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In-shoe center of pressure: Indirect force plate vs. direct insole measurement

Eytan M. Debbi [a,*], Alon Wolf [a], Yulia Goryachev [a], Ziva Yizhar [b], Elhanan Luger [c], Ronen Debi [d], Amir Haim [a,c]

[a] Biorobotics and Biomechanics Lab, Faculty of Mechanical Engineering, Technion-Israel Institute of Technology, Haifa 32000, Israel
[b] Department of Physical Therapy, Faculty of Medicine, Tel Aviv University, Tel Aviv, Israel
[c] Department of Orthopedic Surgery, Sourasky Medical Center, Tel Aviv, Israel
[d] Department of Orthopedic Surgery, Barzilai Medical Center, Ashkelon, Israel

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ABSTRACT

Background: In-shoe center of pressure (COP) measurement is essential in biomechanics. COP can be measured directly utilizing pressure-sensitive insoles, or calculated indirectly via force plate-generated data. While the latter does not require the use of additional measurement hardware (shoe insoles), its precision at calculating in-shoe COP has not been determined. Our purpose was to ascertain the precision of force plate in-shoe COP calculations and enhance their accuracy through a mathematical algorithm.
Methods: Twelve male students participated in the study. In-shoe COP was measured synchronously via the Pedar-X insole system and AMTI force plates, comparing the measurements of both systems. A mathematical algorithm was created to improve agreement between the systems and comparisons were recalculated.
Results: The two methods showed different measurements of in-shoe COP. The medio-lateral (ML) and anterior-posterior (AP) Pearson correlation coefficients between systems were 0.44 ± 0.35 and 0.99 ± 0.01, and the ML and AP RMS errors were 6.3 ± 3.0 mm and 43.0 ± 12.5 mm, respectively. Using a mathematical algorithm, the differences between the measurements of each system could be reduced significantly (all $P < 0.001$).
Conclusions: Without adjustment, force plates give an approximate location of the COP. Using an adjustment model greatly improves the accuracy of the COP trajectory during stance.

1. Introduction

The foot center of pressure (COP) is a virtual site about the foot's plantar surface that is the average location of all pressures acting on the foot at any given time [1,2]. During the stance phase of gait, the COP is located at the medial aspect of the hind foot during initial contact and moves anteriorly, curving laterally at mid-stance and progressing medially to the first two metatarsal heads at terminal stance [2–4]. 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 [3,5–8]. The location of the COP is a critical measure of gait and reflects balance [2,3], the severity of lower limb diseases [3,4] and the actions of different treatment modalities on the lower limb [5,6].

Substantial biomechanical research has been focused on the effect of footwear-generated biomechanical interventions (e.g. foot orthoses, insoles, shoes which generate alterations in kinetic patterns), which have necessitated an accurate assessment of in-shoe COP. When wearing shoes, there is a unique COP between the foot and the shoe sole and between the shoe sole and the floor. The instantaneous in-shoe COP is the virtual locus of the average forces acting on the foot at any given time [7]. Currently, the in-shoe COP is measured directly using insole measuring systems such as the Pedar-X system (Novel Electronics, St. Paul MN, USA) [8–10]. These systems consist of insoles fitted within a subject's shoes that are able to accurately and precisely measure the real-time location and magnitude of the pressures between the foot and shoe. These directly record the pressure on the foot using a matrix of pressure sensors embedded within the insole and calculate the distribution of pressure over time. While these systems give relatively good measurements of the center of force acting on the foot, they have several drawbacks. Firstly, such measuring devises are not frequently used in routine gait studies. Secondly, these systems mandate the application of a transmitter or other sensors that may interfere with normal locomotion. In addition, although the insoles are not bulky, they do add additional material to the subject's feet. Inserting them within a subject's shoe may alter the biomechanical properties of the shoe and the natural gait patterns of the subject. Lastly, the premade insoles do not always match every type of shoe.

Alternatively, the in-shoe COP can be calculated indirectly using force plates that measure the location and magnitude of the ground reaction force (GRF), which is the average vector exerted by the

* Corresponding author. Tel.: +972 4 8292087; fax: +972 4 8295711.
E-mail addresses: eytan.debbi@aya.yale.edu, edebbi@technion.ac.il (E.M. Debbi).

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ground on the body (between the shoe and the floor) [3]. Each of the four corners of the force plates are equipped with three piezoelectric sensors oriented orthogonally, thereby enabling calculation of the location and orientation of the GRF vector, which is the hypothetical mean force vector generated by the ground on the body. When walking barefoot, the locus of COP is where the GRF crosses the force plates. In order to calculate the COP, we have to calculate the location of the GRF vector in the plane of the sole of the foot (fpCOP). By definition, the projection of the fpCOP on the foot and the trajectory of the in-shoe COP measured by the Pedar-X during the single limb support of gait should overlap [1,9].

Chesnin et al. compared the in-shoe Parotec System (London Orthotics, London UK) to an Advanced Medical Technology Inc. (AMTI) force plate (AMTI, Watertown MA, USA). Between the two systems, they found a 5.6 ± 3 mm RMS error in the ML direction with a correlation greater than 0.7 and a 13.7 ± 5.9 mm RMS error in the anterior–posterior (AP) direction with a correlation greater than 0.9 [7]. The results of this study suggest that differences exist between the systems and that the linear relation between the systems is greater in the AP direction than in the ML direction. The study is also limited by several methodological problems that are claimed to have led to significant errors [7]. In addition, the study does not attempt to improve the approximation of the fpCOP to the in-shoe COP.

Our group recently described an algorithm utilizing force plate data to indirectly evaluate in-shoe COP [3,11]. In brief, the instantaneous coordinates of the fpCOP recorded by the force plate were extracted in conjunction with matching instantaneous coordinates of the heel and toe markers (defining a fixed sagittal axis reference in the foot segment), and the instantaneous distances in the horizontal and sagittal plane from the foot axis to the in-shoe COP were then calculated. However, the accuracy of this algorithm has not been tested.

The goal of the present study was to compare indirect force plate-based fpCOP calculations to the simultaneous direct in-shoe COP measured by the Pedar-X system. The study was designed to test the hypothesis that significant differences exist between the force plate and Pedar-X systems and that a properly designed mathematical algorithm can improve the ability of the force plate to predict in-shoe COP. If this is possible, the mathematical algorithm can be implemented in future studies that use the force plate instead of the Pedar-X to determine in-shoe COP location.

2. Methods

2.1. Subjects

Twelve healthy male students (20–30 years old) with an average ± SD BMI of 23.2 ± 3.2 kg/m^2 and height of 177.3 ± 5.0 cm volunteered for the study. All subjects had an equivalent shoe size. The subjects had no medical conditions related to their lower limbs or to balance. The study was approved by the institutional ethics committee and all subjects gave written informed consent prior to participating in the study.

2.2. Instrumentation

The Pedar-X System (Novel Electronics, St. Paul MN, USA) consisted of two in-shoe insoles that measure pressure. The insoles were 2.5 mm wide and contained 99 pressure-sensitive capacitor sensors. The sensors sampled at a rate of 100 Hz and created a compound pressure–time data series that provided the centroid of loading at each point in time. The system gathers and analyzes large amounts of quantitative data that are related to dynamic foot pressure. Each insole was connected to a unit attached to the subject's waist with a belt. This unit wirelessly transmitted the data to the lab's computer housing the Pedar-X software. The Pedar-X software calculated the instantaneous location of the in-shoe COP throughout the stance phase of gait. Prior to analysis, the insoles were calibrated according to the manufacturer's instructions.

Two AMTI OR6-7-1000 force plates (AMTI, Watertown MA, USA) measured the instantaneous fpCOP of each foot at a sampling rate of 960 Hz. Six passive reflective markers were placed on the subjects' shoes: (1) on the second metatarsal head; (2) at the heel; (3) medial to first metatarsal head; (4) medial side under medial malleolus; (5) lateral to fifth metatarsal head; (6) lateral side under lateral malleolus. All markers were placed on the same plane (i.e. at the same vertical distance from the ground during quiet standing). The six markers create an elliptical outline of the foot. Only the toe and heel markers were used for analysis, while the other makers served as redundant markers for visualization and orientation. Their location was acquired by a multi-camera infrared Vicon motion analysis system (Oxford Metrics Ltd., Oxford, UK) at a sampling rate of 120 Hz. The plates were placed in tandem along a 10 m walkway around which were placed the eight infrared cameras. The force plate and marker position data were transferred to the same computer running the Vicon Nexus Software (version 1.5.2). The force plates and the Vicon systems were synchronized in time and space. The Pedar and force plate systems acquired data simultaneously during each trial. Data acquisition was started at the same time.

2.3. Data collection

Subjects were measured for anthropometric data and fitted with sneakers that were identical for all subjects. Appropriately sized Pedar-X insoles were inserted between the subjects' socks and soles of the shoes. It was verified that the insoles of the Pedar did not fold when the subject inserted his foot.

The subjects were instructed to walk several times along a 10 m walkway containing the two AMTI force plates at a self-selected speed. Once prepared for each trial, the participant was asked to carry out several walks before data acquisition. Once the subject reported being comfortable, he was asked to carry out the trial walks at approximately the same pace. Data were collected simultaneously by the Pedar-X, force plate and Vicon systems. Only steps on the force plates were saved for analysis. Six trials per subject were saved and exported to MATLAB™ for analysis.

2.4. Data analysis

A schema of the analyses is presented in Fig. 1. When the patient walked on the force plates, data was acquired simultaneously from the Pedar-X system and the force plates (Fig. 2a). The Pedar-X software records the location of the in-shoe COP as *X*–*Y* coordinates relative to an origin located at the most medial and posterior points of the insole (Fig. 2b). The in-shoe COP was calculated as a graph of sampled points throughout the stance phase of gait (Fig. 2b).

2.5. Manual adjustment

The force plate measures the fpCOP as *X*–*Y* coordinates relative to a predetermined lab origin. To place these coordinates within the frame of the subject's shoe, a new origin was needed. For this, the instantaneous coordinates of the shoe markers were used to define fixed axes of the foot. The line between the toe and heel markers was considered the *y*-axis of the foot. The horizontal (*x* axis) was defined as a perpendicular line to the *y*-axis that intersects at the heel marker (new origin) (Fig. 2c). Here, too, the fpCOP was calculated as a graph of sampled points throughout the stance phase of gait (Fig. 2c).

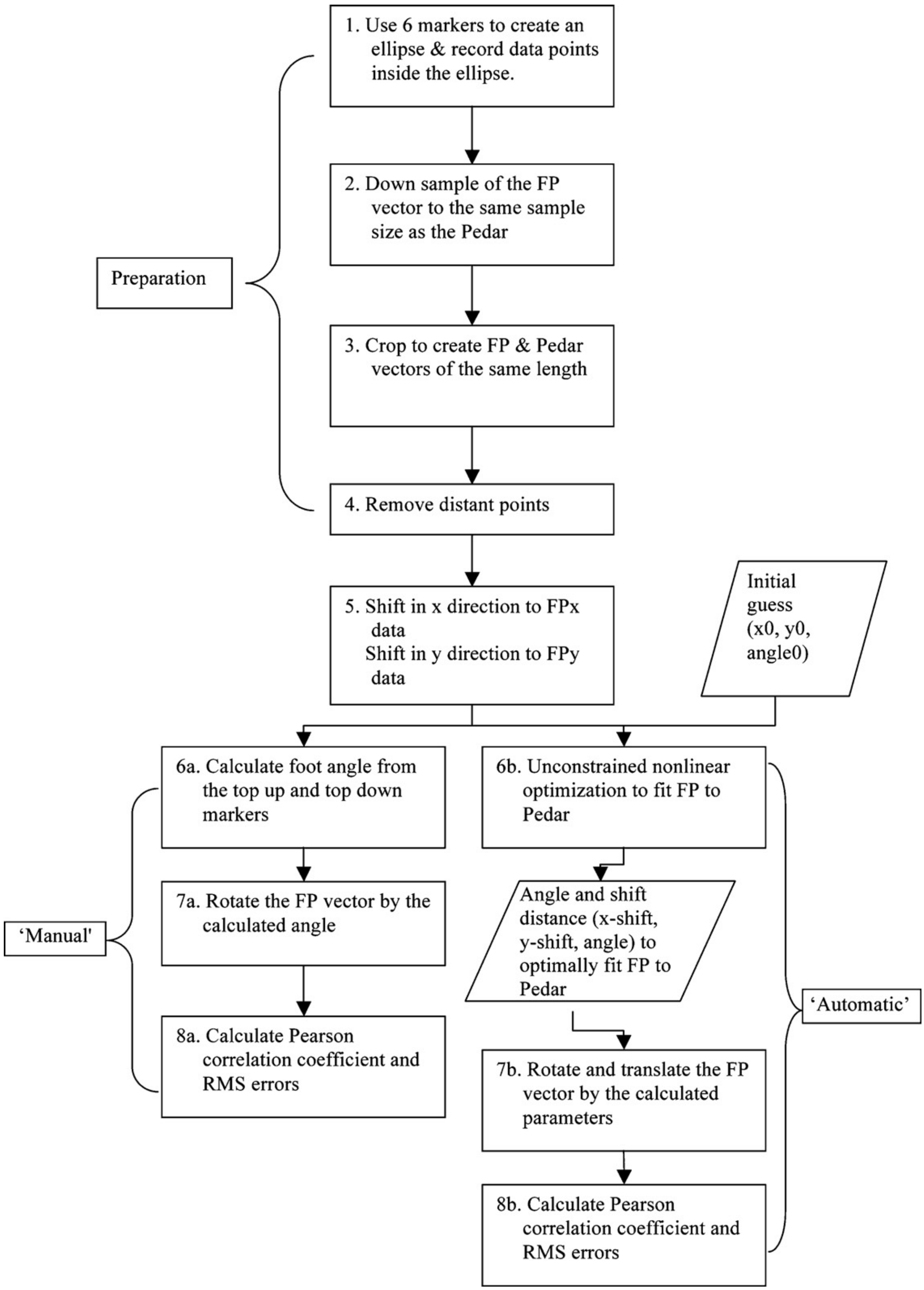

Fig. 1. Schematic diagram of study analyses. The ellipse refers to the outline of the foot using the six reflective markers. FP refers to force plate.

The origins of the Pedar-X and the force plate were then superimposed (Fig. 2d). The *y*-axis of the foot model of the markers was rotated (angle = α) with the graph of the fpCOP, so that it was superimposed with the *y*-axis of the Pedar-X (Fig. 2d). The *y*-axis of the foot model from the force plate was then translated 43.0 mm to align with the posterior-most part of the shoe (Fig. 2e) (43.0 mm is half the width of the shoe at the heel). Graphs for the right foot were shifted 43.0 mm to the right and graphs for the left foot were shifted 43.0 mm to the left. At this point, COP graphs from the two systems were as equated as much as possible (Fig. 2e). Angles refer to the rotation of the fpCOP and translations refer to the distance by which the fpCOP was shifted. External rotation was considered positive and lateral translation was considered positive.

Fig. 3 presents the actual graphical data and adjustments made to the COP graphs of one individual. After this adjustment, a small

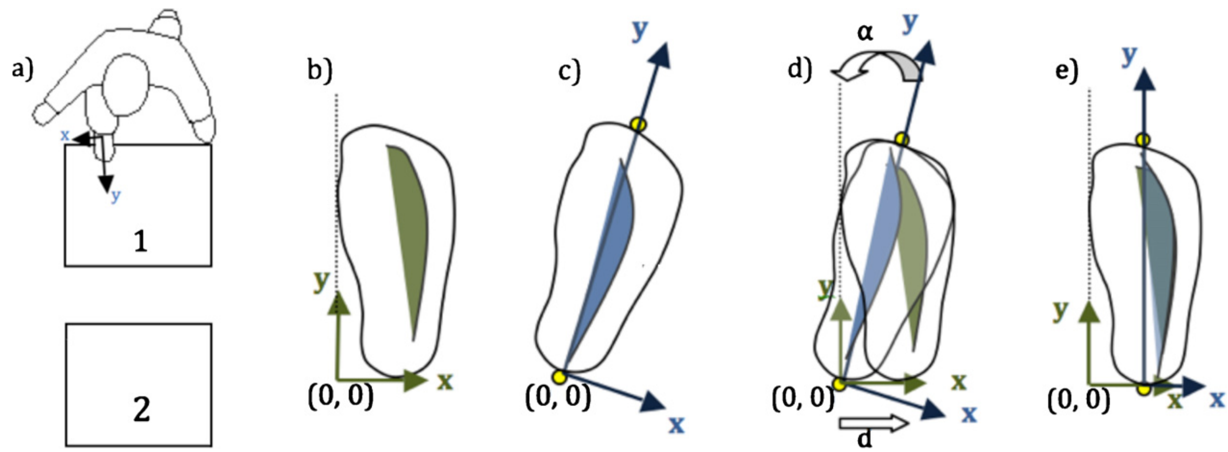

Fig. 2. Manual alignment of the fpCOP from the AMTI force plate and in-shoe COP from the Pedar-X sytem. All images are drawn with the viewer's perspective looking from above downward toward the subject walking. (a) Subject walking on force plates. (b) Pedar sole: origin is marked and the in-shoe COP over time is plotted in green. (c) Force plate COP: origin is marked and the fpCOP over time is plotted in blue. The anterior and posterior reflective markers are denoted in yellow. (d) Alignment: the Pedar and shoe origins are placed at same point. The force plate COP is rotated by an angle of α and translated by d. (e) Superimposition of graphs. (For interpretation of the references to color in this figure legend, the reader is referred to the web version of the article.)

difference in angle and location remained between the COP graphs of both systems (Fig. 3c).

Pearson correlation coefficients (PCCs) were calculated to compare the COP changes in the ML and AP directions throughout stance calculated by the Pedar and force plate. In addition, the differences in COP displacements in the ML and AP directions were compared between the two systems, using root means square (RMS) error. Data from one trial for two subjects were unusable - there were too many sampling data artifacts that contaminated these trials and, therefore, they could not be analyzed. This left 70 trials for the right foot, 70 trials for the left foot and 140 trials in total.

2.6. Mathematical algorithm

A mathematical algorithm was then constructed to match the fpCOP graph of the force plate to the in-shoe COP graph of the Pedar-X. The algorithm finds the homogeneous transformation (i.e. rotation and translation) of the COP of the two systems. The parameters of the homogeneous transformation matrix are solved by a minimization process for which the magnitude of the adjustments – fpCOP coordinates adjusted to that of the Pedar-X – serves as the objective function. The initial adjustment was defined as (x0, y0, angle0)-X distance = 0, Y distance = 0, angle0 = −pi/4.

We used the Matlab function 'fminunc' with the original parameters, the parameters defined by the Matlab function 'optimset' (medium scale, 1000 iterations), the initial parameters as described and the least squares minimization function to obtained the optimal fit (X-shift, Y-shift, angle) between vectors. The following transformation matrix was then used to align the fpCOP with the Pedar-X in-shoe COP (planar case):

$$^{\text{pedarx}}T_{\text{forceplate}} = \begin{bmatrix} \cos(\text{angle}) & \sin(\text{angle}) & x\,\text{shift} \\ -\sin(\text{angle}) & \cos(\text{angle}) & y\,\text{shift} \\ 0 & 0 & 1 \end{bmatrix}$$

Once the parameters of the transformation matrix T were revealed per subject, the two graphs were aligned again for each participant (Fig. 3d). The PCC and RMS errors were recalculated and compared to results after manual adjustment using a t-test. A Kolmogorov–Smirnov test was calculated to evaluate the distribution of each dataset. The results for the left and right feet before and after adjustment with the mathematical algorithm were also evaluated using a Kolmogorov–Smirnov test and compared using a t-test.

3. Results

The PCCs and RMS errors in the ML and AP directions before and after adjustment are presented in Table 1. Before adjustment, the RMS error was higher in the AP direction than in the ML direction. The PCC was greater in the AP direction than in the ML direction. Through the mathematical minimization process, the fpCOP was mathematically rotated an average of 0.16 ± 0.05 rad for the left foot and 0.25 ± 0.07 rad for the right foot, and translated an average distance of 12.7 ± 5.74 mm in the ML and -44.9 ± 11.0 in the AP directions for both feet. After this adjustment, the PCC in the ML direction significantly increased by 110% ($P < 0.001$) and the RMS error significantly decreased by 75% in the ML direction ($P < 0.001$) and by 86% in the AP direction ($P < 0.001$). The PCC in the AP direction remained high.

The results for the left and right feet differed significantly in the angle adjustments (right > left; $P = 0.002$ for manual adjustment and $P = 0.003$ using the mathematical algorithm). The right foot also had a greater ML RMS error before adjustment with the mathematical algorithm ($P = 0.027$). All other results from both feet were comparable before and after adjustment with the mathematical algorithm.

4. Discussion

The first portion of the present study compared the fpCOP measured by the force plate to the in-shoe COP location of the Pedar-X system. Before comparison, the fpCOP graph was rotated and translated to best fit the Pedar-X in-shoe COP graph. The magnitude of translation and rotation was based on marker orientation and size of shoe and Pedar-X insole. The ML fpCOP location measured by the force plate differed by an RMS error of 6.3 mm and the AP location differed by an RMS error of 43.0 mm from the Pedar-X in-shoe COP. The higher RMS error in the AP direction compared to the ML direction may be because the foot is longer in this direction. The trajectory of the COP on the foot during stance travels from the calcaneus during heel strike to the lateral ray of the foot during midstance and then finally travels beneath the first metatarsal head just before toe-off. Thus, the relative change in COP is significantly greater in the AP direction than in the ML direction. Accordingly, any recorded error in the AP direction would be significantly greater than in the ML direction. The AP PCC was high (PCC = 0.99) while the ML PCC was lower (PCC = 0.44).

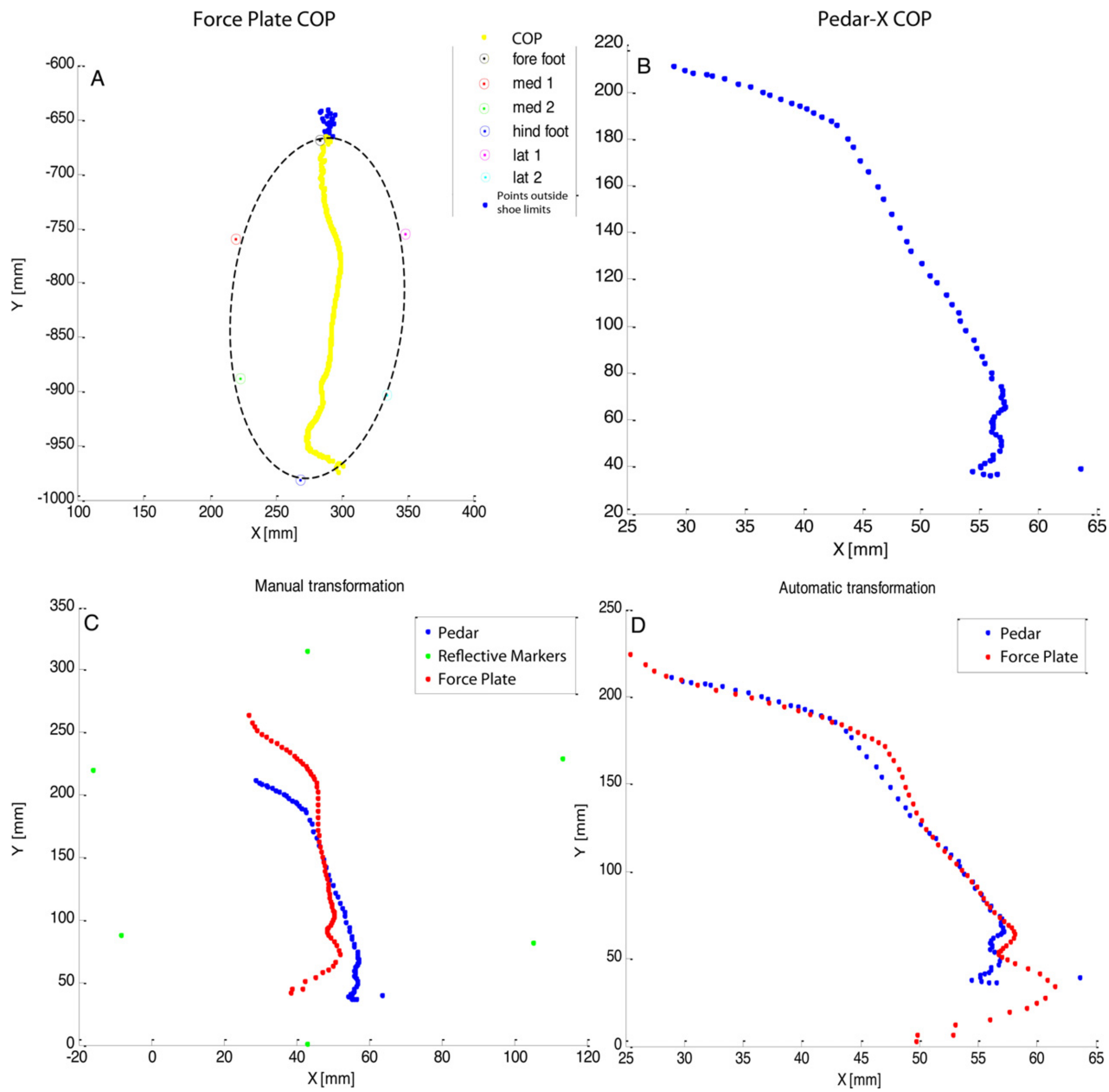

Fig. 3. Center of pressure (COP) graph from the force plate and from the Pedar-X System for the right foot of one individual. (A) The fpCOP graph from the force plate. med 1, med 2, lat 1 and lat 2 refer to the medial and lateral reflective markers of the foot, respectively. (B) The in-shoe COP graph from the Pedar-X System. (C) Superimposition of the fpCOP graph from the force plate onto the in-shoe COP graph of the Pedar-X system. The force plate graph was slightly rotated manually in order to align with the axes of the Pedar-X system. (D) Best fit of the fpCOP graph from the force plate onto the in-shoe COP graph of the Pedar-X system. The force plate graph was adjusted by a mathematical algorithm via rotation and translation in the *X* and *Y* directions in order to best fit the Pedar-X graph.

These results are similar to those obtained in a previous study by Chesnin et al. using different insole systems [7]. After axial alignment, they found a 5.6 ± 3 mm RMS error in the ML direction with a PCC greater than 0.7 and a 13.7 ± 5.9 mm RMS error in the AP direction with a PCC greater than 0.9. The ML RMS error obtained by Chesnin et al. was similar to that of the present study. On the other hand, the AP RMS error obtained by Chesnin et al. was significantly smaller than that obtained by the present study. This is most probably due to differences in methodology. Chesnin et al. performed their study with the insoles placed between two socks and not within shoes. The presence of the shoes in the current study may have added to the RMS error in the AP direction. As in the present study, Chesnin et al. also found higher correlations in the AP direction than in the ML direction. Chesnin et al. suggested that the low correlations found in the ML direction were due to methodological problems in their data collection. We found an even lower correlation in the ML direction in the present study. Chesin et al. suggested that their experimental design with two socks led to inherent instability between the insole and the foot and caused a high number of artifacts. In our study there was much less relative translation between the insole and the foot. Thus, it does not appear that our results are due to artifacts. Rather, it is more likely that the results we observed are true phenomena. Therefore, it is likely that a true low correlation exists between the ML changes measured by the force plate and the shoe insoles. This difference may be because the force plate takes vertical and shear forces into calculation while the Pedar-X only measures vertical forces. The higher correlation in the AP direction in both studies may be because this axis is along the primary line of progression during walking.

In the second portion of the study a mathematical algorithm was constructed to use the fpCOP measurements of the force plate to predict the in-shoe COP location of the Pedar-X. The

Table 1
Comparison of the medial-lateral (ML) and anterior-posterior (AP) location of the fpCOP and in-shoe COP.

Initial alignment of axes	Left foot (Average ± SD)	Right foot (Average ± SD)	Both feet (Average ± SD)
Angle adjustment (rad)	0.096 ± 0.051	0.18 ± 0.06	0.14 ± 0.07
Translation adjustment (mm)	43.0 ± 0.0	43.0 ± 0.0	43.0 ± 0.0
PCC ML	0.53 ± 0.31	0.35 ± 0.38	0.44 ± 0.35
PCC AP	0.99 ± 0.01	0.99 ± 0.01	0.99 ± 0.01
RMS ML (mm)	5.0 ± 1.8	7.6 ± 3.4	6.3 ± 3.0
RMS AP (mm)	39.7 ± 12.0	46.3 ± 12.7	43.0 ± 12.5

Adjustment with mathematical algorithm	Left foot (Average ± SD)	Right foot (Average ± SD)	Both feet (Average ± SD)	Comparison to initial alignment[a]	
				Mean difference [95% CI]	*P* value
PCC ML	0.93 ± 0.04	0.90 ± 0.12	0.91 ± 0.09	0.48 [0.63; 0.330]	<0.000[a]
PCC AP	0.99 ± 0.01	0.99 ± 0.01	0.99 ± 0.01	0.00018 [-0.0055; 0.0052]	0.947
RMS ML (mm)	1.3 ± 0.5	1.9 ± 1.9	1.6 ± 1.4	4.7 [3.4; 6.0]	<0.000[a]
RMS AP (mm)	6.0 ± 2.5	6.3 ± 3.2	6.1 ± 2.9	36.8 [31.6; 42.1]	<0.000[a]

Note: Angle adjustment = manual angle change (radians) needed to align the force plate and Pedar-X axes. Positive angles were considered external rotation. Translation adjustment = manual translation needed to align the force plate and Pedar-X axes. The force plate axes were translated by half the width of the Pedar insole (full width = 86 mm; half width = 43 mm). Lateral translation was considered positive. PCC = Pearson correlation coefficient. RMS error = root mean square error.

[a] Distributions were normal according to Kolmogorov–Smirnov tests. A *t*-test was used to compare means. Significance levels were set at $P < 0.05$.

program rotated and translated the fpCOP coordinates in the ML and AP directions. A significant reduction in RMS error was found in both the ML and AP directions. In addition, a large and significant improvement in correlation was found in the ML directions (new ML PCC = 0.91).

These results suggest that a mathematical algorithm can be used to define in-shoe COP using the fpCOP measurements of the force plate. However, an initial comparison between the force plate and an insole system, such as the Pedar-X, should be carried out to find the exact adjustment necessary for each gait lab. Using our mathematical algorithm would only give an approximate result. Future studies should apply the initial alignment or the mathematical algorithm to the force plate system in several contexts and determine how well they predict the results from the insole system. Additionally, a larger study with a more diverse population would be required to test the generalizability of the adjustment process.

The present study also compared the results for each foot. The amount of adjustment in rotation needed for the right foot was significantly greater than for the left both for initial alignment and the mathematical algorithm. These findings suggest that the difference in angle adjustment needed is a difference between the feet. In addition, the RMS error in the ML direction for the right foot before adjustment with the mathematical algorithm was 2.61 mm greater than for the left foot. The differences in angular adjustments between feet may be due to differences in the right and left Pedar-X insoles used or in marker placement. Despite this, the difference may be small enough (0.14 rad) to tolerate when analyzing the in-shoe COP from a force plate instead of an insole system. Nonetheless, it is recommended that any gait lab that intends to use force plates to measure in-shoe COP should adequately assess the rotational angle needed for each foot separately.

The difference between legs may also be due to an effect of a dominant leg. The present study did not record the dominant leg of each subject. Future studies should take note of subjects' dominant legs and compare the results between the Pedar-X and force plate for each leg to determine if differences exists between limbs. Aside from limb differences, future studies should also implement trials with different shoes to determine the accuracy of the force plates in different contexts.

Several limitations arising from the current study should be noted. Firstly, it is important to mention the possible effects of different shoes with soles made of different materials and created at different heights. Different soles may cause a variable distance between the foot and floor during the stance phase of gait. However, due to the nature of the algorithm that carries out the mathematical corrections, the reference points of the markers in the sagittal axis are placed on the shoe at the level of the sole of the foot (i.e. the transverse plane of the second metatarsal head and the calcaneal fat pad). The motion capture system tracks the exact location of these markers relevant to each other, continuously, and compares them in relation to the relative fpCOP position recorded by the force plates. Since the markers are placed above the true sole of the shoe, the markers, as well as the Pedar-X insoles, should not be affected by the changes in compressibility of the sole of the shoe. Nevertheless, since all testing in the current study was carried out with the same shoe, it is possible that different shoe materials could theoretically influence the accuracy of this algorithm. Further studies must be undertaken in order to validate this algorithm. In addition, any such studies should have the markers placed above the sole of the shoe used. If the shoe design does not allow for this type of placement, a bias may be introduced into the algorithm. Another important limitation may be in the repeatability of the algorithm in different research facilities. Every lab uses different machinery and models to analyze motion. This may introduce a bias into our proposed algorithm. If this is the case, we recommend that, before implementing this algorithm in clinical use, the lab should recalculate the algorithm in the new setting. Finally, the current study focused on a unique group (i.e. young subjects with similar anthropometric characteristics). These results are therefore applicable only for subjects with characteristics similar to those of the study cohort.

In conclusion, the current study provides information on the variation between direct and indirect methods of measuring in-shoe COP. Additionally, it offers a novel mathematical algorithm to adjust the indirect methods of in-shoe COP measurement (i.e. force plates) to the direct methods of in-shoe COP measurement. This data could be applicable for biomechanical studies as well as in clinical practice.

Conflict of interest statement

No author has any conflict of interest to declare.

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