NEUROMECHANICAL ADAPTATIONS TO REAL-TIME BIOFEEDBACK OF THE
CENTER OF PRESSURE DURING HUMAN WALKING

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ABSTRACT

Michael Gordon Browne: Neuromechanical adaptations to real-time biofeedback of the center of pressure during human walking
(Under the direction of Gregory S. Sawicki)

The purpose of this study was to understand the effects of adjustments to the center of pressure (COP) via real-time visual biofeedback on joint loading in the frontal and sagittal planes while walking. Eight subjects walked on an instrumented treadmill while provided bilateral targets for toe-off on a visual display alongside real-time COP trajectories. Toe-off targets included a neutral location along with medial, lateral, anterior and posterior shifts. Resultant COP shifts caused compensations in joint mechanics; anteriorly/posteriorly shifted COP, was achieved by velocity changes to COP progression, and lead to increases/decreases in plantarflexor angle and reductions in hip extension moment while laterally/medially shifted COP, was achieved through spatial changes to COP progression, lead to increases/decreases in both peak inversion ankle angle and moment. Temporal modifications to peak muscle activities drove mechanical changes. Results suggest that COP biofeedback could be a useful tool or shaping joint kinematics/kinetics during functional locomotion tasks.

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I want to thank everyone who has been a part of my graduate experience so far. Greg Sawicki, thank you for mentoring me and my research through my confused undergraduate days into my [less confused] graduate days. Andrew DiMeo, thank you for supporting my varied interests in teaching and product development and brainstorming how to break the system. Jason Franz, thank you for your continued patience, support and input through my completion of this project. I need to thank all the current and past members of the Human PoWeR Lab but specifically Rich Nuckols and Kota Takahashi for all their help on this project being a sounding board whenever I needed to discuss a new approach. Last but not least, thank you to my family who believed in me all along.

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LIST OF ABBREVIATIONS

A/P   Anterior/Posterior

COP   Center of Pressure

GRF   Ground Reaction Force

M/L   Medial/Lateral

MTPJ1   First Metatarsophalangeal Joint

MTPJ5   Fifth Metatarsophalangeal Joint

CHAPTER 1: CENTER OF PRESSURE MODIFICATION VIA VISUAL BIOFEEDBACK

Introduction

Motor learning and re-learning (*e.g.* rehabilitation) is more effective and can happen more quickly when feedback is provided to enforce corrected movements. Visual biofeedback, providing an individual with visible cues, has shown promising effects on a variety of mobility outcomes with significant research looking at its use in static stability control (Cofré Lizama et al., 2015; D'Anna et al., 2015; Lakhani and Mansfield, 2015).Visual biofeedback during gait has also been attempted with a multitude of gait metrics (*e.g.* EMG, distorted stride length, etc.) (Franz et al., 2014; Kim and Mugisha, 2014) and with visual modalities ranging from simple mirrors (Willy et al., 2012) to complex delayed contralateral limb mirroring in a virtual reality system (Barton et al., 2014).

The center of pressure (COP), defined as the centroid of all external forces acting on the plantar surface of the foot (Lugade and Kaufman, 2014), is a promising cue for visual biofeedback as it resides at the base of the kinetic chain. During normal walking, the COP propagates from heel to toe on the lateral aspect of the foot until late stance when it quickly progresses medially during push-off (Lugade and Kaufman, 2014). Perhaps most importantly, the 3 dimensional COP location with reference to the ankle joint center influences the moment arm of the ground reaction force (GRF), thereby affecting leg joint moments (Farris and Sawicki, 2012; Huang et al., 2015). Furthermore, joint mechanics have been shown to have high sensitivity to changes in COP (Camargo-Junior et al., 2013; McCaw and DeVita, 1995). This implies that modification of this single variable has the potential to alter the dynamics (kinematics and kinetics) of multiple lower extremity joints.
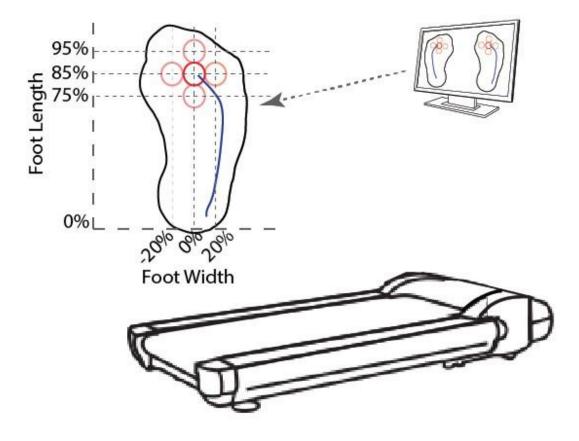
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In spite of this, COP has been used primarily as an outcome measure through a wide variety of biomechanical manipulations and foot pathology evaluations (*i.e.,* through pedobarography). Although COP is considered to be modulated specifically by joint torques, specifically at the ankle (Gruben and Boehm, 2014), it is poorly understood how humans modulate COP and what effects more proximal joints (*i.e.,* knee and hip) play in its modification. The purpose of this study was to determine if visual biofeedback of COP during healthy walking could induce systematic changes on lower extremity joint mechanics (*i.e.,* kinetics and kinematics).

Clinically, modification of the COP has been used for pain relief and in attempts to improve joint alignment and dynamic moments. Modified footwear has been developed to guide both the mediolateral and anteroposterior propagation of the COP using movable domelike attachments to shoe soles (Khoury et al., 2015). Additionally, research to decrease the external knee adduction moment in patients suffering from knee osteoarthritis have implemented simple shoe wedges under the lateral portion of the heel (Chapman et al., 2015; Jones et al., 2015), essentially modifying the COP during loading. To our knowledge, however, no research has focused on the implications that intentional changes in COP propagation have on motor coordination and joint kinematics and kinetics, even in healthy, young individuals.

Considering the ability of the COP to modify lower extremity joint moments, we strove to investigate whether real-time COP biofeedback during gait could function as a translational replacement for more complex mechanical and/or biofeedback based treatments. We developed a system to visually portray COP trajectory in the transverse plane of the foot along with target locations for toe off (Figure 1). Our goal was to determine healthy individuals' ability to intentionally modulate their COP in real-time in response to targets while walking, and furthermore, to evaluate biplanar correlations between COP and joint moments. We hypothesized that 1) subjects would volitionally modify and maintain an

altered COP trajectory when provided with real-time visual biofeedback of COP, 2) shifting the COP trajectory anteriorly/posteriorly (A/P) would increase/decrease sagittal plane plantarflexor ankle moments, respectively, 3) shifting COP trajectories medially/laterally would decrease/increase frontal plane inversion ankle moments, respectively, and lastly 4) we would see increases in coactivation of the triceps surae muscles, specifically with the anteroposterior shifts in COP.

Figure 1 – Biofeedback Schematic

Split belt BERTEC treadmill with an eye-level computer monitor was utilized to modulate COP targets bilaterally. Each axis denote % shifts of respective targets. COP trajectories were shown with a target representing an anthropometric (NEUTRAL) location, M/L shifts (MEDIAL/LATERAL) and A/P shifts (ANTERIOR/POSTERIOR).

Methods

<u>Participants</u>

We recruited eight healthy, young adult subjects who provide written, informed consent (University of North Carolina IRB Office) to be in this study. The group consisted of five males and three females with mean (SD) age: 26.9 (2.7) years and mass: 70.49 (13.57) kg. None reported any musculoskeletal injuries within six months of testing nor had any neurological impairments.

<u>Procedures</u>

41 retroreflective markers were used in order to collect motion capture data for use in the creation of an inverse dynamic model as well as real-time COP tracking. For dynamic trials (*i.e.,* walking) markers were placed bilaterally over the calcaneus, in a 3-marker cluster on top of the foot, in 4-marker clusters on the shank and thigh, and a 3-marker cluster on the pelvis. Static trials also included anatomical landmarks to complete the model with markers on the base of the 2nd metatarsal, medial and lateral malleoli, medial and lateral femoral epicondyles, greater trochanters and iliac crests. All subjects walked at 1.25 m/s barefoot for all trials. We collected basic anthropometric data (foot width, foot length, and distance from toe to end of foot) for inverse dynamic calculations and assistance with real-time COP plotting. All subjects walked with six different conditions including no feedback (NOFEED), a neutral position defined by foot size (NEUTRAL), and four shifted toe-off locations: medial shift (MEDIAL), lateral shift (LATERAL), posterior shift (POSTERIOR), and anterior shift (ANTERIOR). All trials after the NOFEED condition were randomized to avoid bias ordering effects based on learning adaptation or fatigue. After a minimum of 3 minutes walking with biofeedback, we collected 10 consecutive steps of marker trajectories and analog data for analysis.

<u>Biofeedback Display</u>

We used a pre-collection static capture to establish an accurate biofeedback projection of the transverse plane of the foot. Subjects stood with feet parallel to establish an original rotation matrix for each foot. These 3x3 matrixes were calculated using two perpendicular vectors formed by the right triangle alignment of the 3-marker cluster attached to the top of each foot and their cross-product. The difference between this rotation matrix and the lab's coordinate system provided an offset, at each frame, to preserve the visual representation of of each subjects' foot.

A computer monitor centered at eye-level in front of the treadmill to displayed the biofeedback (Figure 1). A software development kit paired with the motion capture software (Vicon Nexus) assisted in passing marker and ground reaction force data in real time to a custom script written in Matlab, a computation software (MathWorks, Natick, MA). The scripts first determined heel-strike with a vertical GRF threshold of 20N. The treadmill COP location, extracted from the treadmill in its internal coordinate system, was subtracted from the calcaneus marker position to translate its position. Then, using the same cluster on top of the foot to calculate a frame-by-frame, 3x3 rotation matrix, the COP location was transformed from each belt into the appropriate foot's coordinate system. The entire COP trajectory for each foot was plotted real-time. A delay of 3 frames (0.025s) allowed for the last 3 frames of data to be truncated to eliminate large shifts in COP contributed to ground reaction force noise at low forces. This method was utilized in place of active filtering of analog and marker data to maximize frame rate. Furthermore, to compensate for delay in the system, a rolling average of the previous 3 strides was also shown. A different red circle was shown as a target for the toe-off location. All data were plotted on top of a graphic representing a transverse plane image of the bottom of the foot with both length and width scaled to each subjects' anthropometry (Figure 1). Subjects were instructed to use any technique necessary to achieve targets while maintaining a heel-to-toe-style walking pattern.

<u>Targets</u>

Target size diameter was defined as 10% of the foot width between MTPJ1 and MTPJ5. The NEUTRAL location was defined as along the midline of the foot (50% foot width) to ensure there would be physical space medially and laterally for each subsequent shift. Research has shown that COP propagates approximately 83% of the foot length (Han et al., 1999; Lugade and Kaufman, 2014) so the NEUTRAL location along the anteroposterior axis of the foot was set at 85% of the foot length from the calcaneus. Each shift (MEDIAL, LATERAL, ANTERIOR, POSTERIOR) in toe-off location was 10% of foot length along the respective direction of the foot away from the NEUTRAL location (Figure 1).

<u>Data Analysis</u>

Marker trajectories (Vicon Nexus, Denver, CO) were collected at 120 Hz with analog data collected at 960 Hz. Raw marker positions were filtered using a second-order low pass Butterworth filter with a cut-off frequency of 8 Hz. Raw force data for use in the inverse dynamic calculations were filtered with a second-order low pass Butterworth filter with a cut-off frequency of 35 Hz. We incorporated these data into an inverse dynamic model to estimate joint centers in relation to cluster locations. We collected EMG from the medial gastrocnemius (MGAS), lateral gastrocnemius (LGAS), and soleus (SOL) then band-pass filtered (20-460 Hz), rectified, and low pass filtered with a cutoff frequency of 6 Hz these data. Finally, we integrated data across time and normalized by the peak values. We calculated all joint kinetics and kinematics using Visual3D Software (C-Motion, Inc., Germantown, MD). We evaluated data for the right leg in all conditions.

We calculated the average location of the COP location along both the M/L and A/P axes of the foot across the entire stance phase as a modality to determine total resultant shift in COP over stance. We extracted relevant peak values as well as averages across stance (0-60% of the gait cycle) from each

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mechanical outcome measure (*i.e.,* peak angles, moments) as denoted by vertical green lines in Figures 3 and 8. We additionally averaged EMG activity across early-to-mid stance (0-40%), across the entire stance phase (0-60%) and extracted its peak value. Lastly, for the anteroposterior shifts, we extracted peak propulsive force values.

Statistics

We used SPSS Statistics 23 (IBM, Armonk, North Castle, NY) to compute all statistical outcomes. We used one-way repeated measures Analysis of Variance (ANOVA) to calculate main effects ($p < 0.05$) and pairwise t-tests ($p < 0.05$) to evaluate specific effects between conditions for each variable as well as to determine the main/individual effects of biofeedback on COP modification.

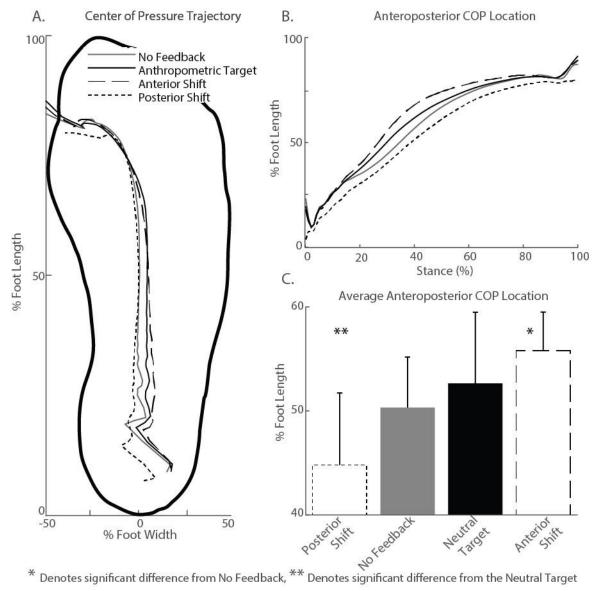**Results**

COP Target-Matching Accuracy

Across stance, we saw no significant differences between the NEUTRAL and NOFEED COP locations in neither the frontal nor sagittal planes (Figures 2C and 7C) (p=0.25 for anteroposterior location and p=0.68 for mediolateral location). Furthermore, we observed no significant changes in the final point of COP (Figure 2A). We did see, however, significant shifts in COP location when observing the location averaged across stance.

Anteroposterior Shift

Biofeedback of a posteriorly shifted toe-off location significantly shifted the average COP posteriorly when compared to NEUTRAL (p=0.021) and ANTERIOR (p=0.001) as well as a nearly significant shift compared to NOFEED (p=0.051) (Figure 2B). The anteriorly shifted toe-off

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biofeedback location elicited a significant anterior shift of the average COP when compared to
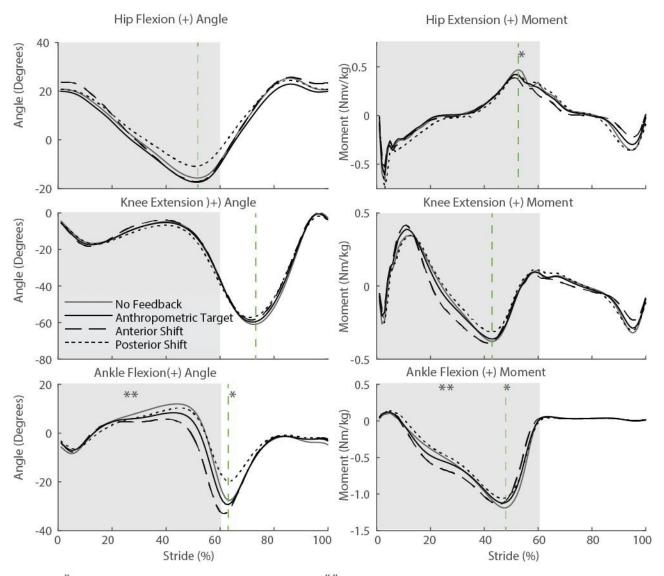
NOFEED (p=0.016) and POSTERIOR (p=0.001) (Figure 2B).

Figure 2 - Anteroposterior COP Location

A. Transverse plane COP trace for shifts along the anteroposterior axis of the foot

B. Anterior (positive) propagation of the COP normalized to foot length over time

C. Average anteroposterior location of COP over stance phase.

For anteroposterior targets, the primary kinematic adaptations were seen at the ankle and hip, with posterior shifts in COP leading to less ankle plantarflexion and less hip extension and conversely, anterior shifts in the COP leading to increased ankle plantarflexion (Figure 3). This trend was supported statistically with significant increases in peak plantarflexion for the anterior shift ($p=0.023$) and significant decreases for the posterior shift ($p=0.012$) when compared to the NEUTRAL target (Figure 4).

We found no significant difference in peak plantarflexion moments. When averaged across stance, however, the anteriorly shifted COP caused a significant increase in ankle moment when compared to the NEUTRAL ($p=0.046$) condition (Figure 3).

Peak hip extension moments for both ANTERIOR and POSTERIOR were significantly reduced than NOFEED ($p=0.026$, $p=0.021$ respectively) condition (Figure 4) though not with respect to NEUTRAL.

Figure 3 - Sagittal plane joint mechanics for anteroposterior shifts
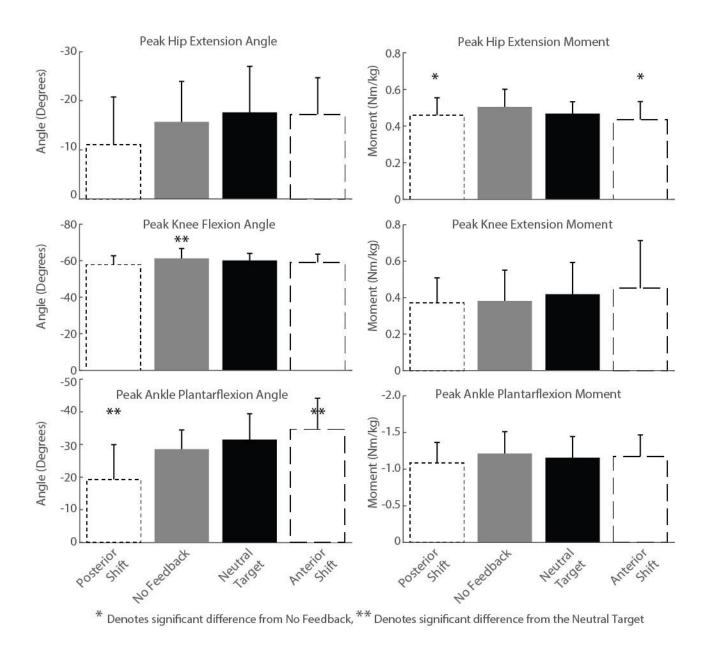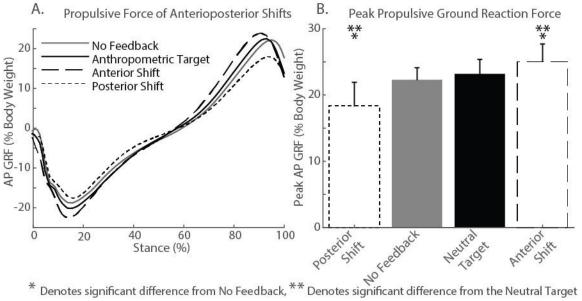
Figure 4 - Peak sagittal plane joint mechanics for anteroposterior shifts

Peak values of sagittal plane hip, knee and ankle angles and moments across stride. (Peak knee moment is the peak extension moment)

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Peak propulsive force was both significantly increased and decreased for the ANTERIOR and POSTERIOR shifts, respectively, when compared to NOFEED (p=0.007 and p=0.018, respectively) and NEUTRAL (p=0.020 and p=0.006, respectively) (Figure 5B).

Figure 5 - Propulsive force for anteroposterior shifts

A. Anteroposterior (AP) Ground Reaction Force (GRF) traces for COP shifts along the anteroposterior axis of the foot.

B. Peak Propulsive GRF values demonstrate a reduction with the posterior shift and an increase with the anterior shift.

A/P COP adjustments had no significant effects on triceps surae muscle activations (Figure 6).
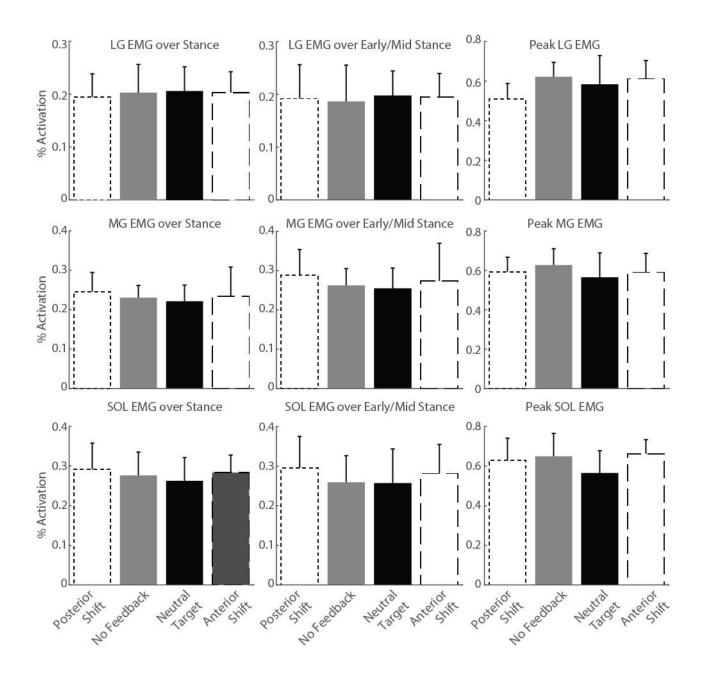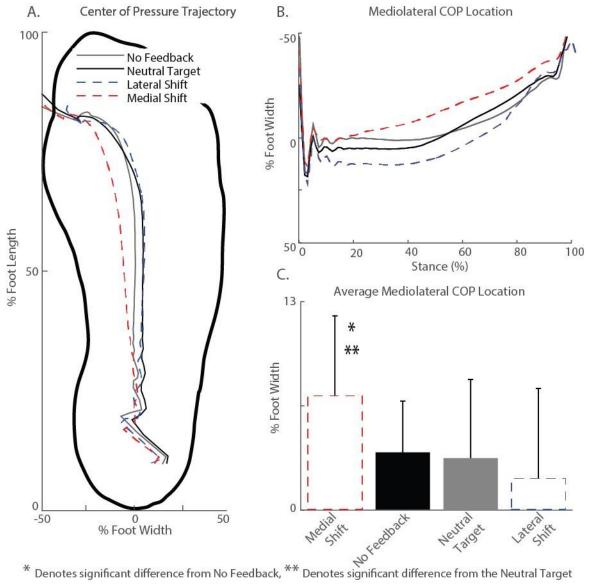

Figure 6 - Triceps surae EMG activity for anteroposterior shifts

EMG activity of the triceps surae muscles (Lateral Gastrocnemius: LG, Medial Gastrocnemius: MG, Soleus: SOL) over the entire stance, early to mid-stance (1-40% of gait cycle) and the peak activations.

Mediolateral Shift

Biofeedback of a medially shifted toe-off location significantly shifted average COP medially when compared to NEUTRAL (p=0.001), NOFEED (p=0.014) and LATERAL (p=0.014) (Figure 10).
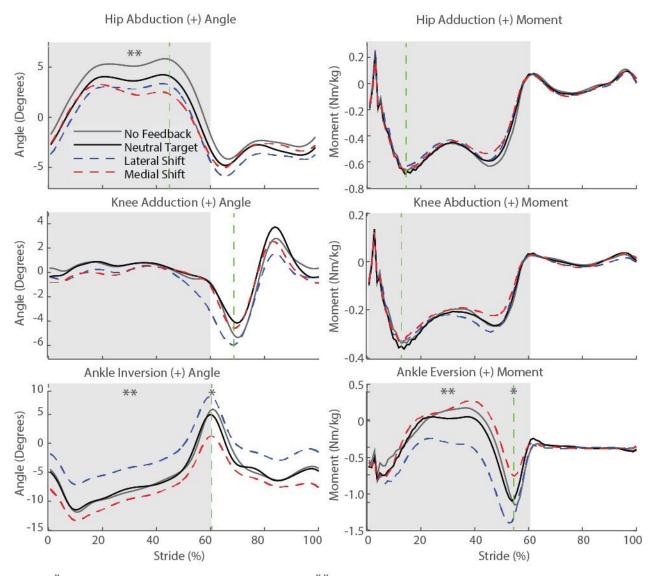

Figure 7 - Mediolateral COP Location

A. Transverse plane COP trace for shifts along the mediolateral axis of the foot

B. Medial/Lateral (up/down) propagation of the COP normalized to foot length over time

C. Average mediolateral location of COP over stance phase

Across the mediolateral condition changes, the primary mechanical adaptations were made by the ankle angle and moment. As COP shifted from medial to lateral, peak ankle angle also shifted from eversion to inversion, respectively. We observed this adaptation both across stance as well as at the peak values. Pairwise comparisons showed that MEDIAL was significantly more everted than NOFEED condition ($p=0.033$) although not significantly everted than NEUTRAL ($p=0.064$). LATERAL was significantly more inverted than NEUTRAL ($p=0.026$).

A shift in COP demonstrated a significant main effect on peak and average values of the frontal plane ankle moment during stance. LATERAL caused a significant peak ankle moment increase from the NOFEED ($p=0.042$) condition (Figure 9).

Lastly, we observed a shift to a less abducted hip posture across all biofeedback trials, supported by a significant main effect across stance ($p=0.042$) (Figure 8)

Figure 8 - Frontal plane joint mechanics for mediolateral shifts

Figure 9 - Peak frontal plane joint mechanics for mediolateral

Peak values of frontal plane hip, knee and ankle angles and moments across stride. (Peak knee moment is the peak adduction moment)

M/L COP adjustments had no significant effects on triceps surae muscle activations (Figure 10).

Figure 10 - Triceps surae EMG activity for mediolateral shifts

EMG activity of the triceps surae muscles (Lateral Gastrocnemius: LG, Medial Gastrocnemius: MG, Soleus: SOL) during anteroposterior shifts to COP over the entire stance, early to mid-stance (1-40% of gait cycle) and the peak activations.

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Discussion

This study strove to determine the ability of individuals to intentionally modify their COP trajectory while walking with visual biofeedback and to evaluate the sensitivity of lower extremity joints to the respective changes. We hypothesized that 1) subjects would volitionally modify and maintain an altered COP trajectory when provided with real-time visual biofeedback of COP, 2) shifting the COP trajectory anteriorly/posteriorly (A/P) would increase/decrease sagittal plane plantarflexor ankle moments, respectively, 3) shifting COP trajectories medially/laterally would decrease/increase frontal plane inversion ankle moments, respectively, and lastly 4) we would see increases in coactivation of the triceps surae muscles, specifically with the anteroposterior shifts in COP. Our data largely support the first three hypotheses but reject the fourth. In general, subjects responded to COP toe off target locations by altering the spatiotemporal trajectory. In order to achieve the COP shifts along the anteroposterior axis of the foot subjects resulted to using the temporal characteristic of COP. Alternatively, to achieve shifts in the mediolateral axis of the foot, subjects predominantly modified spatial coordinates of the COP. As hypothesized, we observed the ankle mechanics (*i.e.,* angles and moments) being the most sensitive to both mediolateral and anteroposterior shifts to the COP. Interestingly, we observed no resultant changes in the magnitude of EMG activations, although this is likely due to the temporal nature of COP leading to unaccounted for temporal shifts in EMG activity.

Anteroposterior Shift

Along the anteroposterior axis of the foot, biofeedback of a posteriorly shifted toe-off location was effective in inducing a posterior shift of about 5% foot length (*i.e.,* shorter moment arm of ground reaction force) to the average COP trace while an anterior target also induced an anterior shift of about 5% foot length (ie. longer moment arm of ground reaction force) when compared to normal. The slope (velocity) of the COP propagation through mid-stance across percent stride (Figure 2B) suggests that

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subjects used temporal changes to COP to induce these changes. The anterior shift to the COP was achieved through more rapidly reaching the anterior portion of the foot. Approximately 75% of the foot length was utilized during the first 50% of stance. Conversely, the opposite trend was seen with the posterior shift with approximately 60% of the foot length being used in the first 50% of stance. Furthermore, although the final locations of the COP were not significantly different, the respective increase and decrease in average ankle moment across stride can be attributed to the associated increase and decrease in propulsive ground reaction force.

Mediolateral Shift

Along the mediolateral axis of the foot, a medially shifted COP target resulted in a medial shift of the average COP of about 3% foot length and, while no significant change was observed, a lateral shift in target also induced a lateral shift in average COP of about 3% foot length. In a very similar manner as the anteroposterior shifts COP converged to the same general area in the medial and anterior aspect of the foot with reduced forces. The COP velocity (slope of Figure 7B) was largely uneffected such as seen in the anteroposterior shifts, however, subjects started relatively inverted or everted and maintained the position until late stance (approximately around 75% of stance phase).

The observations at the ankle during the mediolateral shifts were consistent with the hypothesized ankle moment changes. Peak ankle inversion moment increased with lateral shifts to the COP and there were non-significant reductions in peak ankle moment for medial shifts. The medial shifts to the COP also showed altered ankle range of motion. This may create opportunities to translate the biofeedback approach to improve equinovarus posture for stroke populations (Khallaf et al., 2014) or as an alternative to bracing in ambulating children with clubfoot to maintain an abducted and everted foot posture while walking (Dimeo et al., 2012). While these moment changes are consistent with expectations, we did not observe any reductions in knee adduction moments as would be relevant in

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knee osteoarthritis populations. Furthermore, while research has shown that peak knee adduction moment can be reduced with lateral wedges forcing the COP medially (Sawada et al., 2016), this study suggests that such a shift in COP volitionally does not necessarily elicit a response at the knee.

Electromyography

The lack of magnitude changes to triceps surae muscle activations was somewhat surprising. Work performed by Goryachev et al. used mechanical modification of the COP through custom footwear and saw variations in the lateral gastrocnemius activity during terminal stance and pre-swing (Goryachev et al., 2011). These authors also demonstrated modifications to the COP by the tibialis anterior, biceps femoris, and vastus lateralis which were not measured in this study but may have contributed to A/P and M/L COP shifts based on their architecture and relation to the joints. Additionally, no muscles relating specifically to ankle inversion/eversion such as the peroneus longus were collected. Future studies should investigate the utilization of more proximal muscles such as the hip abductor/adductor muscles in addition to ankle inverter/everter muscles. An alternative possibility is that passive structures at the ankle (ie. the Achilles tendon) are playing a larger role with alterations of the COP thus eliminating the need to increase/decrease activation of the muscles. Additionally, it is important to consider that small changes in the location of the center of mass can create large shifts in the COP simply by capitalizing on a longer moment arm.

Limitations

Gait speed may have played a role in the difficulty of subjects to fully maintain accurate COP trajectories. Modulation of a 2 dimensional parameter bilaterally may have been a difficult task to perform while maintaining the speed of 1.25 m/s. Anecdotally, subjects mentioned often switching focus from one foot to the other after many consecutive steps, suggesting a cognitive overload which was not

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accounted for in this study. The complexity of volitionally adjusting movement for two feet each in two degrees of freedom (each target had both an anteroposterior and mediolateral position) could have been taxing. This may imply that, should real-time COP biofeedback be implemented, a single degree of freedom (*i.e.* visual display of anteroposterior or mediolateral displacement only) which is more outcome derived should be used. Future research should utilize either larger target quadrants of the foot or visual feedback of the displacement only along the anteroposterior or mediolateral axis.

Implications and Future Work

As observed in our results, timing was a large consideration when modifying COP propagation, seen in COP traces as well as angle and moment profiles. COP shifts along the anteroposterior axis of the foot relied almost exclusively on this temporal component to COP while mediolateral shifts acted primarily within the realm of time-independent mechanical adaptations. Joint loading for various pathologies also has a temporal aspect, such as peak knee adduction moment occurring in early stages of stance phase. Future research should investigate targeting COP changes at different locations on the foot that may more accurately align with more specific phases of joint mechanics. For instance, providing a target shifted along the mediolateral axis of the foot just after heel-strike may improve the chances of reducing the knee adduction moment that is observed during loading.

Potentially the most translational contributions from the anteroposterior shifts to the COP was the increase (anterior) and decrease (posterior) in propulsive force demonstrated. Propulsive force is closely related to preferred walking speed in healthy and impaired populations, such as stroke (Bowden et al., 2006). One limitation of in-sole technologies is their inability to accurately measure shear forces (ie. propulsive forces). Utilizing anterior shifts to the COP through in-shoe sensors may be an effective modality to attempt to increase propulsive forces through correlation and thus preferred walking speed similar to work performed by Franz and Kram (Franz et al., 2014). Surprisingly here, we did not observe

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an increase in peak ankle moment to go along with our increase in propulsive force. One possible explanation could lie in the metatarsal phalangeal joint, and its capacity to generate force (Goldmann and Brüggemann, 2012). Increasing and decreasing the moments generated by the metatarsal phalangeal joint could have played some role in the propulsive GRF while not affecting ankle moments.

One factor making the utilization of COP exciting is its potential for translation. Most lab-based technologies for measuring COP involve the use of force plates, either imbedded in a treadmill or on over-ground walkways. Significant effort, however, has gone into the validation and development of various commercially available wearable insoles such as the Pedar-X insole system (Novel, Munich, Germany), the Parotec System (Paromed, Neubeuern, Germany), and the F-Scan (Tekscan, South Boston, MA) (Chesnin et al., 2000; Debbi et al., 2012; Han et al., 1999). Furthermore, newer research has pushed to develop systems to reduce cost and noninvasively measure plantar pressures, such as using capacitive pressure sensing fabrics (Shu et al., 2010) or estimate to COP, GRFs and ankle moments using combination load cells and pressure transducers (Jacobs and Ferris, 2015). These technologies are creating new possibilities for the real-time capture and implementation of the COP.

Conclusion

Providing visual biofeedback of COP trajectory via toe-off targeting effectively shifted average COP locations along the direction of each target. Changes in COP acting in the sagittal plane were controlled by temporal adjustments while those in the frontal plane were controlled with angle and moment changes at the ankle. A lack of changes in triceps surae EMG magnitudes likely signifies that muscle activation was also time dependent for sagittal plane shifts and that the triceps surae complex plays little role in frontal plane COP modification. The biofeedback modified joint loading largely as hypothesized with most compensations being observed in angles and moments at the ankle and small effects on more proximal joints. COP biofeedback should prove an efficacious tool for goal-oriented

tasks. For example, a shift medially of the COP demonstrated well-defined compensations at the in

ankle angle which may benefit individuals suffering from equinovarus or ankle instability as an

alternative option from bracing.

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