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The influence of deformation height on estimating the center of pressure during level and cross-slope walking on sand

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ABSTRACT

Force plates are frequently used to collect the ground reaction forces (GRF) and center of pressure (COP) during gait. The calculated COP is affected by the material type and thickness covering the top surface. If the surface is deformable, these effects can be significant. The purpose of this study is to simulate and evaluate the effects of deformation height when calculating the COP in a deformable surface during gait. The GRF and COP data during normal gait were collected from 20 healthy adult males on sand in two conditions (level and cross-slope of 10°). The COP differences in the anteroposterior (AP) and mediolateral (ML) directions were modeled for constant deformation heights (10–50 mm, 10 mm increments). The results showed the magnitude of COP changes in the AP and ML directions were different in both level and cross-slope conditions. A significantly larger COP_{ML} difference was shown for the cross-slope condition than level condition for the same deformation height. The COP was more sensitive to the deformation height for the downhill limb than uphill limb in the cross-slope condition. The results of this study suggest that the maximum allowable deformation height before a correction for surface deformation is needed is 20 mm for level condition and 10 mm for cross-slope condition, where 3 mm difference in COP is considered as the tolerance limit. Surface deformations beyond these thresholds may lead to an inaccurate interpretation and evaluation of joint kinetics during gait on deformable surfaces.

1. Introduction

Force plates are key instruments in biomechanics research to provide the ground reaction force (GRF) and the center of pressure (COP) information. The COP is the single point of origin of the ground reaction force vector on a force plate's surface that is used for inverse dynamics. However, it is not directly determined from the force plate and is usually calculated from the analog signals using several parameters which characterize the force plate. When a floor covering exists above the top surface of a force plate, the thickness of the covering effects the calculated COP. However, this parameter may not always be constant and continuously changes during gait on deformable surfaces like sand. Although human locomotion on smooth, hard level surfaces is well described, there is a paucity of studies investigating gait on other surfaces, such as found in outdoor environments including construction work, military activities and railroad work. These often include walking over inconsistent and deformable surfaces like sand or gravel and cross-slope (transversely inclined) conditions. Some previous research has focused on the energy cost of human locomotion on different terrains but did not describe the COP [1–3]. Lejeune et al. investigated the mechanics and energetics of human locomotion on sand [4]. The studies performed by Kim and Yoo analyzed upper and lower limb biomechanics when carrying a military backpack while walking on cross-slope sand surfaces [5,6]. Other studies have focused on the effects of walking on gravel, like railroad ballast. Wade et al. compared the difference of joint kinetics when subjects walked on smooth and ballast surfaces [7]. Merryweather investigated the lower limb biomechanics when walking on cross-slope and level railroad ballast with 10 subjects [8]. A follow-up study using the same data by Xu discussed the effect of cross-slope and ballast surface for the knee contact force during gait [9]. In these studies, the calculated COP from a force plate only accounted for the initial sand or ballast depth. Any

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height change as a result of sand or ballast displacement during contact was neglected or not considered. The COP is a key parameter used for evaluation of joint moments, muscle forces and joint loading during walking; therefore, differences in actual and computed COP on deformable surface could result in erroneous conclusions.

Since the location of the COP affects the estimation of joint kinetics including muscle and joint forces [10–13], and studies of movements on various deformable terrains continue to increase, this research provides valuable guidelines for when it might be acceptable to neglect the effect of a deformable surface on COP. The effect of surface deformation on determining the COP needs to be evaluated to prevent misrepresentation of the results of joint biomechanics of gait on deformable surfaces.

The purpose of this study is to simulate and evaluate the differences in calculating the COP during gait on a deformable surface. Specifically, we would like to know the magnitude of this difference as it relates to the deformation height and whether this is the same for different surface conditions (level and cross-slope), and for different directions (anteroposterior (AP) and mediolateral (ML)).

2. Methods

2.1. Experimental set-up

Twenty healthy male adults (age 24.9 ± 3.5 years; height 1.76 ± 0.04 m; weight 77.2 ± 5.7 kg) volunteered for this study. Participants were carefully selected from a healthy young population who were not currently experiencing injury or pain in the lower extremities that may affect normal gait. All participants reviewed and signed an informed consent document approved by the University of Utah Institutional Review Board. Two adjustable walkways (7.3 m long, 0.76 m wide and 0.23 m deep) and custom isolation fixtures were constructed (Fig. 1) to perform this work. One walkway was covered with 3/4 in. reinforced plywood with two embedded force plates (OR6-5-1000 and OR6-7, AMTI, Watertown, MA) to replicate a hard surface environment. The other track was filled with sand to simulate a common deformable surface environment. Two force plates were embedded 20 cm beneath the sand surface in this track with the isolation fixtures. The isolation fixture consisted of two welded steel rectangles concentrically aligned with 6.4 mm clearance between walls. The outer frame was securely attached to the base of each track, and the inner frame securely fit on the force plate by use of four alignment tabs. The alignment tabs ensured all shear forces were transmitted to the surface of the force plate. Previous studies confirmed that this force plate isolation technique could serve well to reduce the dissipation of force and accurately identify measured GRF on ballast and sand [5,8].

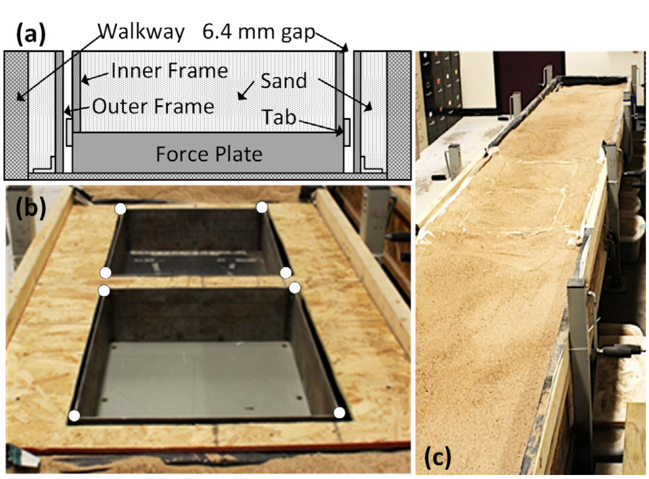

Fig. 1. The adjustable walkways and custom isolation fixture. (a) Schematic diagram of the isolation fixture; (b) the isolation fixture with markers on the inner frame; and (c) the adjustable walkways with embedded force plate.

2.2. Data acquisition

Prior to data collection with participants, eight 10 mm diameter markers were carefully placed at the corners of the force plate using precision machined aluminum jig blocks. These markers where also placed on the inner rectangle of the isolation fixture for sand on both level and cross-slope conditions (Fig. 1). The marker location data were collected using a 16-camera motion capture system (NaturalPoint, Inc., Corvallis, USA) at 100 Hz for 5 s. These data were used to align the force platforms with the laboratory fixed coordinate system and serve as a reference truth for the height of the force plate surface compared with the walking surface (these were equal for the hard surface). Participants received a pair of BELLEVILLE 790G Gore-Tex boots to standardize footwear. The initial calibration trial was collected using the same motion capture system at 100 Hz with a modified Helen Hays marker set for five seconds when the participant stood on one force plate. For the dynamic trials, participants were asked to become accustomed to the walking surface by traversing the walkway, and an optimal starting range was identified to increase the likelihood of two sequential force plate contacts, which was critical for determining a successful trial. Three successful trials per participant at self-selected speed within the range of 1.20–1.40 m/s [14–16] on the tracks were collected for both level and cross-slope of 10° conditions over hard and sand surfaces. The walking direction was same for all trials to keep the left limb in the uphill for the cross-slope condition. Kinematic data were recorded at 100 Hz with digitally low-pass filtered at 6 Hz. The GRF and COP were collected at 2000 Hz with low-pass filtered at 20 Hz.

2.3. Data process

Major gait events (heel strike and toe off) were defined via force plate activation with a 20 N threshold to identify the stance phase of the gait cycle. The GRF and COP data during the stance phase were normalized by 101 data points. In order to evaluate the effect of deformation height for calculating the COP, the deformation height (t) was defined as the maximum vertical slippage depth of the sand surface in both level and cross-slope conditions (Fig. 2). This value was determined by calculating the difference between the height of heel marker on the shoe (h_1) obtained from the calibration trial and the height of heel marker above the reference plane (h_2) defined by the calibration markers on the inner frame (Fig. 2). This was done using MATLAB program (The Mathworks, Inc., Natick, MA). The results showed that the average deformation height during stance was 29 ± 8 mm and ranged from 11 to 41 mm on level condition. The deformation heights on the cross-slope condition were 36 ± 11 mm (14–60 mm) for the uphill limb and 49 ± 11 mm (18–62 mm) for the downhill limb. Therefore, the deformation heights from 10 to 50 mm, in 10 mm increments were chosen for simulation in the present study and analyzed for the total 120 trials with the sand surface (subject (20) $\times$ trial (3) $\times$ condition (2)).

The equations used to calculate the COP in the force plate coordinates were:

$$\mathrm{COP}_x = \frac{-(h-t) \times F_x - M_y}{F_z} \quad (1)$$

$$\mathrm{COP}_y = \frac{-(h-t) \times F_y - M_x}{F_z} \quad (2)$$

where COP_x and COP_y are the coordinates of COP in ML and AP directions, respectively. h is the original sand thickness above the

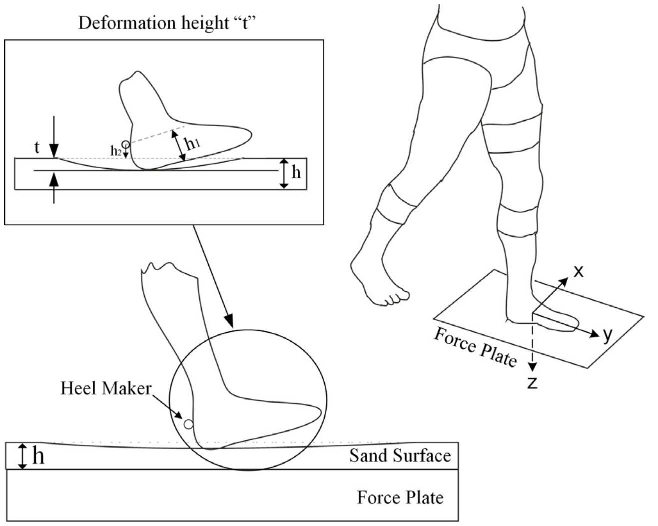

Fig. 2. The force plate coordinate system is *X*-axis to the left, *Y*-axis forward and *Z*-axis downward. The original sand thickness above the top surface of force plate is represented by "*h*". The deformation height is represented by "*t*". The height of heel marker on the shoe is represented by "h_1" and the height of heel marker above the reference plane is represented by "h_2".

top surface of force plate, which is 20 cm in this study, and *t* is the deformation height (Fig. 2). The *F*'s and *M*'s are the force and moment components in the force transducer coordinate system.

The calculated COPs in force plate coordinates were transformed to the laboratory fixed coordinate system before evaluating the magnitude of the difference in COP caused by the deformation height. Root mean square error (RMSE) was used to calculate the average COP differences in the AP or ML direction (Eq. (3)), respectively when comparing no deformation (h = 20 cm and t = 0) with different deformation heights. All the results were firstly averaged across three within-subject trials and then across subjects to obtain group-averaged data.

$$\mathrm{RMSE} = \sqrt{\frac{\sum_{i=1}^{n}(x_{1,i}-x_{2,i})^2}{n}} \tag{3}$$

where $x_{1,i}$ and $x_{2,i}$ represented the corresponding normalized data point of COP for no deformation condition and five different deformable heights, n represented the normalized data point during stance phase.

2.4. Statistics analysis

The statistical analysis of COP difference was performed using SPSS 20 for Windows (IBM Corporation, Armonk, NY). The regression analysis was performed to determine the relationship of the RMSE of COP to the deformation height. The repeated measures analysis of variance (RM ANOVA) was used to determine the effect of surface conditions (level, uphill and downhill limbs) for the difference in COP. The Greenhouse–Geisser correction was used when the assumption of sphericity was violated. Post hoc tests were conducted using Tukey HSD test for pairwise comparison. Paired *t*-test was used for comparing the difference in COP between AP and ML direction in level and cross-slope conditions. The results were considered statistically significant when $p < 0.05$.

3. Results

The GRF was similar (waveform and magnitude) in the vertical and AP directions when comparing limbs in the level and cross-slope conditions on the hard and sand surfaces (Fig. 3). However, the GRF had a higher peak and trough in the AP direction and higher second peak in vertical direction on hard surface than the peak values on sand. For the ML direction, the GRF for the hard surface remained mostly within the range of standard deviation for the level condition and uphill limb in the cross-slope condition. However, this similarity was not observed for the downhill limb.

A linear growth was found between the deformation height and the RMSE of COP in AP and ML direction on both level ($R^2_{AP} = 0.986$ and $R^2_{ML} = 0.995$) and cross-slope conditions ($R^2_{AP} = 0.981$ and 0.983, $R^2_{ML} = 0.978$ and 0.975 for uphill and downhill limbs, separately) based on the regression analysis. The effects were significant among surface conditions for the RMSEs of COP_{ML} ($F(2, 38) = 431.710$, $p < 0.05$). The pairwise comparisons showed a statistical difference between level and uphill limbs ($p < 0.001$), level and downhill limbs ($p < 0.001$), and uphill and downhill limbs ($p < 0.001$). The average RMSEs were 0.5 mm, 1.5 mm and 2.0 mm for level, uphill and downhill limbs, respectively, when the deformation height was 10 mm (Fig. 4a). However, the RMSEs were not significantly different for COP_{AP} ($F(2, 38) = 1.177$, $p = 0.319$) when comparing level, uphill and downhill limbs. The average RMSEs were about 1.3 mm when deformation height was 10 mm (Fig. 4b). The RMSEs were statistically different between COP_{ML} and COP_{AP} for both level and cross-slope conditions (Fig. 4c). The difference was significantly larger for COP_{AP} than COP_{ML} in the level condition ($t(19) = -34.438$, $p < 0.05$). But the opposite results were observed in the cross-slope condition for both uphill limb ($t(19) = 3.409$, $p < 0.05$) and downhill limb ($t(19) = 3.759$, $p < 0.05$).

The difference in COP in both AP and ML directions had a significant increase after 90% of the stance phase for level and cross-slope conditions except for COP_{ML} in the downhill limb (Fig. 5b). When comparing COP between the level and cross-slope conditions, the results indicated the curves of COP_{AP} were quite similar (waveform and magnitude) among level, uphill and downhill limbs, which showed a trough around 50% of stance phase and increased quickly after 80% of the stance phase (Fig. 5a–c). For the ML direction, the RMSEs of COP were relatively stable between 20% and 80% of the gait cycle for both level and cross-slope conditions, but was significantly larger for the uphill and downhill limbs than the level limb.

4. Discussion

The purpose of this study was to simulate and evaluate the difference in COP in the AP and ML direction caused by deformation height when walking on sand in both level and cross-slope conditions. The GRFs were firstly investigated by surface types (hard and sand) and surface conditions (level and cross-slope) since calculated COP was a function of GRF (Eqs. (1) and (2)). The GRFs were similar in the vertical and AP directions when comparing level and cross-slope conditions for both hard and sand surfaces, which aligns with previous research [8,17]. The GRF in the AP direction for sand surface had smaller peaks during heel strike and toe off than hard surface. This is likely the result of sand shifting and the increased energy required to propel the center of mass forward when walking on sand. A lower second GRF peak in the vertical direction was observed for the sand surface, which may be explained by the walking surface deforming under the force applied during toe off. The observed GRF pattern in the ML direction was different for the cross-slope trials compared with level trials, as was expected. The GRF increased laterally for the uphill limb and medially for the downhill limb to oppose the additional shear force acting down the slope in order to preserve the body balance.

The linear relationships between the deformation height and the COP difference in both level and cross-slope conditions were found, as was expected (Fig. 4a and b) since the COP is a function of three factors. These factors are (1) deformation height, (2) the ratio of GRFs in different directions and (3) the transformation matrix between force plate and laboratory fixed coordinate systems. The significant difference of average COP_{ML} among level, uphill and downhill limbs (Fig. 4a) was well reflected by the difference of the COP_{ML} curves in the stance phase (Fig. 5). The similar curves of COP_{AP} in the stance phase (Fig. 5) aligned with the findings that only a slight COP_{AP} difference existed between level and cross-slope conditions, which were not statistically significant (Fig. 4b). The rapid increase of COP_{AP} after 60% of the stance phase (Fig. 5) resulted in the significantly larger differences for COP_{AP} than COP_{ML} in the level condition (Fig. 4c). However, the COP_{ML} difference was shown to be statistically larger than COP_{AP}

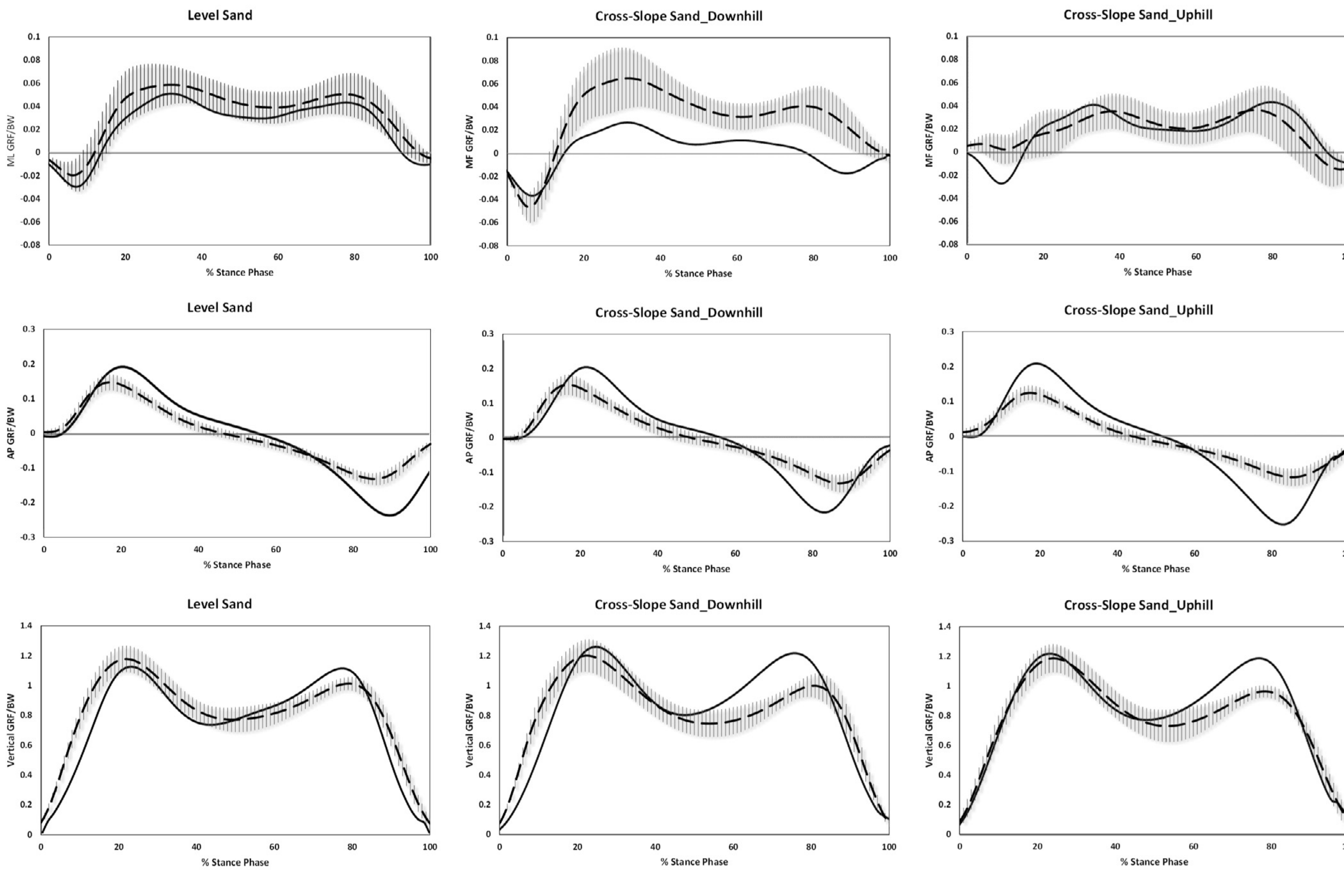

Fig. 3. Comparison average GRF in three directions between sand (dashed line) and hard (solid line) surfaces for corresponding level and cross-slope conditions. The shaded regions indicated standard deviation for sand surface.

difference for both uphill and downhill limbs in the cross-slope condition. This was mainly because of the variance in COP between the AP and ML directions in the middle of stance phase (Fig. 5a and b).

Some previous research focused on the effect of COP on the joint moment during gait. McCaw and DeVita used two-dimensional kinematic and kinetic data to evaluate the COP_{AP} for lower extremity torques. They found that ±5 mm and ±10 mm shifts in the COP_{AP} caused mean changes of 7% and 14% in maximum joint torque, respectively [12]. Kim et al. extended the evaluation to the three-dimensional lower limb joint, they shifted the COP toward the anteroposterior and mediolateral direction by ±10 mm, ±20 mm and ±30 mm. The findings of their study indicated the joint moment magnitudes were more sensitive to the COP change in the ML direction than in the AP direction [18]. Camargo et al. further investigated the effect of COP during gait at different velocities and concluded that the absolute peak moment uncertainties decreased from distal to proximal joint and also from the lower to higher gait velocity [13]. In these studies, the researchers suggest that the COP errors should be less than 3 mm on average and less than 5 mm of maximum to prevent remarkable changes of joint moment. In our study, the overall difference in COP was more sensitive in the cross-slope condition compared to the level condition, especially for the downhill limb. When 3 mm COP error is considered as the tolerance limit, 10 mm and 20 mm deformation heights could be the threshold in the cross-slope and level conditions prior to adjusting the deformation height (t) in Eqs. (1) and (2).

This study has practical implications for other gait research on deformable surfaces. Joint kinetics are critical to predict muscle forces and joint contact forces during gait. Therefore, it is highly possible that the difference in calculated COP caused by deformation height affects the estimation of joint moments and further for an accurate interpretation and evaluation of risk while traversing deformable surfaces.

We acknowledge several limitations in this study. First, the deformation height was considered to be a constant value during the stance phase in the simulations, which is actually changing continuously when walking on deformable surfaces. Second, the method used to calculate deformation height likely overestimated the true value since the angle between the deformable sand and the shoe was neglected. These two limitations should not significantly affect the estimate of deformation height, which considered as the maximum deformation depth in our study, since it happened in the midstance phase after the contralateral toe off when the angle is really small. A more sophisticated method could be implemented to measure actual deformation height in future studies.

In summary, the COP differences caused by the constant deformation height while walking on a deformable surface (sand) in level and cross-slope conditions was characterized in the present study. It was found that (1) the COP differences in the AP and ML direction were significantly different in both level and cross-slope conditions, (2) the COP_{ML} difference was larger for the cross-slope condition than level condition when deformation height was the same, and (3) the overall difference in COP was more sensitive to the change of deformation height for the downhill limb than uphill limb in the cross-slope condition. These results provide guidance for those conducting studies examining gait on deformable surfaces and show evidence that minor surface

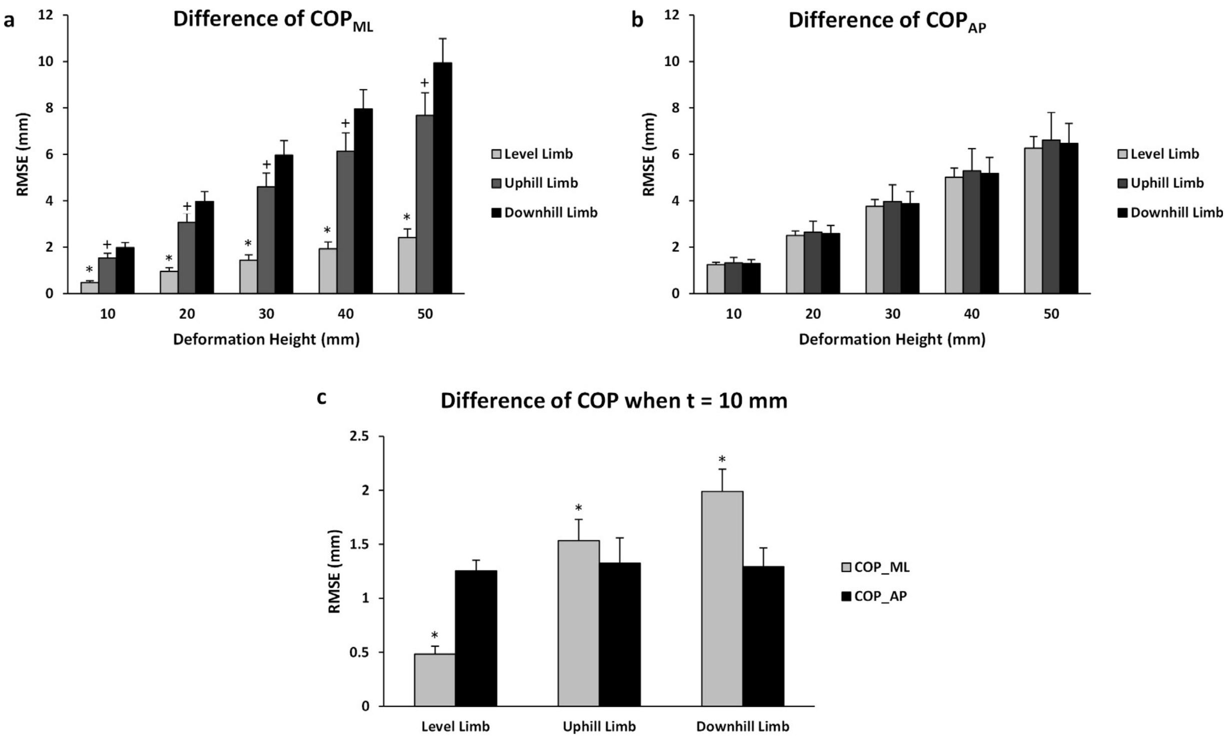

Fig. 4. Simulation results of RMSE for t = 10–50 mm (a) RMSE of COP_{ML}. * indicates a significant difference between level and cross-slope conditions (both uphill and downhill limbs), + indicates a significant difference between uphill and downhill limbs; (b) RMSE of COP_{AP}; and (c) RMSE when deformation height is 10 mm. * indicates a significant difference between COP_{AP} and COP_{ML} in the same condition.

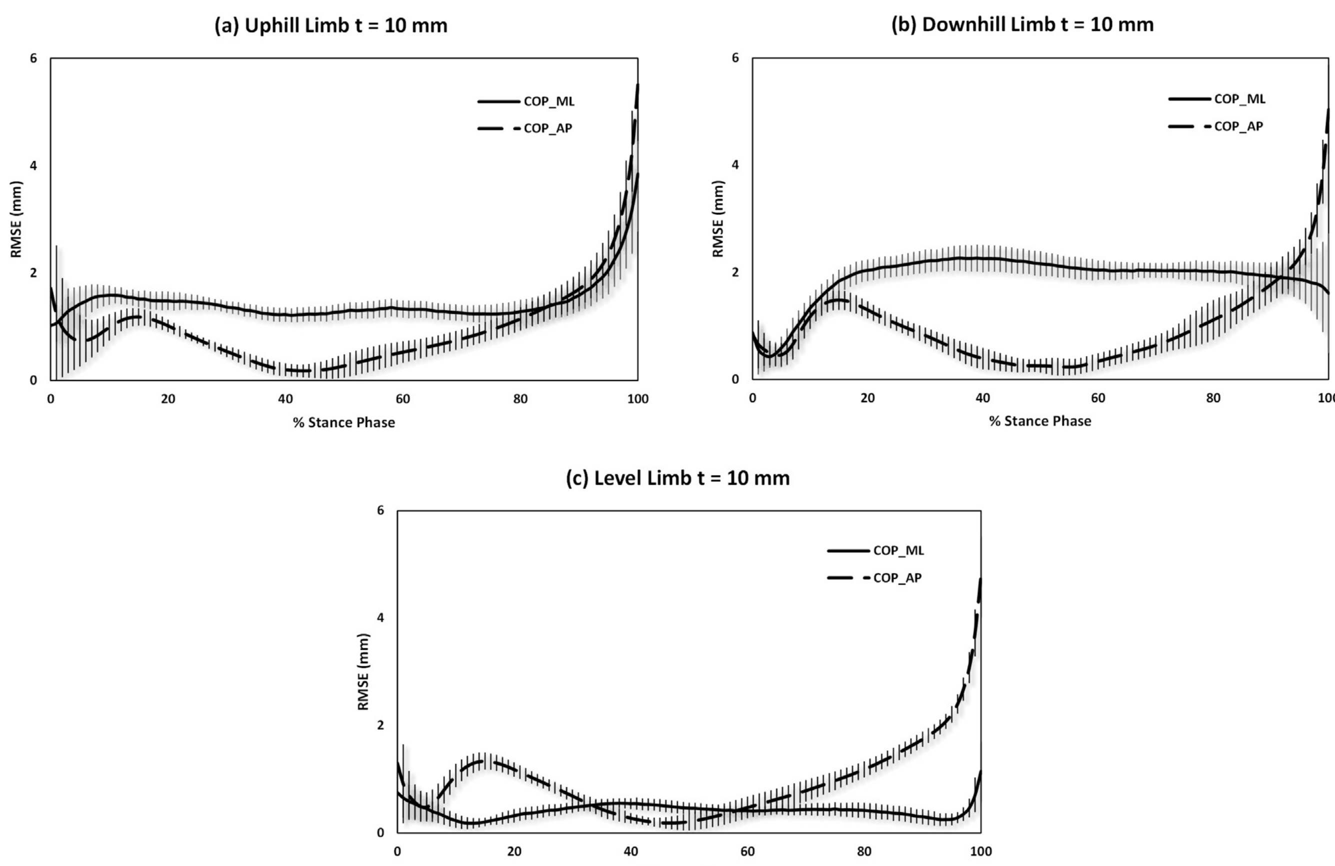

Fig. 5. RMSE of COP_{ML} and COP_{AP} during stance phase of gait cycle when deformation height (t) is 10 mm. (a) Uphill limb; (b) downhill limb; and (c) level limb.

deformation may lead to unacceptable changes in calculated COP. We suggest that maximum deformation limits are 20 mm for the level condition and 10 mm for the cross-slope (10°) condition without implementing correction methods.

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Conflict of interest statement: The authors attest that there are no conflicts of interest to disclose.

References

[1] Soule RG, Goldman RF. Terrain coefficients for energy cost prediction. J Appl Physiol 1972;32:706–8.
[2] Zamparo P, Perini R, Orizio C, Sacher M, Ferretti G. The energy cost of walking or running on sand. Eur J Appl Physiol Occup Physiol 1992;65:183–7.
[3] Pinnington HC, Dawson B. The energy cost of running on grass compared to soft dry beach sand. J Sci Med Sport 2001;4:416–30.
[4] Lejeune TM, Willems PA, Heglund NC. Mechanics and energetics of human locomotion on sand. J Exp Biol 1998;201:2071–80.
[5] Yoo B. The effect of carrying a military backpack on a transverse slope and sand surface on lower limb during gait. Salt Lake City: The University of Utah; 2014
[6] Kim S. Ergonomic analysis of army backpack designs: back and shoulder stresses and their implications. Salt Lake City: The University of Utah; 2014.
[7] Wade C, Redfern MS, Andres RO, Breloff SP. Joint kinetics and muscle activity while walking on ballast. Hum Factors 2010;52(October):560–73.
[8] Merryweather AS. Lower limb biomechanics of walking on slanted and level railroad ballast. Salt Lake City: The University of Utah; 2008.
[9] Xu H. Development of a musculoskeletal model to determine knee contact force during walking on ballast using opensim simulation. Salt Lake City: The University of Utah; 2013.
[10] Cappozzo A, Leo T, Pedotti A. A general computing method for the analysis of human locomotion. J Biomech 1975;8(September):307–20.
[11] Takeshita D, Shibayama A, Fukashiro S. Effect of errors of the center of pressure (CoP) on lower limbs joint moment during vertical jump. Jpn J Biomech Sports Exerc 2000;4:63–70 [in Japanese].
[12] McCaw ST, DeVita P. Errors in alignment of center of pressure and foot coordinates affect predicted lower extremity torques. J Biomech 1995;28:985–8.
[13] Camargo-Junior F, Ackermann M, Loss JF, Sacco I. Influence of center of pressure estimation errors on 3D inverse dynamics solutions during gait at different velocities. J Appl Biomech 2013;29:790–7.
[14] Delval A, Salleron J, Bourriez J-L, Bleuse S, Moreau C, Krystkowiak P, et al. Kinematic angular parameters in PD: reliability of joint angle curves and comparison with healthy subjects. Gait Posture 2008;28:495–501.
[15] Robbins SM, Maly MR. The effect of gait speed on the knee adduction moment depends on waveform summary measures. Gait Posture 2009;30:543–6.
[16] Heiden TL, Sanderson DJ, Inglis JT, Siegmund GP. Adaptations to normal human gait on potentially slippery surfaces: the effects of awareness and prior slip experience. Gait Posture 2006;24:237–46.
[17] Giakas G, Baltzopoulos V. Time and frequency domain analysis of ground reaction forces during walking: an investigation of variability and symmetry. Gait Posture 1997;5:189–97.
[18] HeungYoul K, Sakurai S, JaeHan A. Errors in the measurement of center of pressure (CoP) computed with force plate affect on 3D lower limb joint moment during gait. Int J Sport Health Sci 2007;5:71–82.