Reduction of Frontal-Plane Hip Joint Reaction Force Via Medio-Lateral Foot Center of Pressure Manipulation: A Pilot Study

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ABSTRACT: Footwear-generated biomechanical manipulation of lower-limb joints has been shown to influence lower-limb biomechanics. Numerous studies report the influence of such interventions on the knee, however little is known about the influence of these interventions on the hip. The present study analyzed kinetic and kinematic changes about the hip of 12 healthy young males who underwent biomechanical manipulation utilizing the APOS biomechanical device (APOS–Medical and Sports Technologies Ltd., Herzliya, Israel) allowing controlled foot center of pressure manipulation. Subjects underwent gait testing in four para-sagittal device configurations: Medial, lateral, neutral, and regular shoes. In the medial configuration, subjects demonstrated no change in step width (i.e., distance between right and left foot center of pressure), however inter-malleolar distance significantly increased. Likewise with the medial setting, greater hip abduction was recorded, while hip adduction moment and joint reaction force decreased significantly. We speculate that subjects adopt a modified gait pattern aimed to maintain constant base of support. As a result, hip abductor muscle moment arm increases and adduction moment and joint reaction force decreases. To the best of our knowledge this is the first study to show this relationship. These results contribute to the understanding of lower-limb biomechanics and warrant further investigation. © 2014 Orthopaedic Research Society. Published by Wiley Periodicals, Inc. J Orthop Res 33:261–269, 2015.

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Footwear-generated biomechanical manipulation (e.g., wedge insoles and foot orthotics), commonly used in clinical practice, has been the focus of vast biomechanical research and has been shown to alter lower-limb biomechanics and reduce joint loads in both healthy and pathological subjects.[1–8] These devices have been reported to shift the foot's center of pressure (COP) thus changing the locus and orientation of the resultant ground reaction force (GRF). They work on the principle that the lower limbs act as a system of chained links forming a functional kinetic unit.[9] Thus the effect of changing the COP is carried up the chain starting from the most proximal joint (i.e., ankle, knee, hip). By changing the locus and orientation of the GRF, and hence the perpendicular distance from the GRF to the center of the joint being investigated, the kinematics, kinetics, and neuromuscular activity about the joint are also affected.[1,7,8,10–12] This in turn may alter the loading conditions of the joint. Footwear-generated COP manipulation which displaces the COP in such a way to reduce joint loads via this mechanism has been the focus of vast research since recommendations for degenerative joint diseases include reducing load on the pathological joint.[13] However, in-depth studies of influence of such interventions on the hip joint, specifically with an eye toward clinical implications, are scarce and, in general, there is a void of biomechanical analyses, with specific respect to the hip joint, of footwear capable of manipulating lower-limb biomechanics.

Loads in the hip joint during gait have been measured to be 2 to over 5 times body weight.[14–17] These extreme loads may be detrimental to the joint, causing further damage, pain, and disease progression in the case of degenerative diseases.[18] The abductor muscles' force acting across the hip joint contributes a major component of the load on the hip joint.[19,20] In fact, it has been shown that the gluteus medius, a major hip abductor muscle, contributes the most to the vertical and medio-lateral components of the hip joint contact force out of all of the key muscles that span the hip joint, including abductors, adductors, flexors, extensors, and rotators, and also more than gravitational and centrifugal forces, as well as all other muscles of the ipsilateral limb.[21] In addition, it was shown that the gluteus medius and minimus, also a hip abductor, were among the muscles that contributed the most to the first peak of the total GRF and were nearly solely responsible for the midstance GRF.[22] It is also suggested that abductor muscles contribute more to the joint force than body weight.[23] According to the standard frontal-plane model of the hip-pelvis complex, in single limb stance, the hip acts as a fulcrum to keep the pelvis parallel to the ground (Fig. 1a). The moment about the hip produced by the body weight minus the weight of the ipsilateral leg (K*a) is counteracted by the moment produced by contraction of the abductor muscles (M*b). The effective body weight (K) plus the abductor muscles' force (M) produces a resultant joint reaction force (JRF).

Gait in the frontal, sagittal, and transverse planes occurs due to a balance or specific imbalance of internal and external forces acting about joints of the lower limbs. External forces about the joint result from the GRF, while internal forces result from active and passive anatomical structures including muscles, bones,
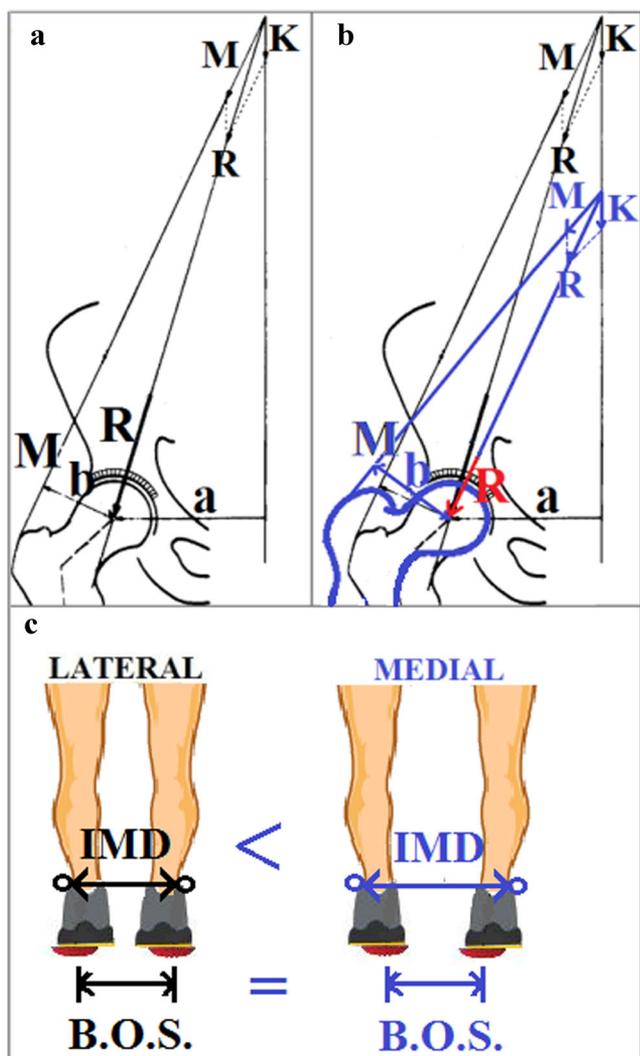
Figure 1. (a) Standard free body diagram of normal hip joint, (b) superimposed free body diagram (in blue) for medially shifted COP with demonstrated increased hip abduction, increased hip abductor muscle moment arm, decreased abductor muscle force, and decreased resultant JRF (illustration is exaggerated for visual clarity), and (c) demonstration of study finding in which subjects maintain constant base of support (distance between biomechanical elements' centers) for medial and lateral configurations while consequently increasing IMD for medial compared to lateral configuration. M = hip abductor muscle force, b = abductor muscle moment arm, K = body weight minus weight of ipsilateral limb, a = body weight moment arm, R = resultant joint reaction force.

and viscoelastic tissues. These forces produce moments that act about the joints. The magnitude of the external frontal, sagittal, and transverse-plane hip moments are equal to the magnitude of the vector component of the GRF in the plane of interest times the perpendicular frontal, sagittal, or transverse plane distance, respectively, from the GRF vector component to the center of hip rotation. Using three-dimensional gait analysis apparatus, such as a camera and force plate system, the GRF and associated kinematics can be measured. The kinetics can then be calculated from these measures using an inverse dynamics approach.[24,25] When

there is no movement, a balance of internal and external moments exists (i.e., net moment equals zero), while when movement occurs, there is an imbalance of moments (i.e., net moment is not equal to zero and at least one moment must be greater than its corresponding counteracting moment). Counteractive muscle pairs may create internal moments, which combined with the external moments, act in synchrony to allow for coordinated movement of the joint. This critical imbalance is what enables controlled human locomotion. Using an inverse dynamics approach, external hip moments and inter-segmental forces can be calculated. The calculations reveal approximate values since certain parameters, such as segment masses, gravitational forces, passive moments, etc., cannot, for all practical purposes, be accurately measured. It is assumed that these external kinetic parameters represent a reasonably close estimation of the true kinetics acting about the hip.

In a previous work by our group, we used a biomechanical apparatus which enabled controlled COP manipulation. We demonstrated that a measured alteration of the device lead to a shift in the COP trajectory during stance phase, which generated a reduction in knee medial-compartment joint loads and external knee adduction moment in both healthy[1] and knee osteoarthritis (OA) patients.[2] The present study was therefore devised to establish the outcome of this biomechanical intervention on the hip joint. Since the hip is a ball-and-socket joint as opposed to the knee, and has a very different function in gait, the theoretical framework for the present study was based on the aforementioned free body diagram and specific biomechanical principles of the hip joint.

Past research showed that step width (distance between right and left COP) remained constant during medio-lateral COP manipulation.[1] We therefore hypothesized that in order to maintain constant step width in a medial COP device configuration, internal-leolar distance (IMD) will increase by means of relative hip abduction. Hip abduction has been shown to increase the abductor muscles' moment arm, thus decreasing the amount of abductor muscle force required to balance body weight.[26] We thereby further hypothesized that the relative hip abduction would lead to a decrease in external hip adduction moment (due to decrease in abductor muscle force) and hence a decrease in hip JRF, in direct accordance with Figure 1a and b.

METHODS
Participants
Existing data from a previous study focusing on the knee joint was utilized for analysis with respect to the hip joint. The study group consisted of 12 healthy anthropometrically similar male subjects (Dominant leg = Right, Shoe size = French 43, Age = 25.7 ± 2.13 years, Height = 177 ± 3.8 cm, Mass = 73.3 ± 4.87 kg).[1] Subject exclusion criteria were any orthopedic, musculoskeletal, or neuromuscular pathology. Approval of the Ethics Sub-Committee was obtained and

informed consent was given by all participants. The purpose and methods of the study were explained to the subjects.

The Biomechanical System

The APOS biomechanical device (APOS System, APOS—Medical and Sports Technologies Ltd. Herzliya, Israel) was used. A detailed description of the device was previously reported.[1] In brief, COP manipulation is accomplished using a platform in the form of a shoe in which two adjustable convex-shaped biomechanical elements are attached to the feet by means of a shoe sole specially designed with two mounting rails (Fig. 2a). A medial or lateral shift of the elements along the medio-lateral foot axis causes a corresponding directional shift of COP.[1]

Experimental Protocol

Prior to the study, functional assessment of each subject was performed by a single physician (A. Haim). Subsequently, the functional neutral configuration (FNC) was custom-defined and documented by a single trained physiotherapist. The FNC was defined for each subject as the position of the elements in which the least varus, valgus, plantar, and dorsal torque was exerted by the apparatus about the ankle. The physiotherapist set the position of this configuration by observing the subjects' gait and making adjustments until he was satisfied that the proper positioning was achieved. The medial and lateral COP configurations were defined as 0.8-cm medial and 1.5-cm lateral deviations of the biomechanical elements from the neutral sagittal axis (Fig. 2b–d).
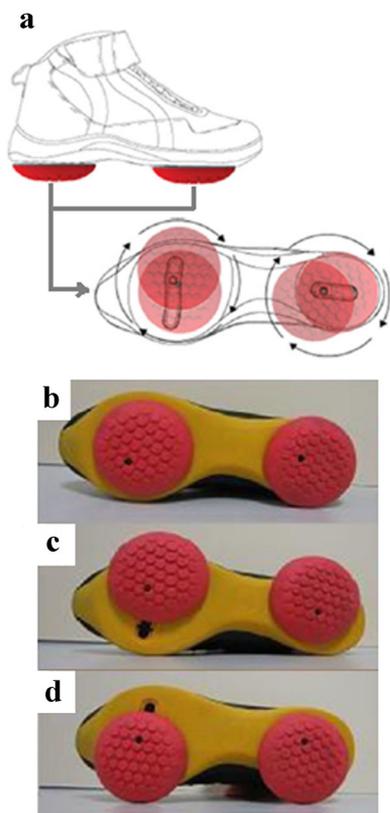
Figure 2.   (a) Biomechanical device with adjustable elements in (b) neutral, (c) lateral, and (d) medial configurations.

Subjects were given a several-minute period prior to data acquisition to walk at a comfortable self-selected speed in order to become accustomed to the shoes. During this period, a metronome was set to the subject-selected walking speed and was used during every gait trial in which data were collected to ensure constant cadence. After the accustomization period, gait analyses were performed in the four COP conditions—medial (M), neutral (N), lateral (L), and the biomechanical device with no biomechanical elements attached (NB) (similar to regular shoes)—at random order on the same day.

Data Acquisition and Processing

Three-dimensional motion analysis was performed using an 8-camera Vicon motion analysis system (Oxford Metrics Ltd., Oxford, UK) for kinematic data capture, at a sampling frequency of 120 Hz. GRFs were recorded by two 3-dimensional AMTI OR6–7-1000 force plates placed in tandem in the center of a 10-m walkway, at a sampling frequency of 960 Hz. Kinematic and kinetic data were collected simultaneously while subjects walked over the walkway. A standard marker set was used to define joint centers and axes of rotation.[27] A knee alignment device (KAD; Motion Lab Systems Inc., Baton Rouge, LA) was used to estimate three-dimensional alignment of the knee flexion axis during a static trial. Joint angles were calculated based on marker locations using 'PlugInGait' (Oxford Metrics Ltd., Oxford, UK), and joint forces and torques were calculated via 'PlugInGait' using inverse dynamic analyses from kinematic data and force plate measures. All analyses were performed for the dominant leg. Joint moments were normalized for body mass.

Various gait parameters were recorded with respect to each of the four different device configurations. The following parameters were calculated: The hip joint adduction angle, external adduction moment, frontal-plane hip JRF (i.e., the inter-segmental force between the thigh and pelvis segments in the link-segment model), and JRF angle (i.e., the angle formed by the frontal-plane JRF and the transverse pelvic plane). All aforementioned kinetic and kinematic parameters were calculated at 2 distinct time points during the stance phase—the first (load-bearing) and the second (push-off) peaks of the GRF. We elected to do so since the two peaks in GRF were easily identifiable in all gait trials of all subjects, and these peaks approximately coincided with the peaks in JRFs and adduction moment. In addition, the following spatio-temporal parameters were calculated: Step width, IMD, speed, and cadence. Step width was calculated as the distance between COPs of the right and left foot by means of the force plates, while IMD was calculated as the distance between the lateral ankle markers, located on the lateral malleoli, at peak 1 of the right and left foot GRFs for the same steps on the force plates.

Statistical Analysis

Data was analyzed by a biomedical statistician (A.Herman) using R© version 2.11.1 (Vienna, Austria).[28] The Wilcoxon signed-rank test was used as a paired test to compare each variable between different shoe component configurations (M, N, L, NB). Bivariate nonparametric correlation was done using the Spearman's correlation coefficient. All p-values reported are two-sided. A p-value below 0.05 is considered statistically significant.

RESULTS

Spatio-Temporal Parameters

Results for spatio-temporal parameters are shown in Table 1. Step width, speed, and cadence did not differ significantly between any of the four COP conditions. The IMD was 21% and 13% increased with M compared to N and NB, respectively, and 31% increased compared to L.

Hip Kinetics and Kinematics

Figure 3a–c shows representative graphs of hip adduction/abduction angle, adduction/abduction moment, and frontal-plane JRF versus percent stance phase, respectively, in the four different walking conditions. The adduction angle (Fig. 3a) is clearly reduced with M throughout most of stance phase, from loading response to the end of terminal stance. The adduction moment (Fig. 3b) is reduced around peak 1 for M, most markedly in the first half of midstance. The frontal-plane JRF (Fig. 3c) is also clearly reduced in magnitude in the first half of midstance.

Results for values of the kinetic and kinematic parameters tested in the different device configurations, recorded at the time of 1st and 2nd peaks of the GRF, are listed in Tables 2 and 3, respectively. On average, the hip adduction angle at peak 1 was significantly reduced by 29% with M configuration compared to N, reduced 32% with M compared to NB, and reduced 37% with M compared to L. Correspondingly magnitude of the adduction moment was significantly reduced by 14% with M configuration compared to N, reduced 8% with M compared to NB, and reduced 12% with M compared to L. The frontal-plane JRF was significantly reduced by 8% with M configuration compared to both N and L. The angle between the resultant frontal-plane JRF and the horizontal pelvis line (Hilgenreiner's line) was significantly increased (JRF became more vertical), although only 1% and 2% with M compared to NB and L, respectively, whereas the angle was 1% significantly decreased with L compared to N.

At peak 2 of the GRF, on average, the adduction moment was significantly reduced by 7% with M compared to L only. Contrary to peak 1, the JRF at peak 2 was reduced with L compared to N, but only by 2%. The adduction angle at peak 2 did not significantly differ between any of the four COP
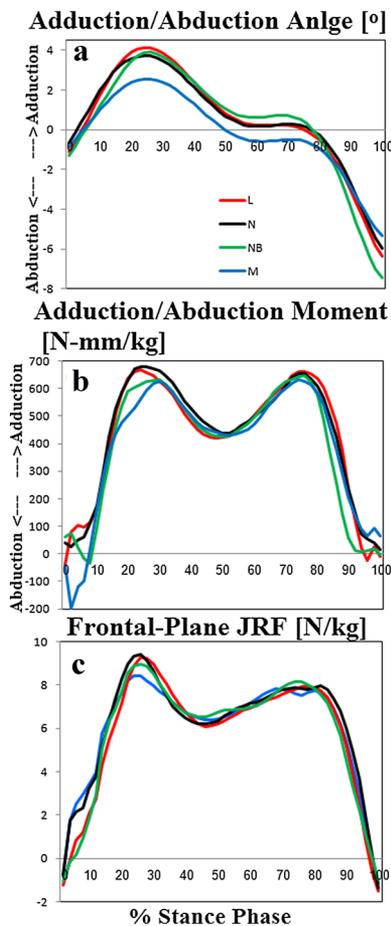
Figure 3. Representative graph of (a) adduction/abduction angle, (b) adduction/abduction moment, and (c) frontal-plane JRF for all four walking conditions versus percent stance phase.

conditions. The JRF angle was significantly increased 1% with M compared to L, while it was significantly decreased 1% and 2% with L compared to N and NB, respectively.

Table 4 shows results for statistical correlations between peaks 1 and 2 of the adduction moment and adduction/abduction angle and the magnitude of the resultant frontal-plane JRF. Peaks 1 and 2 of the adduction moment showed significant positive correlations to their associated adduction angle and JRF. Adduction angle was positively correlated to the JRF only at peak 2.

Table 1. Comparison of Spatio-Temporal Parameters ($n = 12$)

COP Position	M	N	L	NB
Spatio-temporal				
Step width [m]	0.16 (0.03)	0.16 (0.04)	0.17 (0.03)	0.15 (0.03)
IMD [m]	0.17 (0.34)	0.14 (0.30)[a(p = 0.005)]	0.13 (0.23)[a(p = 0.001)]	0.15 (0.31)[b(p = 0.003) a(p = 0.042)]
Speed [m/s]	1.13 (0.15)	1.13 (0.16)	1.13 (0.20)	1.13 (0.16)
Cadence [steps/min]	100.9 (8.62)	100.1 (8.99)	102.0 (7.64)	100.2 (10.5)

Mean (standard deviation) of spatio-temporal parameters ($n = 12$); [a] $= p$ value <0.05 for comparison with medial element position; [b] $= p$ value <0.05 for comparison with lateral element position.

Table 2.   Comparison of Hip Kinetics and Kinematics at Peak 1 of GRF ($n = 12$)

COP Position	M	N	L	NB
Kinetics				
Hip Adduction Moment at Peak 1 [N-mm/kg]	630.73 (106.3)	734.47 (123.6)[a($p = 0.012$)]	716 (123.4)[a($p = 0.002$)]	685.92 (82.6)[a($p = 0.042$)]
Magnitude of Resultant Frontal-Plane JRF at Peak 1 [N/kg]	7.85 (0.68)	8.49 (0.66)[a($p = 0.009$)]	8.53 (0.39)[a($p = 0.005$)]	8.24 (0.77)
Kinematics				
Hip Adduction Angle at Peak 1 [Degrees]	2.53 (3.00)	3.54 (3.00)[a($p = 0.005$)]	4.03 (3.22)[a($p = 0.002$)]	3.7 (3.11)[a($p = 0.003$)]
Angle between Resultant Frontal-Plane JRF at Peak 1 and the Horizontal [Degrees]	87.19 (2.19)	86.71 (2.02)[b($p = 0.021$)]	85.53 (2.75)[a($p = 0.001$)]	86.19 (1.56)[a($p = 0.027$) c($p = 0.034$)]

Mean (standard deviation) of kinetic and kinematic parameters associated with peak 1 of GRF ($n = 12$); [a] $= p$ value <0.05 for comparison with medial element position; [b] $= p$ value <0.05 for comparison with lateral element position; [c] $= p$ value <0.05 for comparison with neutral element position.

DISCUSSION

The results presented establish a quantitative relationship between footwear-generated COP manipulation along the medio-lateral foot axis and hip joint kinetics and kinematics. To the best of our knowledge such quantitative association has not been previously reported. In accordance with the hypothesis, with the medial condition of the apparatus the inter-malleolar distance was increased and step width as measured by the force plates remained constant (Fig. 1c). This increase was accompanied by increased relative hip abduction (decreased adduction angle). Likewise, with medial transposition of the elements, a significantly lower hip adduction moment and frontal-plane JRF were recorded during stance phase. This is in agreement with a previous study which showed that maximum isometric hip adduction moment decreases with increasing hip abduction angle.[29] Hip abductor muscles' moment arm has been shown to increase continually from 30 degrees of hip adduction to 45

Table 3.   Comparison of Hip Kinetics and Kinematics at Peak 2 of GRF ($n = 12$)

COP Position	M	N	L	NB
Kinetics				
Hip Adduction Moment at Peak 2 [N-mm/kg]	629.37 (159.2)	676.19 (149)	677.95 (165.68)[a($p = 0.009$)]	669.48 (135.25)
Magnitude of Resultant Frontal-Plane JRF at Peak 2 [N/kg]	8.02 (0.56)	8.19 (0.45)[b($p = 0.021$)]	7.99 (0.41)	8.00 (0.61)
Kinematics				
Hip Adduction Angle at Peak 2 [Degrees]	−0.42 (3.46)	0.13 (3.19)	0.03 (3.63)	0.53 (3.18)
Angle between Resultant Frontal-Plane JRF at Peak 2 and the Horizontal [Degrees]	85.66 (3.72)	85.48 (2.97)[b($p = 0.042$)]	84.41 (3.63)[a($p = 0.001$)]	86.31 (3.39)[b($p = 0.021$)]

Mean (standard deviation) of kinetic and kinematic parameters associated with peak 2 of GRF ($n = 12$); [a] $= p$ value <0.05 for comparison with medial element position; [b] $= p$ value <0.05 for comparison with lateral element position.

Table 4. Correlations Between Hip Kinetic and Kinematic Parameters ($n = 12$)

	Adduction Angle @ Peak 1	Frontal-Plane JRF @ Peak 1
Adduction moment peak1	0.5 ($p = 0$)	0.39 ($p = 0.006$)
Adduction angle @ peak 1		0.057 ($p = 0.703$)
	Adduction angle @ peak 2	**Frontal-plane JRF @ peak 2**
Adduction moment peak 2	0.298 ($p = 0.04$)	0.513 ($p = 0$)
Adduction angle @ peak 2		0.362 ($p = 0.011$)

Correlations r (p) between selected kinetic and kinematic parameters recorded at peaks 1 and 2 of GRF.

degrees of abduction.[26] This could clarify the above findings, where relative hip abduction shifts the femoral insertion of the hip abductors to a more lateral position, thus increasing the moment arm of these muscles, resulting in a decrease in the external hip adduction moment required to counteract the body weight moment and maintain the pelvis parallel to the ground (Fig. 1b). A decrease in hip abductor muscle force acting about the hip joint, as seen by the decrease in hip adduction moment, would therefore decrease the frontal-plane JRF. We speculate that subjects in this study adopted a gait strategy intended to maintain a constant effective base of support (i.e., the horizontal distance between the biomechanical elements, measured as step width using the force plates). When the elements where shifted medially, subjects assumed a more abducted hip gait in order to maintain base of support, and as a result the hip JRF was significantly reduced.

Another important finding was that medial transposition of the biomechanical elements caused the frontal-plane JRF vector to become more vertical. This phenomenon may have occurred because of the more neutral angle (adduction angle approaching closer to zero degrees) with the medial configuration. In this case, the more neutral angle may have reduced the forces being transferred along the horizontal axis in conjunction with the reduction in abductor muscles' force. It must be noted, however, that the change in the JRF angle between device configurations, although statistically significant, was very small and may not be substantial.

Hip joint OA is one of the most common joint disorders in the United States and in Europe and is a leading cause of disability in the elderly, with more than 200,000 total hip replacements performed annually in the United States, according to the American Association of Orthopaedic Surgeons. Although the pathogenesis of OA is not completely understood, there is substantial evidence that biomechanical factors play a pivotal role in disease progression.[20,30] Compressive and shear stresses on a joint may be a major mechanism for mechanical failure of cartilage as seen in degenerative diseases.[31,32] Nevertheless, for the purpose of treating hip OA, advice concerning appropriate footwear is based on expert opinion alone.[13] In this study, walking with the medial configuration of the

apparatus significantly reduced frontal-plane JRF. Moreover, this intervention also significantly reduced peak 1 of the adduction moment. In a cohort of subjects scheduled for total hip replacement, peak hip adduction moment was associated with an accelerated joint space narrowing of the contralateral hip.[33] Peak hip adduction moment was also suggested to be a major determinant of the peak hip contact force.[34] This further emphasizes that a COP displacement which reduces adduction moment and joint loads may also lead to a slowing or stopping of joint space narrowing in hip OA patients. In this study, however, due to the scarcity of studies validating the biomechanical factors of cartilage degeneration in the hip, we did not use the observed reduction in adduction moment as the main outcome of the COP manipulation, but rather the reduction in JRF. It has been established that OA in general, and hip OA in particular, has onset/progression associated with excessive joint loads.[20,30,35–39] We therefore used the adduction moment as an intermittent parameter to explain our observed reduction in JRF. Under a quasistatic assumption during stance phase, the external adduction moment is equal and opposite to the internal abductor muscles' moment. Thus if, as we suggested, abductor muscle force is reduced in the medial condition, this would be reflected by a reduction in the external adduction moment as was observed. We thus suggest that a reduction in abductor muscle force, as seen by the reduction in external adduction moment, contributed to the reduction in JRF that we observed.

Footwear-generated manipulation of joint mechanics by means of wedged insoles, which essentially shift the COP along the medio-lateral foot axis, is a common and accepted therapy for knee OA. They may reduce pathologically enhanced moments about the knee, thereby reducing load on the diseased joint compartment.[40,41] However, shoe wedges revealed no effect on hip kinetics or kinematics in knee OA patients[42] and healthy subjects.[12] In contrast, the present study revealed significant impact on hip joint kinetics and kinematics with medio-lateral COP manipulation. This may be due to a more extreme displacement of COP compared to wedges. The degree of displacement of COP using the device of our study can be adjusted continuously in a large portion of the transverse plane of the foot, whereas the degree of displacement using

wedges is constant and determined by the angular incline of the insole. Additionally, the convexity of the elements attached to the shoe sole of the device of this study induces an unstable platform that the user must adapt to. In order to maintain stability, the device may "force" the user to walk in such a way that he/she must adopt the new COP along with the associated altered gait and neuromuscular pattern. We believe that the unique design of the device which merges both a custom-positioned COP along with an element of instability may be the reason that we were successful in showing statistically significant results, where the studies using shoe wedges failed to do so.

Several limitations arising from the current study should be noted. Firstly, employment of the apparatus with no elements attached was defined as a control. This setting was elected to ensure consistency of the kinematic model (biomechanical elements were attached and modulated without repositioning of the retro-reflective markers). The flat rather than curved bottom presents a more stable contact interface and could result in lesser demands on the neuromuscular system to balance. To limit this potential bias, the neutral configuration was used as a reference and a secondary control for evaluation of the medial and lateral configurations. Nevertheless, it should be noted that the COP in the neutral axis was medially deviated by 6.27 mm with respect to the control.[1] In addition, it should be emphasized that the participants in this study comprised a distinctive homogeneous cohort (i.e., healthy, young male adults). These results are therefore valid only for individuals with characteristics similar to those of the tested group. Further studies are needed before these findings can be extended to other populations. In addition, the absolute difference of the presented gait parameters between COP conditions in some cases, though statistically significant, was relatively small. We demonstrated statistical significance of hip kinetics and kinematics of the four walking conditions measured at peak 1 of the GRF, whereas there was relatively minor significance at peak 2. Previous studies have reported greater variability of gait parameters at peak 2, which may account for this.[1,43] Additionally, we acknowledge that our small sample size may present a limitation. Despite this, we found statistical significance between COP conditions of the tested parameters. In addition, patients served as their own controls for each walking condition, while the test setting did not change (i.e., markers remained fixed and were not moved and testing was performed on the same day). We believe that this further validates our results despite the small sample size, as we eliminated factors such as marker movement, between-day variability, and variability among different control and test subjects. Past studies used to analyze hip kinetics and kinematics in different walking conditions have used similar sample sizes.[10,44–47] We also acknowledge that anatomical variations in the neck of the femur among subjects would likely impact the hip kinetics. This is a bias shared by all gait analysis studies of this type. Again, since subjects served as their own control we believe our results show the true relative relationship between COP manipulation and hip joint kinetics and kinematics. We also acknowledge that positioning of the FNC for the neutral control position was subjective with respect to the physiotherapist who positioned the elements of the device based on observational gait analysis to achieve this condition. Nevertheless, since each subject served as their own control for the different walking conditions we tested, we believe that if there was a small error in the positioning of the FNC, it would not greatly affect the relationship of the tested parameters in the control compared to the medial and lateral COP configurations. Lastly, the use of a straightforward inverse dynamics approach to calculate external joint forces, described in the literature in detail,[24,25,48–50] although very commonly used in gait analysis studies of lower-limb joints,[43,51–54] presents a well-known limitation. This method likely underestimates the JRF to some extent since muscle and soft tissue dynamics are not taken into account. Several studies have attempted to use musculoskeletal models with a combined inverse dynamics approach to estimate these dynamics, however these models are also not without limitations as they are frequently unvalidated, rely on many assumptions, and have been shown in many instances to substantially overestimate joint forces.[55] Regardless, we are confident that any error incurred in our results due to assumptions made by the inverse dynamics model would be adequately controlled for by our experimental protocol since subjects served as their own controls in the manner aforementioned. We thus expect that, despite this limitation, the outcome of our study in which a medially displaced COP reduces the frontal-plane hip joint force would not be affected by a potential underestimation of the joint force.

Finally, it is necessary to declare that we do not intend for the reader to draw any immediate clinical conclusions from the results of this study. This study was strictly a pilot study whose objective was to analyze the relationship between specific COP displacements in the frontal plane and the resulting changes in kinematics and kinetics of the hip, and to determine if these changes would be statistically significant. We intend that the statistical significance found for our results illuminate certain aspects of the effect of footwear-generated biomechanical manipulation of the hip joint on biomechanical parameters of the hip, and be used as a foundation for further research, including research in a pathological cohort. Although we have exhibited that a medial COP reduces the frontal-plane JRF in our study cohort, this result must first be confirmed in a cohort of subjects with degenerative hip disease before making the claim of clinical implications. Nevertheless, the results presented are still valid and could contribute

to the knowledge in the field of biomechanics and footwear design.

ACKNOWLEDGEMENT

REFERENCES

1. Haim A, Rozen N, Dekel S, et al. 2008. Control of knee coronal plane moment via modulation of center of pressure: a prospective gait analysis study. J Biomech 41:3010–3016.
2. Haim A, Wolf A, Rubin G, et al. 2011. Effect of center of pressure modulation on knee adduction moment in medial compartment knee osteoarthritis. J Orthop Res 29:-1668.
3. Haim A, Rubin G, Rozen N, et al. 2012. Reduction in knee adduction moment via non-invasive biomechanical training: a longitudinal gait analysis study. J Biomech 45:41–45.
4. Erhart JC, Mündermann A, Elspas B, et al. 2008. A variable-stiffness shoe lowers the knee adduction moment in subjects with symptoms of medial compartment knee osteoarthritis. J Biomech 41:2720–2725.
5. Erhart JC, Dyrby CO, D'Lima DD, et al. 2010. Changes in in vivo knee loading with a variable-stiffness intervention shoe correlate with changes in the knee adduction moment. J Orthop Res 28:1548–1553.
6. Erhart JC, Mündermann A, Elspas B, et al. 2010. Changes in knee adduction moment, pain, and functionality with a variable-stiffness walking shoe after 6 months. J Orthop Res 28:873–879.
7. Crenshaw SJ, Pollo FE, Calton EF. 2000. Effects of lateral-wedged insoles on kinetics at the knee. Clin Orthop Relat Res 375:185–192.
8. Shelburne KB, Torry MR, Steadman JR, Pandy MG. 2008. Effects of foot orthoses and valgus bracing on the knee adduction moment and medial joint load during gait. Clin Biomech 23:814–821.
9. Zajac FE, Neptune RR, Kautz SA. 2002. Biomechanics and muscle coordination of human walking. Part I: introduction to concepts, power transfer, dynamics and simulations. Gait Posture 16:215–232.
10. Haim A, Rozen N, Wolf A. 2010. The influence of sagittal center of pressure offset on gait kinematics and kinetics. J Biomech 43:969–977.
11. Goryachev Y, Debbi EM, Haim A, Wolf A. 2011. The effect of manipulation of the center of pressure of the foot during gait on the activation patterns of the lower limb musculature. J Electromyogr Kinesiol 21:333–339.
12. Nester CJ, van der Linden ML, Bowker P. 2003. Effect of foot orthoses on the kinematics and kinetics of normal walking gait. Gait Posture 17:180–187.
13. Zhang W, Moskowitz RW, Nuki G, et al. 2008. OARSI recommendations for the management of hip and knee osteoarthritis, part II: OARSI evidence-based, expert consensus guidelines. Osteoarthritis Cartilage 16:137–162.
14. Bergmann G, Graichen F, Rohlmann A. 1993. Hip joint loading during walking and running, measured in two patients. J Biomech 26:969–990.
15. Bergmann G, Deuretzbacher G, Heller M, et al. 2001. Hip contact forces and gait patterns from routine activities. J Biomech 34:859–871.
16. Davy DT, Kotzar GM, Brown RH, et al. 1988. Telemetric force measurements across the hip after total arthroplasty. J Bone Joint Surg Am 70:45–50.
17. Kotzar GM, Davy DT, Goldberg VM, et al. 1991. Telemeterized in vivo hip joint force data: a report on two patients after total hip surgery. J Orthop Res 9:621–633.
18. Andriacchi TP, Mündermann A, Smith RL, et al. 2004. A framework for the in vivo pathomechanics of osteoarthritis at the knee. Ann Biomed Eng 32:447–457.
19. Neumann DA, Cook TM. 1985. Effect of load and carrying position on the electromyographic activity of the gluteus medius muscle during walking. Phys Ther 65:305–311.
20. Sims K. 1999. The development of hip osteoarthritis: implication for conservative management. Man Ther 4: 127–135.
21. Correa TA, Crossley KM, Kim HJ, Pandy MG. 2010. Contributions of individual muscles to hip joint contact force in normal walking. J Biomech 43:1618–1622.
22. Anderson FC, Pandy MG. 2003. Individual muscle contributions to support in normal walking. Gait Posture 17: 159–169.
23. Krebs DE, Robbins CE, Lavine L, Mann RW. 1998. Hip biomechanics during gait. J Orthop Sports Phys Ther 28: 51–59.
24. Andriacchi TP, Strickland AB. 1985. Lower limb kinetics applied to the study of normal and abnormal walking. In: Berme N Engin AE Correia Da Silva KM, editor. *Biomechanics of Normal and Pathoiogical Human Articulating Joints*. NATO SI Series. pp 83–102.
25. Andriacchi TP, Natarajan RN, Hurwitz DE. 1997. Musculoskeletal dynamic locomotion and clinical applications. Basic Orthopaedic Biomechuniq, 2nd edition. Philadelphia. Lippincott-Raven. 37–69.
26. Henderson ER, Marulanda GA, Cheong D, et al. 2011. Hip abductor moment arm – a mathematical analysis for proximal femoral replacement. J Orthop Surg Res 6:-6.
27. Kadaba MP, Ramakrishnan HK, Wootten ME. 1990. Measurement of lower extremity kinematics during level walking. J Orthop Res 8:383–392.
28. R Development Core Team. 2010. *R: A language and environment for statistical computing*. Vienna, Austria: R Foundation for Statistical Computing http://www.R-project.org.
29. Delp SL, Maloney W. 1993. Effects of hip center location on the moment-generating capacity of the muscles. J Biomech 26:485–499.
30. Horak Z, Kubovy P, Stupka M, Horakova J. 2011. Biomechanical factors influencing the beginning and development of osteoarthritis in the hip joint. Wien Med Wochenschr 161: 486–492.
31. Mansour JM. 2003. Biomechanics of cartilage. In: Oatis CA editor., *Kinesiology: the mechanics and pathomechanics of human movement*. Philadelphia: Lippincott Williams and Wilkins. pp 66–79.
32. Radin EL, Burr DB, Caterson B, et al. 1991. Mechanical determinants of osteoarthrosis. Semin Arthritis Rheum 21: 12–21.
33. Hurwitz DE, Sumner DR, Block JA. 2001. Bone density, dynamic joint loading and joint degeneration: a review. Cells Tissues Organs 169:201–209.
34. Foucher KC, Hurwitz DE, Andriacchi TP. 1999. *Gait adaptations in postoperative total hip replacement patients are associated with reduced hip contact forces*. San Diego: American Society of Biomechanics pp 134–135.
35. Frosi G, Sulli A, Testa M, Cutolo M. 2001. [Physiopathology and biomechanics of hip osteoarthritis]. Reumatismo 53: 271–279.
36. Egloff C, Hügle T, Valderrabano V. 2012. Biomechanics and pathomechanisms of osteoarthritis. Swiss Med Wkly 142: w13583.
37. Felson DT, Lawrence RC, Dieppe PA, et al. 2000. Osteoarthritis: new insights. Part 1: the disease and its risk factors. Ann Intern Med 133:635–646.

38. Arden N, Nevitt MC. 2006. Osteoarthritis: epidemiology. Best Pract Res Clin Rheumatol 20:3–25.
39. Griffin TM, Guilak F. 2005. The role of mechanical loading in the onset and progression of osteoarthritis. Exerc Sport Sci Rev 33:195–200.
40. Hinman RS, Bowles KA, Bennell KL. 2009. Laterally wedged insoles in knee osteoarthritis: do biomechanical effects decline after one month of wear. BMC Musculoskelet Disord 10:-146.
41. Rodrigues PT, Ferreira AF, Pereira RM, et al. 2008. Effectiveness of medial-wedged insole treatment for valgus knee osteoarthritis. Arthritis Rheum 59:603–608.
42. Butler RJ, Barrios JA, Royer T, Davis IS. 2009. Effect of laterally wedged foot orthoses on rearfoot and hip mechanics in patients with medial knee osteoarthritis. Prosthet Orthot Int 33:107–116.
43. Hurwitz DE, Ryals AB, Case JP, et al. 2002. The knee adduction moment during gait in subjects with knee osteoarthritis is more closely correlated with static alignment than radiographic disease severity, toe out angle and pain. J Orthop Res 20:101–107.
44. Opila-Correia KA. 1990. Kinematics of high-heeled gait. Arch Phys Med Rehabil 71:304–309.
45. Romkes J, Rudmann C, Brunner R. 2006. Changes in gait and EMG when walking with the Masai Barefoot Technique. Clin Biomech 21:75–81.
46. Nigg B, Hintzen S, Ferber R. 2006. Effect of an unstable shoe construction on lower extremity gait characteristics. Clin Biomech 21:82–88.
47. Ajemian S, Thon D, Clare P, et al. 2004. Cane-assisted gait biomechanics and electromyography after total hip arthroplasty. Arch Phys Med Rehabil 85:1966.
48. Crowninshield RD, Johnston RC, Andrews JG, Brand RA. 1978. A biomechanical investigation of the human hip. J Biomech 11:75–85.
49. Soutas-Little RW. 1998. Motion analysis and biomechanics. In: DeLisa JA , editor. *Gait Analysis in the Science of Rehabilitation*. Baltimore, MD: Department of Veteran Affairs. pp 49–68.
50. Winter DA. 2009. *Biomechanics and Motor Control of Human Movement, 4th ed.*. Hoboken, NJ: John Wiley & Sons, Inc..
51. Hurwitz DE, Hulet CH, Andriacchi TP, et al. 1997. Gait compensations in patients with osteoarthritis of the hip and their relationship to pain and passive hip motion. J Orthop Res 15:629–635.
52. Hurwitz DE, Ryals AR, Block JA, et al. 2000. Knee pain and joint loading in subjects with osteoarthritis of the knee. J Orthop Res 18:572–579.
53. Sell TC, Ferris CM, Abt JT, et al. 2007. Predictors of proximal tibia anterior shear force during a vertical stop-jump. J Orthop Res 25:1589–1597.
54. Nagura T, Dyrby CO, Alexander EJ, Andriacchi TP. 2002. Mechanical loads at the knee joint during deep flexion. J Orthop Res 20:881–886.
55. Brand RA, Pedersen DR, Davy DT, et al. 1994. Comparison of hip force calculations and measurements in the same patient. J Arthroplasty 9:45–51.