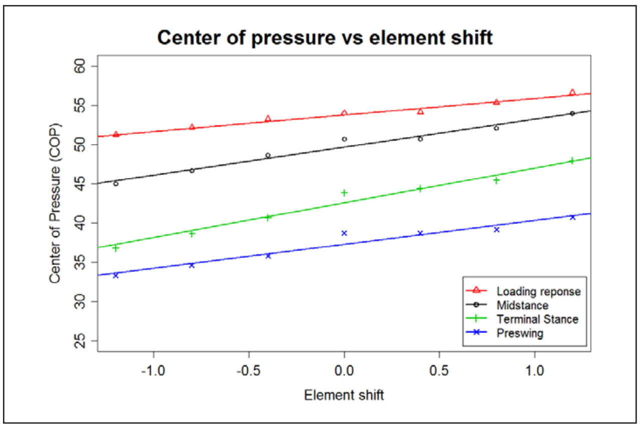

Figure 4. Linear regression plots of the mean COP coordinate versus the shift of the elements at loading response, midstance, terminal stance, and preswing. Signs represent the empirical means of the data, and the solid lines represent the linear models.

may be due to the fact that, in these stages of stance, the foot is in single limb support, thus any change in the foot position would have a more profound impact on the COP.

Kirby reviewed the interrelationships between foot and lower extremity function and mechanically based pathology of the foot and lower extremity, with an emphasis on the subtalar joint.[9] Thorough analysis of walking biomechanics clearly demonstrated that, without normal foot function, the rest of the lower extremity and body cannot function normally during walking. Because the GRF vector must first go through the foot on its way through the kinetic chain, the foot-ground interface may be expected to have a substantial effect on the lower extremity.[12]

The effect of the wedge position on different joints and segments has been widely explored. In a previous study by Eslami et al,[5] it was suggested that a wedge position could change the orientation of the subtalar joint axis of motion, thus playing a major role in the angle variability of different joints and segments. Lateral wedge insoles, for instance, are commonly used in clinical practice to counter the effect of elevated KAMs.[13,14,21] Previous studies have indicated that accurate management of knee moments can be attained by controlled shifts of the COP.[7,16] Our results suggest a positive association between the extent of the biomechanical manipulation (ie, the distance of element shift in the coronal plane from the neutral axis) and the location of the COP during gait and, therefore, may be a major contribution to understanding knee as well as other lower limb joint loading. Future studies incorporating a full examination of subtalar and ankle joint motions are required to quantify their relation to the observed correlation between the shift of the elements and the COP.

Several limitations arising from the current study should be noted. First, since we based our results only on in-shoe plantar pressure measurements, and we did not estimate any kinematics or kinetic measures. Nevertheless, since the GRF and COP were consistent for all patients, we assume that there were no dramatic changes in the kinematic or kinetic data. However, further studies incorporating full gait analysis could provide greater validity to our speculations. Second, the biomechanical testing was performed shortly after the device was first used by the participants. Continued usage of such an apparatus may lead to substantial gait adaptations and could influence the outcome of these interventions. Another limitation of this study was the employment of the apparatus at neutral position as control. This setting was selected to assure consistency of the kinematic model. Finally, it should be noted that the current study focused on a unique group (ie, healthy, young male adults). Therefore, these results are valid only for individuals with characteristics similar to those of the study cohort. Different populations (eg, females who tend to have different lower extremity joint motions than males) may respond differently to such interventions. Further studies are needed before these findings can be validated in other populations.

In conclusion, the current study indicates that external biomechanical manipulations enable controlled manipulation of the location of the COP. These results confirm the hypothesis of a direct association between manipulation of the biomechanical apparatus via translation of elements in the coronal plane (ie, from medial to lateral) and the direct shift of the COP trajectory, and offer a new understanding of lower limb biomechanics that may have implications in the field of biomechanical apparatus design and practice.

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Declaration of Conflicting Interests

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