Gait & Posture 40 (2014) 719–722

Contents lists available at ScienceDirect

Gait & Posture

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

Short Communication

Center of pressure trajectory during gait: A comparison of four foot positions

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ARTICLE INFO

Article history:
Received 16 May 2013
Received in revised form 20 May 2014
Accepted 1 July 2014

Keywords:
Center of pressure
Gait analysis
Foot pathology
Regression

ABSTRACT

Knowledge of the center of pressure (COP) trajectory during stance can elucidate possible foot pathology, provide comparative effectiveness of foot orthotics, and allow for appropriate calculation of balance control and joint kinetics during gait. Therefore, the goal of this study was to investigate the COP movement when walking at self-selected speeds with plantigrade, equinus, inverted, and everted foot positions. A total of 13 healthy subjects were asked to walk barefoot across an 8-m walkway with embedded force plates. The COP was computed for each stance limb using the ground reaction forces and moments collected from three force plates. Results demonstrated that the COP excursion was 83% of the foot length and 27% of the foot width in the anterior–posterior and medial lateral directions for plantigrade walking, respectively. Regression equations explained 94% and 44% of the anterior–posterior and medial–lateral COP variability during plantigrade walking, respectively. While the range of motion and COP velocity were similar for inverted and everted walking, the COP remained on the lateral and medial aspects of the foot for these two walking conditions, respectively. A reduced anterior–posterior COP range of motion and velocity were demonstrated during equinus walking. Ankle joint motion in the frontal and sagittal planes supported this COP movement, with increased inversion and plantar flexion demonstrated during inverted and equinus conditions, respectively. Results from this study demonstrated the COP kinematics during simulated pathological gait conditions, with the COP trajectory providing an additional tool for the evaluation of patients with pathology.

1. Introduction

The center of pressure (COP) movement has been identified as a measure of neuromuscular control during posture and gait. Defined as the centroid of all the external forces acting on the plantar surface of the foot, the COP movement has further been used to identify balance control, foot function, and treatment efficacy [1,2]. The COP velocity has additionally been shown to be a reliable measure of gait efficiency, with its clinical usefulness hypothesized for patients with hallux limitus or rigidus, metatarsalgia, hallux abducto valgus, or lower-limb amputation [3]. Among patients with hallux valgus and metatarsalgia an increased COP velocity was previously demonstrated during gait, when compared to normal feet [4].

While studies have demonstrated the efficacy of using both plantar pressure devices and force plates to record COP [1,5,6], with normative COP trajectories and velocities determined during walking [3] and running [7], no investigations have demonstrated the differences in COP kinematics during various gait conditions. Therefore, the purpose of this study was to investigate the COP movement when walking under normal and modified gait conditions. We hypothesized that the COP range of motion (ROM) would be greatest during plantigrade gait, with reduced COP movement and increased COP velocity demonstrated during simulated pathological gait.

2. Methods

A total of 13 healthy young adults (8 females, age 25.1 $\pm$ 2.9 years), were asked to walk barefoot across an 8-m walkway using four different foot conditions: (1) plantigrade; (2) equinus; (3) inverted; and (4) everted. During equinus, inverted, and everted walking, subjects ambulated on their toes, lateral borders of their feet, and medial borders of their feet, respectively, in order to simulate walking with pathology. All participants provided written informed consent

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http://dx.doi.org/10.1016/j.gaitpost.2014.07.001
0966-6362/

prior to involvement in the study. The study protocol was approved by the Mayo Clinic Institutional Review Board.

Three-dimensional trajectories of 12 reflective markers bilaterally placed on the feet (calcaneus, midpoint of the 2nd and 3rd metatarsal–phalangeal joint, 1st proximal metatarsal, 1st distal metatarsal, 5th proximal metatarsal, and 5th distal metatarsal) and eight reflective markers bilaterally placed on the shank (lateral malleolus, medial malleolus, lateral epicondyle and midpoint of the lateral epicondyle, and lateral malleolus) were collected using a 10-camera motion analysis system (Motion Analysis Inc., Santa Rosa, CA). Ground reaction forces and moments were collected from three force plates (AMTI Inc., Watertown, MA and Kistler Inc., Amherst, NY). Kinematic and kinetic data was collected at 120 Hz and 720 Hz, respectively. Foot anthropometrics collected included navicular height, foot length, and foot width.

The COP was computed for each limb throughout stance from the measured ground reaction forces and moments. The COP was converted into the foot coordinate system, with data normalized in the anterior–posterior and medial–lateral direction based on the foot length and foot width, respectively. The COP velocity was calculated using the Savitzky–Golay least squares method of differentiation, with the polynomial order set to 5 and the window length set to 11 [8]. Ankle joint kinematics were calculated using a *y–x–z* Cardan sequence, where *x* represents the anterior–posterior axis, *y* the medial–lateral axis, and *z* the superior–inferior axis. The arch index was calculated as the ratio of navicular height and the distance from the calcaneus to the first metatarsal–phalangeal joint during single-leg quiet standing [9]. All computations were performed using custom MATLAB programs (MathWorks Inc., Natick, MA).

Differences in the COP ROM and velocity (VEL) across walking conditions were evaluated using a one-way ANOVA. A comparison of loaded foot arch index and the COP ROM was performed using a Spearman's rank correlation in order to investigate COP changes due to foot anthropometrics. A forward stepwise linear regression was performed to estimate the medial–lateral and anterior–posterior COP during all walking conditions (IBM SPSS Inc., Chicago, IL). An increasing order polynomial of percent stance was inserted or removed from the regression equation if the *F*-test was less than 0.05 or greater than 0.10, respectively.

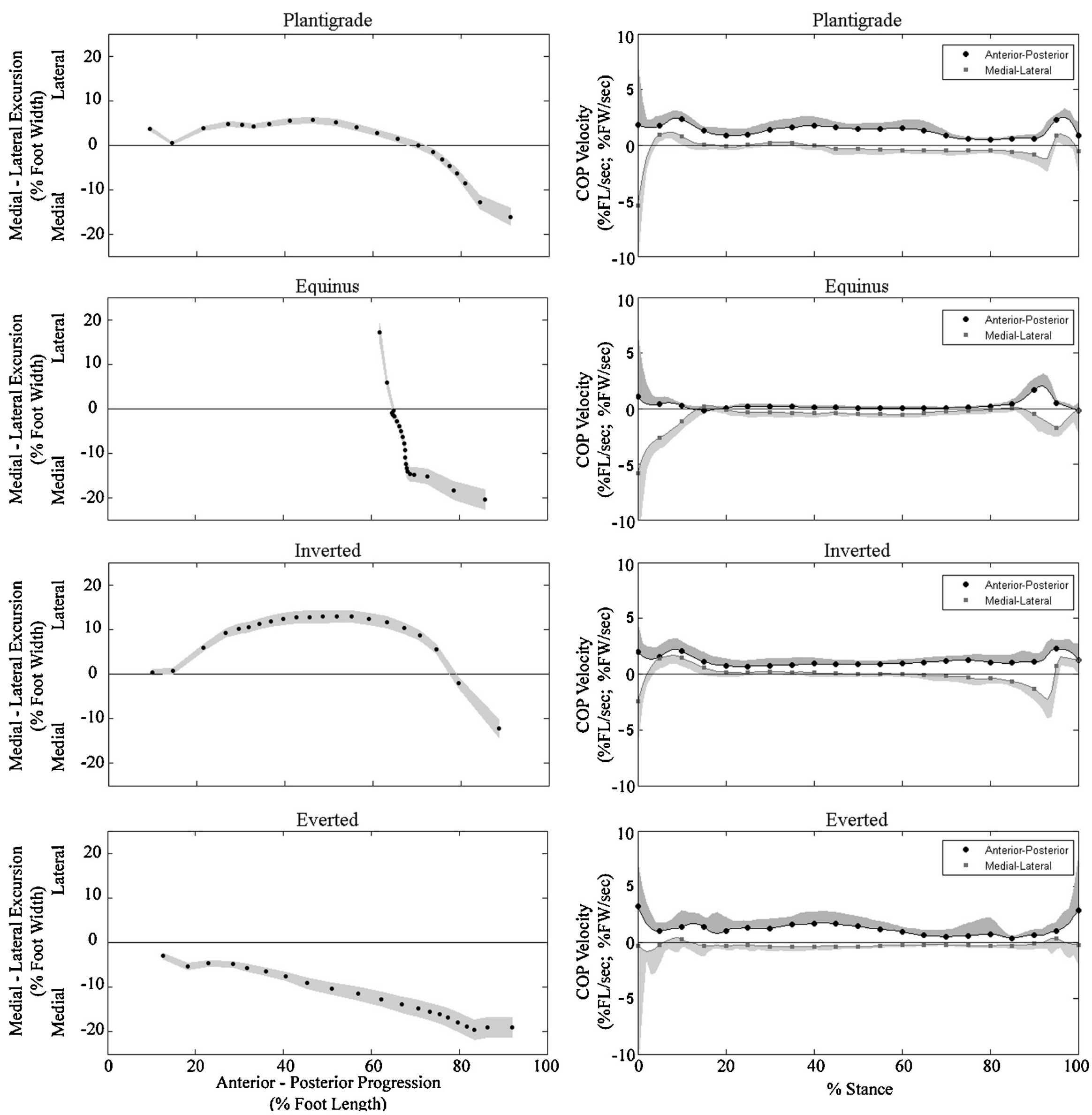

Fig. 1. Average (±1 SD) excursion of the COP displacement and velocity over the four walking conditions. Each point of the COP displacement and velocity represents 5% of stance.

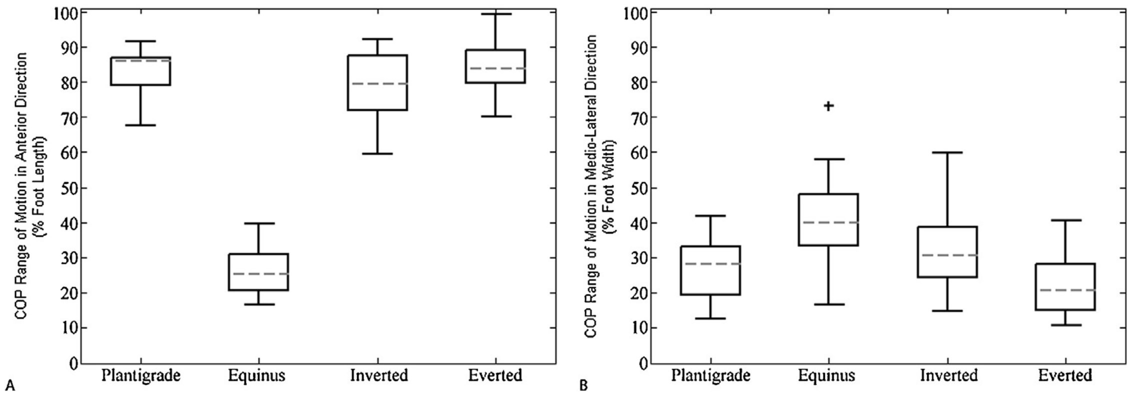

Fig. 2. COP range of motion across four conditions. The central dashed line represents the median, the box edges are the 25th and 75th percentiles, and the whiskers extend to ±2.7 SD. Outliers are labeled as +.

3. Results

A total of 26 feet were evaluated across all four walking conditions, with the COP traversing the forefoot, lateral boundary, or medial boundary of the foot during equinus, inverted, and everted walking, respectively. On average, the COP remained along the midline, the lateral portion, and the medial portion of the foot, during midstance for plantigrade, inverted, and everted walking, respectively (Fig. 1). Toward the end of stance, the COP progressed to the 1st metatarsal–phalangeal joint for all conditions. The COP demonstrated a relatively large medially directed velocity at heel strike and toe-off, with a constant anteriorly directed velocity throughout stance for plantigrade, inverted, and everted walking.

Participants demonstrated a significant condition effect for both the anterior–posterior and medial lateral COP ROM (Fig. 2; $P < 0.001$). The equinus walking condition demonstrated differences from all other conditions ($P < 0.001$), with the COP traversing across approximately 26.1% (±6.8%) of the foot length and 41.2% (±12.8%) of the foot width. The inverted walking condition also demonstrated a 10% greater COP ROM in the medial lateral direction than everted walking ($P = 0.036$).

Investigation of ankle joint angles across the four gait conditions revealed kinematic differences in all three planes. In the sagittal plane, equinus gait demonstrated on average 15 degrees of increased plantar flexion during stance (Fig. 3; $P < 0.001$). In the frontal plane, inverted and everted gait demonstrated approximately 9.6 and 0.7 degree of inversion, respectively, compared to the 4.5 degrees of inversion during plantigrade and equinus gait ($P < 0.001$).

The variability explained for the COP position in the anterior–posterior direction was greater than 80% for plantigrade, inverted, and everted walking (Table 1). Reduced R^2 values were revealed for the anterior–posterior COP movement during equinus walking and all four medial–lateral COP movements.

Over the 26 feet tested, the arch index ranged from 18.6 to 27.5 (average (SD) of 22.6 (2.6)). The arch index was not correlated with the COP ROM for any measure except for the anterior posterior COP ROM during inverted walking ($r = 0.564$; $P = 0.007$).

4. Discussion

The goal of this study was to describe the COP movement across the plantar foot surface during plantigrade, equinus, inverted, and everted walking. While previous research has modeled the COP progression linearly [10], results of a regression analysis reveal

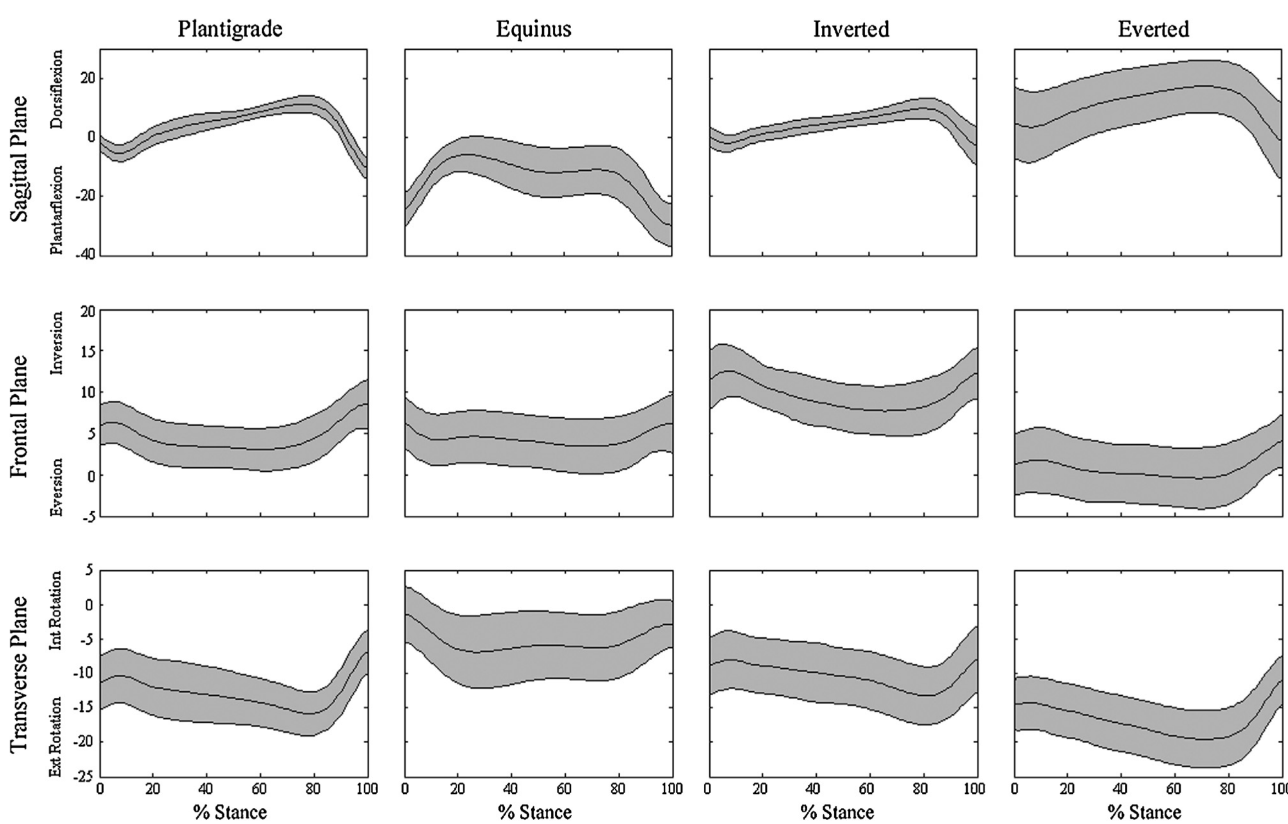

Fig. 3. Average (±1 SD) ankle kinematics during stance over the four walking conditions.

Table 1
Regression results for COP movement in both the anterior–posterior and medial–lateral directions under four foot positions. T indicates the percent of stance. All coefficients included are significant at the level of $P < 0.05$.

	Equation	R^2
Plantigrade		
Anterior–posterior	$0.093 + 1.093 \times T - 0.300 \times T^2$	0.938
Medial–lateral	$0.294 \times T - 0.455 \times T^2$	0.443
Equinus		
Anterior–posterior	$0.611 + 0.436 \times T - 1.092 \times T^2 - 0.863 \times T^3$	0.273
Medial–lateral	$0.091 - 0.620 \times T + 0.731 \times T^2 - 0.401 \times T^3$	0.387
Inverted		
Anterior–posterior	$0.113 + 1.033 \times T - 1.060 \times T^2 + 0.783 \times T^3$	0.875
Medial–lateral	$0.009 + 0.382 \times T - 0.448 \times T^3$	0.331
Everted		
Anterior–posterior	$0.122 + 1.039 \times T - 0.284 \times T^3$	0.797
Medial–lateral	$-0.031 - 0.179 \times T$	0.250

that the COP trajectory is a more complicated movement, involving at least polynomial expressions of the stance time. While there was a reduced amount of variance explained for the COP trajectory in the medial–lateral direction for all foot conditions, among everted, inverted, and plantigrade walking, up to a third order polynomial expression of stance explained greater than 80% of the COP movement in the anterior–posterior direction. In contrast to our hypothesis, the greatest medial–lateral COP ROM was demonstrated during the equinus condition, with similar COP ROM and COP velocity demonstrated during plantigrade, inverted, and everted gait.

The average trajectory of the COP during plantigrade walking is similar to results previously demonstrated using both force plates and foot pressure systems [1,4]. While a similar pattern of movement is seen during inverted walking, the COP is placed approximately 5% of the foot width further laterally during midstance, when compared to plantigrade walking. Alternatively, during everted walking, the COP was always on the medial portion of the foot. Clinically, reducing eversion of the foot during gait can result in reductions in frontal and transverse plane knee joint motion [11]. While orthotic decisions are often based on measurements taken during static posture [11], understanding of the dynamic COP trajectory can provide additional information related to lower extremity moments and balance control during gait [12,13]. The use of a biomechanical device to manipulate the COP in the medial and lateral direction has previously demonstrated significant correlations between the COP shift and the location of the device [14]. The placement of such devices allowed for direct control of the COP, which could have implications for implant design and practice [14].

The COP velocity has previously been shown to range from 16 cm/s to 27 cm/s and 19 cm/s to 24 cm/s for men and women, respectively, from loading response to preswing when wearing flat heeled shoes [6]. While similar results were seen in the current study for participants during midstance, large COP velocities were demonstrated at heel strike and terminal stance across all walking conditions. Differences might be indicative of the effect from shoe insoles on the COP.

Results of the COP ROM are similar to those reported previously, in which the COP displacement corresponded to 83% of the foot contact length and 18% of the foot contact width [6]. The COP ROM further demonstrated up to 10% differences in the medial lateral direction across gait conditions. Interestingly, the arch index did not affect the COP ROM, with the COP trajectories demonstrating small standard deviations across all gait conditions. These results are in contrast to subjects with a higher arch demonstrating a reduced medial–lateral COP ROM during running [7]. Though a limitation of this study is the lack of individuals with foot instability, results of this study provide representative trajectories for the COP among healthy individuals during simulated pathological gait. Future work can hopefully utilize these normative values in the potential diagnosis and evaluation of patients with pathology.

Acknowledgement

Funding was provided by NIH T32 HD07447.

Conflict of interest

The authors have no conflict of interest.

References

[1] Chesnin KJ, Selby-Silverstein L, Besser MP. Comparison of an in-shoe pressure measurement device to a force plate: concurrent validity of center of pressure measurements. Gait Posture 2000;12(2):128–33.
[2] Scherer P, Sobiesk G. The center of pressure index in the evaluation of foot orthoses in shoes. Clin Podiatr Med Surg 1994;11(2):355.
[3] Cornwall M, McPoil T. Velocity of the center of pressure during walking. J Am Podiatr Med Assoc 2000;90(7):334–8.
[4] Grundy M, Tosh P, McLeish R, Smidt L. An investigation of the centres of pressure under the foot while walking. J Bone Joint Surg (Br) 1975;57(1):98–103.
[5] Chevalier TL, Hodgins H, Chockalingam N. Plantar pressure measurements using an in-shoe system and a pressure platform: a comparison. Gait Posture 2010;31(3):397–9.
[6] Han TR, Paik NJ, Im MS. Quantification of the path of center of pressure (COP) using an F-scan in-shoe transducer. Gait Posture 1999;10(3):248–54.
[7] De Cock A, Vanrenterghem J, Willems T, Witvrouw E, De Clercq D. The trajectory of the centre of pressure during barefoot running as a potential measure for foot function. Gait Posture 2008;27(4):669–75.
[8] Savitzky A, Golay MJ. Smoothing and differentiation of data by simplified least squares procedures. Anal Chem 1964;36(8):1627–39.
[9] Kaufman KR, Brodine SK, Shaffer RA, Johnson CW, Cullison TR. The effect of foot structure and range of motion on musculoskeletal overuse injuries. Am J Sports Med 1999;27(5):585–93.
[10] Rosenblatt NJ, Grabiner MD. Measures of frontal plane stability during treadmill and overground walking. Gait Posture 2010;31(3):380–4.
[11] Eng JJ, Pierrynowski MR. The effect of soft foot orthotics on three-dimensional lower-limb kinematics during walking and running. Phys Ther 1994;74(9):836–44.
[12] McCaw ST, DeVita P. Errors in alignment of center of pressure and foot coordinates affect predicted lower extremity torques. J Biomech 1995;28(8):985–8.
[13] Lee H-J, Chou L-S. Detection of gait instability using the center of mass and center of pressure inclination angles. Arch Phys Med Rehabil 2006;87(4):569–75.
[14] Khoury M, Wolf A, Debbi EM, Herman A, Haim A. Foot center of pressure trajectory alteration by biomechanical manipulation of shoe design. Foot Ankle Int 2013;34(4):593–8.