Reduction of Hip Joint Reaction Force via Medio-lateral Foot Center of Pressure Manipulation in Bilateral Hip Osteoarthritis Patients

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ABSTRACT

Loading/excessive loading of the hip joint has been linked to onset and progression of hip osteoarthritis. Footwear-generated biomechanical manipulation in the frontal plane has been previously shown in a cohort of healthy subjects to cause a specific gait adaption when the foot center of pressure trajectory was shifted medially, which thereby significantly reduced hip joint reaction force. The objective of the present study was to validate these results in a cohort of female bilateral hip osteoarthritis patients. Sixteen patients underwent gait analysis while using a footworn biomechanical device, allowing controlled foot center of pressure manipulation, in three para-sagittal configurations: medial, lateral, and neutral. Hip osteoarthritis patients exhibited similar results to those observed in healthy subjects in that a medial center of pressure led to an increase in inter-maleolar distance while step width (i.e., distance between right and left foot center of pressure) remained constant. This adaptation, which we speculate subjects adopt to maintain base of support, was associated with significantly greater hip abduction, significantly decreased hip adduction moment, and significantly reduced joint reaction force compared to the neutral and lateral configurations.

Clinical significance: Recommendations for treatment of hip osteoarthritis emphasize reduction of loads on the pathological joint(s) during daily activities and especially in gait. Our results show that a medially deviated center of pressure causes a reduction in hip joint reaction force. The present study does not prove, but rather suggests, clinical significance and further investigation is required to assess clinical implications.

Keywords:

Center of pressure; Footwear-generated biomechanical manipulations; Gait analysis; Hip osteoarthritis; Frontal-plane kinetics and kinematics of the hip

INTRODUCTION

Hip osteoarthritis (OA) is one of the most common pathologies affecting the elderly with an immense social, economic, and personal burden. It is a chronic debilitating progressive disease characterized by pain, stiffness, loss of articular cartilage and joint space narrowing, formation of osteophytes, and significant gait and physical function abnormalities. It has been estimated by several epidemiologic studies to affect 6.7-9.7 % of people over the age of 45 in the United States.[1,2] As a result of increasing life expectancy and the obesity crisis, the need for total hip arthroplasty (THA) is expected to grow 174%, to 572,000 per year by 2030 in the United States alone, with actual numbers to date suggesting that this is an underestimation.[3]

Although the precise pathogenesis of OA is unknown, based on vast research, biomechanical factors play a critical role in its onset and progression.[4,5] Specifically, excessive loading of the osteoarthritic joint may be detrimental.[6] Mechanical failure of cartilage is caused by compressive and shear stresses on the joint.[7,8] Thus, among the recommended non-pharmacologic and non-surgical treatments for hip OA, reduction of joint load in gait and daily activities is emphasized.[9]

Footwear-generated biomechanical manipulation of lower limbs has been the focus of vast research. This manipulation shifts the foot's center of pressure trajectory, thus changing the locus and orientation of the ground reaction force. This affects biomechanics of all joints in the lower limbs starting with the ankle and progressing to the knee and hip.[10-14] Our previous research has shown that external knee adduction moment and medial-compartment joint loads are reduced in the knee in both healthy[10] and medial compartment knee OA patients[15]. Recently, in a pilot study conducted on a cohort of healthy males, we used a novel biomechanical device capable of controlled foot center of pressure (COP) manipulation to examine the effects of medio-lateral COP displacement on kinematic and kinetics of the hip joint.[12] We showed that hip joint reaction force is significantly reduced with a medial displacement of COP in this cohort. Subjects maintained a constant step width (distance between right and left foot COP) during medio-lateral COP manipulation, while increasing or decreasing intermaleolar distance (distance between lateral maleoli), in order to maintain a constant base of support (Figure 1(c)). With a medially displaced COP, subjects increased inter-malleolar distance (IMD) via increasing hip abduction. Also observed was a concurrent decrease in external hip adduction moment, as well as an 8 % decrease in frontal-plane hip joint reaction force (JRF) at the peak load-bearing portion of stance phase. We speculated that the mechanism for the decrease in frontal-plane JRF was as follows: medial shift in COP → increase in IMD/hip abduction in order to maintain base of support→ suggested increase in abductor muscle moment arm[16] and hence decrease in abductor muscle force required to maintain level pelvis → decrease in external hip adduction moment signifying decrease in abductor muscle force → decrease in frontal-plane hip JRF (Figure 1(a&b)).

The objective of the present study is to validate the results of our pilot study in a cohort of female bilateral hip osteoarthritis patients. We hypothesize that medial and lateral configurations of the biomechanical device elements will shift the COP trajectory accordingly. Further, we hypothesize that JRF will be decreased in the same manner exhibited in healthy subjects. Specifically, we hypothesize that a medial translation of the COP will cause an increase in IMD, while maintaining the same step width, in order to maintain base of support, an increase in hip abduction (decrease in adduction), a decrease in external hip adduction moment, and a resulting decrease in frontal-plane hip JRF. We further hypothesize that a medial COP configuration will increase single support duration and decrease double support duration signifying a decrease in pain due to decreased joint loading.

METHODS

Participants

Sixteen females with bilateral hip osteoarthritis (Age=63.5±6.3 yrs, Height=159.7±56.1 cm, Body mass=73.3±17.5 kg, K-L grade=2.8±0.5) were recruited from a cohort of patients enrolled in a clinical trial in the Department of Orthopedics at Rambam Medical Center, Haifa, Israel and Ha'Emek Medical Center, Afula Israel, investigating the longitudinal effects of biomechanical training with COP modulation on gait kinetics and kinematics. Data for the present study was collected before initiation of the clinical trial. All patients had symptomatic bilateral hip OA according to the American College of Rheumatology criteria for hip OA, with radiographic evidence of Kellgren-Lawrence grade 2 or greater.[17,18] Exclusion criteria were any orthopedic, musculoskeletal, or neurological pathology, previous surgery of the back and lower limbs, any other co-morbidities affecting the back and lower limbs, and use of a walking aid. Approval of the Ethics Sub-Committee was obtained and informed consent was given by all participants. The study was registered in the NIH clinical trial registration system (NCT01450254). 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.[10] 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 (Figure 2(a)).[19] The convex elements can be moved in a continuous fashion in the transverse plane of the foot.

Experimental Protocol

The study is a case series with level of evidence 4. The experimental protocol used in the study is consistent with that outlined in detail in our previous studies using the biomechanical device.[10-12,15] 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.[10,19] The physiotherapist set the position of this configuration by observing the subjects' gait and making adjustments until she 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 (Figure 2(c-d)).

Subjects were given a several-minute period prior to data acquisition to walk at a comfortable self-selected speed in the biomechanical device in order to become accustomed. After the accustomization period, gait analyses were performed in the three COP conditions – medial (M), neutral (N), and lateral (L) - 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 100 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 1000 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.[20] 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. Joint moments and forces were normalized for body mass.

Various gait parameters were recorded with respect to each of the 3 device configurations. Parameters are reported for both the more symptomatic leg (MS-Leg), as reported by the patient, and the less symptomatic leg (LS-Leg). The two distinct time points of the two peaks of the frontal-plane ground reaction force (FGRF) during stance phase were found. Peak 1 of the FGRF represents load-bearing, while peak 2 represents push-off. We elected to calculate the gait parameters discussed below at these 2 time points since visual examination of all of the data revealed that the two peaks in FGRF were identifiable in all gait trials of all subjects, and these peaks closely coincided with peaks in the JRFs and adduction moment. The COP offset was calculated as the perpendicular distance from the GRF coordinates on the force plates to the neutral sagittal foot axis (line connecting the toe and heel markers). The offset was then reported as the COP offset in the M and L conditions from the N condition at the time of peaks 1 and 2 of the FGRF, as well as the average COP offset from N during the entire stance phase. Additionally, the following parameters were calculated at peaks 1 and 2 of the FGRF: 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). In addition, the following spatio-temporal parameters were calculated: step width, inter-malleolar distance (IMD), single support duration, double support duration, 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

The Wilcoxon signed-rank test was used as a paired test to compare each variable between different shoe component configurations (M, N, and L). A p-value below 0.05 is considered statistically significant.

RESULTS

COP Parameters

Results for COP parameters are shown in Table 1. The average stance phase COP, COP at peak 1 of FGRF, and COP at peak 2 of FGRF were significantly offset in both the MS-Leg and LS-Leg in all 16 subjects from that in the neutral device configuration. The offset for M and L was in the direction corresponding to the shift in the biomechanical device elements.

Spatio-temporal Parameters

Results for spatio-temporal parameters are shown in Table 2. Step width was 9 % increased on average with L compared to N. IMD was increased 7 % with M compared to N and 10 % compared to L. Single support duration was increased 2 % with M compared to L for the MS-Leg. Double support duration was decreased 4 % with M compared to L. Speed and cadence did not differ significantly between the 3 biomechanical device configurations.

Hip Kinetics and Kinematics

Figure 3(a-c) shows a subject's representative graphs of hip adduction/abduction angle, adduction/abduction moment, and frontal-plane JRF vs. percent stance phase, respectively, in the 3 different walking conditions for the MS-Leg. For the particular subject, there is an evident reduction in adduction angle (increase in abduction) (Figure 3(a)) with the M configuration from around the middle of loading response throughout terminal stance. The adduction moment and frontal-plane JRF (Figure 3(b,c)) follow a similar pattern around the time of the first peak in which there is a reduction in these parameters with the M configuration from around the end of loading response to around the middle of midstance.

From this point, there is a slight reduction in adduction moment with M throughout most of terminal stance.

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 FGRF, are listed in Table 3 and 4, respectively. On average, the hip adduction angle at peak 1 was significantly reduced with the M configuration for the MS-Leg by 21 % compared to N and 19 % compared to L, as well as for the LS-Leg by 26 % compared to N and 22 % compared to L. Correspondingly magnitude of the adduction moment was significantly reduced with the M configuration for the MS-Leg and LS-Leg by 5 % and 8 %, respectively, compared to N. The frontal-plane JRF was significantly reduced with the M configuration for the MS-Leg by 2 % compared to N and for the LS-Leg by 2 % and 3 %, respectively, compared to 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) with M for the MS-Leg by 1 % compared to L and significantly decreased with L by 1 % compared to N. For the LS-Leg, the JRF angle was significantly increased with M by 1 % compared to N and 1 % compared to L, and significantly decreased with L by 1 % compared to N.

At peak 2 of the FGRF, the adduction moment, adduction angle, and frontal-plane JRF did not differ significantly between any of the COP conditions for either limb. The angle between the resultant frontal-plane JRF and the horizontal pelvis line was significantly increased with M for the MS-Leg by 1 % compared to L, and significantly decreased with L by 1 % compared to N. For the LS-Leg JRF angle was significantly increased with M by 2 % compared to L, and significantly decreased with L by 1 % compared to N.

DISCUSSION

In accordance with our hypothesis, medio-lateral displacement of the elements of the biomechanical device caused a corresponding shift in the COP trajectory. The results presented in this study validate our previous results in healthy young male subjects. Specifically we show a quantitative relationship between COP manipulation in the frontal-plane foot axis and hip joint kinetics and kinematics in a cohort of female bilateral hip OA patients. In accordance with the hypothesis, a medial displacement of COP caused a corresponding increase in IMD, while step width remained constant. Accordingly, hip adduction angle was decreased (increase in abduction angle), hip external adduction moment was decreased, and hip joint frontal-plane joint reaction force was decreased. With respect to spatiotemporal results, single support duration was increased and double support decreased. The results were observed in both the more symptomatic and less symptomatic limb.

In the present study, as observed in the pilot study on healthy subjects, we found an increase in joint reaction force angle (direction became more vertical) with the medial center of pressure. We speculated that this may have occurred due to less force being transferred along the horizontal axis as a result of the change in hip kinematics as well as the reduced abductor muscle force required to maintain a level pelvis. Manipulation of this angle by means of a footworn device may have clinical significance. For example, patients with medial or supero-medial hip OA (loss of cartilage in the medial or supero-medial aspect of the hip joint, respectively) may benefit from rotating the joint force in the frontal plane to a more vertical position thus reducing loads on the more diseased area of the hip. This however cannot be concluded from the present study and requires further investigation.

It must be noted that, as expected, changes in the gait parameters between the foot center of pressure configurations, although statistically significant, were small. Foot center of pressure was shifted 0.8 cm and 1.5 cm from the neutral configuration in the medial and lateral configurations, respectively, and thus in order to maintain base of support, inter-malleolar distance must be increased or decreased by the same amount. Indeed we see that inter-malleolar distance changed on the order of approximately 1 cm between the medial and neutral center of pressure configurations. This would translate into a very small increase in hip abduction angle as seen, and subsequently small changes in all other parameters including hip joint reaction force. In this study we found a 2 % average decrease in joint reaction force with the medial configuration compared to the other two device configurations. This can be interpreted as a 2 % decrease in percent of body weight that loads both joints. Loads on the hip joint have been measured to reach 2 to over 5 times body weight in gait.[21-24] For the cohort in this study this would translate into average loads of 147-367 kg (or 1442-3600 N) on each joint. A 2 % reduction in this load would reduce peak loads during each step by an average of 3-7 kg (or 29-72 N) on each hip joint. However, clinical significance of this load reduction requires further investigation. In addition, the 2 % reduction in joint force that we observed with a medial device configuration was relative to the neutral device configuration which we defined as the control condition. This configuration does not reflect the patient's native condition while walking without the device. We elected not to have a secondary control such as a regular shoe or a barefoot condition, as these conditions represent entirely different walking conditions than those induced by the device, and parameters tested in these conditions would not contribute to proving/disproving the hypothesis. Specifically, the convexivity of the device elements induces an element of instability thus "forcing" the user to adapt to the center of pressure configuration set by the device elements, and the associated biomechanical changes, in order to maintain stability. Contrary to this, regular shoes or barefoot represent more stable walking conditions and thus have substantially less demands on the neuromuscular system. It would therefore not be surprising to find significantly different gait parameters in these conditions as compared to the device configurations, however this is not relevant to the present study. In addition, subjects underwent testing in the biomechanical device only in order to maintain consistency of the kinematic model. In the device conditions, reflective markers remained in the same place, while the device elements on the shoe sole were shifted only. This allowed us to accurately attribute even small changes in gait parameters directly to the foot center of pressure shift. Our ultimate interest beyond the scope of this study is to assess gait while *not* wearing the biomechanical device before and after undergoing gait training in such a device (i.e., assessing results of motor learning after gait training with the device set to medial center of pressure configuration).

We must also note that, as hypothesized, a medial center of pressure shift caused an increase in inter-malleolar distance while step width remained constant. Contrary to the hypothesis, step width in the lateral configuration was significantly increased compared to the neutral configuration while inter-malleolar distance did not change. This may have occurred for several reasons. One reason for this may be that a decrease in inter-malleolar distance would result in the limbs being too close together and cause rubbing of the thighs during gait. Another reason may be that a decrease in inter-maleolar distance and subsequent increase in adduction angle would cause a painful increase in joint reaction force, and thus patients maintain inter-maleolar distance. These possible reasons, however, are speculative and would require further investigation to confirm. The results show that indeed, there was no significant difference between the adduction/abduction angle, moment, or frontal-plane JRF when comparing the lateral configuration to the neutral control. This result reinforces that the change in adduction angle may play an important role, via reduction in abductor muscle forces, in the reduction of the joint reaction force as speculated in our previous study[12] as well as in the present study. Contrary to the laterally deviated center of pressure, a reduction of joint reaction force was achieved in both limbs when the center of pressure was shifted medially.

A noteworthy result from this study is that gait parameters at the time of peak 1 of the frontal-plane ground reaction force changed significantly with the COP shift, while parameters at peak 2, other than the joint reaction force angle, did not show any significance. This finding is consistent with our previous studies using the biomechanical device.[10,12] This may have several explanations. Firstly, it has been suggested that there is greater variability of gait parameters at peak 2, and this may have contributed to the statistical insignificance.[10,25] Secondly, at peak 1 of the frontal-plane ground reaction force, the foot center of pressure is located approximately under the heel, and therefore is defined primarily by the heel element of the device. In this case the ground reaction force passes approximately through the length of the limb. This may allow maximal effect of the center of pressure configuration. At peak 2 of the frontal-plane ground reaction force, the foot center of pressure is located under the forefoot element of the device, and is thus primarily defined by this element. In this case, the ground reaction force does not pass through the limb but rather anterior to the limb. This may render the center of pressure element less effective in influencing frontal-plane hip parameters. Finally, at peak 1, the limb is in a relatively more passive state during the weight acceptance stage of the gait cycle and may be more influenced by center of pressure changes. Contrary to this, at peak 2 the limb is in a relatively more active state during the push-off stage of the gait cycle and may be less susceptible to biomechanical manipulation due to center of pressure changes.

Several limitations of this study must be acknowledged. Firstly, gait testing was performed shortly after patients were outfitted with the biomechanical device. Thus results exhibited in this study may not reflect results that would be obtained after a long period of usage of the device in each center of pressure configuration. Additionally, the results of this study pertain only to the distinct cohort of older female bilateral hip osteoarthritis patients.

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Table 1. COP offset parameters for M and L configurations (with respect to N) (n=16)

COP Position		**M**	**L**
Average Stance Phase COP Offset [mm]	MS-Leg	6.03(2.45) *(p<0.001)	-7.42(2.88) *(p<0.001)
	LS-Leg	6.05(2.38) *(p<0.001)	-8.00(3.42) *(p<0.001)
COP Offset at Peak 1 of FGRF [mm]	MS-Leg	5.84(2.55) *(p<0.001)	-6.29(2.99) *(p<0.001)
	LS-Leg	5.59(3.19) *(p<0.001)	-5.91(3.86) *(p<0.001)
COP Offset at Peak 2 of FGRF [mm]	MS-Leg	5.62(2.97) *(p<0.001)	-8.09(2.95) *(p<0.001)
	LS-Leg	5.64(2.34) *(p<0.001)	-9.22(3.75) *(p<0.001)

Table 1: Mean (standard deviation) of COP offset parameters for MS-Leg and LS-Leg reported as offset relative to neutral element position; * = p value <0.05 for comparison with the neutral element position. Positive values indicate medial COP offset, while negative values indicate lateral COP offset.

Table 2. Spatio-temporal parameters (n=16)

COP Position		**M**	**N**	**L**
Spatio-temporal Parameters				
Step width [m]		0.14 (0.04)	0.14(0.04) b(p=0.008)	0.15(0.04)
IMD [cm]		19.12(3.83)	17.86(3.92) a(p=0.006)	17.34(3.97) a(p=0.026)
Single support duration [% stance phase]	MS-Leg	37.33(3.57)	37.10(2.07)	36.48(2.97) a(p=0.040)
	LS-Leg	37.69(1.92)	37.84(2.37)	37.32(2.43)
Double support duration [% stance phase]		24.99(4.34)	25.08(4.05)	25.97(3.85) a(p=0.008)
Speed [m/s]		0.94(0.12)	0.94(0.13)	0.93(0.12)
Cadence [steps/min]		100.7(9.3)	100.3(9.5)	100.3(8.8)

Table 2: Mean (standard deviation) of spatio-temporal parameters; ^{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.

Table 3. Hip Kinetics and Kinematics at Peak 1 of FGRF (n=16)

COP Position		**M**	**N**	**L**
Kinetics				
Hip Adduction Moment at Peak 1 [N-mm/kg]	MS-Leg	674.89 (182.00)	712.51 (193.91) a(p=0.017)	689.20 (206.51)
	LS-Leg	718.16 (172.68)	776.83 (178.31) a(p=0.002)	742.57 (188.55)
Magnitude of Resultant Frontal-Plane JRF at Peak 1 [N/kg]	MS-Leg	8.37(0.36)	8.50(0.30) a(p=0.039)	8.46(0.38)
	LS-Leg	8.48(0.44)	8.68(0.45) a(p=0.003)	8.75(0.61) a(p=0.011)
Kinematics				
Hip Adduction Angle at Peak 1 [Degrees]	MS-Leg	2.48(3.00)	3.13(3.00) a(p=0.026)	3.06(2.64) a(p=0.023)
	LS-Leg	2.54(3.69)	3.43(4.32) a(p=0.020)	3.25(3.88) a(p=0.039)
Angle between Resultant Frontal-Plane JRF at Peak 1 and the Horizontal [Degrees]	MS-Leg	86.17(2.77)	85.80(3.05) b(p=0.001)	84.95(2.56) a(p=0.0004)
	LS-Leg	85.92(4.70)	85.37(5.26) a(p=0.034), b(p=0.044)	84.87(4.75) a(p=0.0005)

Table 3: Mean (standard deviation) of kinetic and kinematic parameters associated with peak 1 of FGRF for MS-Leg and LS-Leg; [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.

Table 4. Hip Kinetics and Kinematics at Peak 2 of FGRF (*n*=16)

COP Position		M	N	L
Kinetics				
Hip Adduction Moment at Peak 2 [N-mm/kg]	MS-Leg	675.70 (272.03)	675.27 (258.00)	675.41 (280.50)
	LS-Leg	762.72 (186.67)	779.71 (175.76)	770.05 (200.48)
Magnitude of Resultant Frontal-Plane JRF at Peak 2 [N/kg]	MS-Leg	8.41(0.37)	8.48(0.31)	8.43(0.34)
	LS-Leg	8.39(0.53)	8.45(0.56)	8.37(0.54)
Kinematics				
Hip Adduction Angle at Peak 2 [Degrees]	MS-Leg	1.89(3.64)	1.92(3.42)	1.86(3.41)
	LS-Leg	1.72(3.66)	1.98(3.51)	1.84(3.5)
Angle between Resultant Frontal-Plane JRF at Peak 2 and the Horizontal [Degrees]	MS-Leg	83.62(5.03)	83.50(4.81) b(p=0.0004)	82.68(4.95) a(p=0.0005)
	LS-Leg	81.71(3.79)	81.32(3.64) b(p=0.002)	80.43(3.78) a(p=0.0004)

Table 4: Mean (standard deviation) of kinetic and kinematic parameters associated with peak 2 of FGRF for MS-Leg and LS-Leg; [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.

FIGURE LEGENDS

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. (Figure adapted from [12].)

Figure 2: (a) Biomechanical device with adjustable elements in (b) neutral, (c) lateral, and (d) medial configurations. (Figure adapted from [12].)

Figure 3: Representative graph of (a) adduction/abduction angle, (b) adduction/abduction moment, and (c) frontal-plane JRF for the three walking conditions versus percent stance phase.

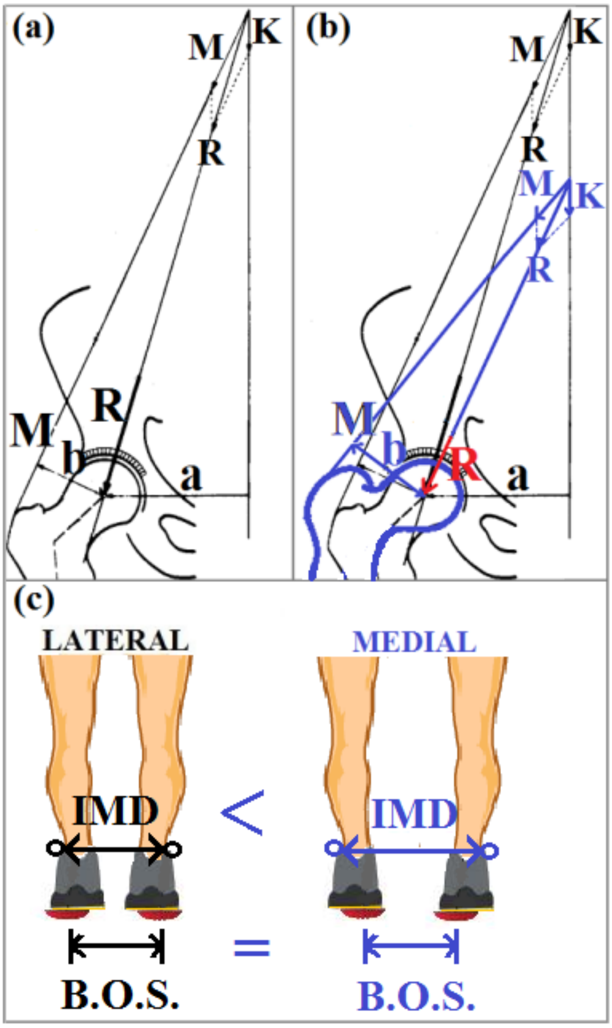


Figure 1

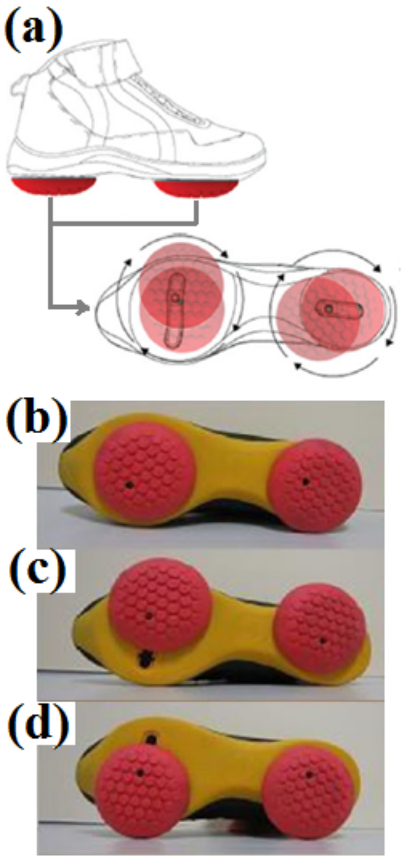
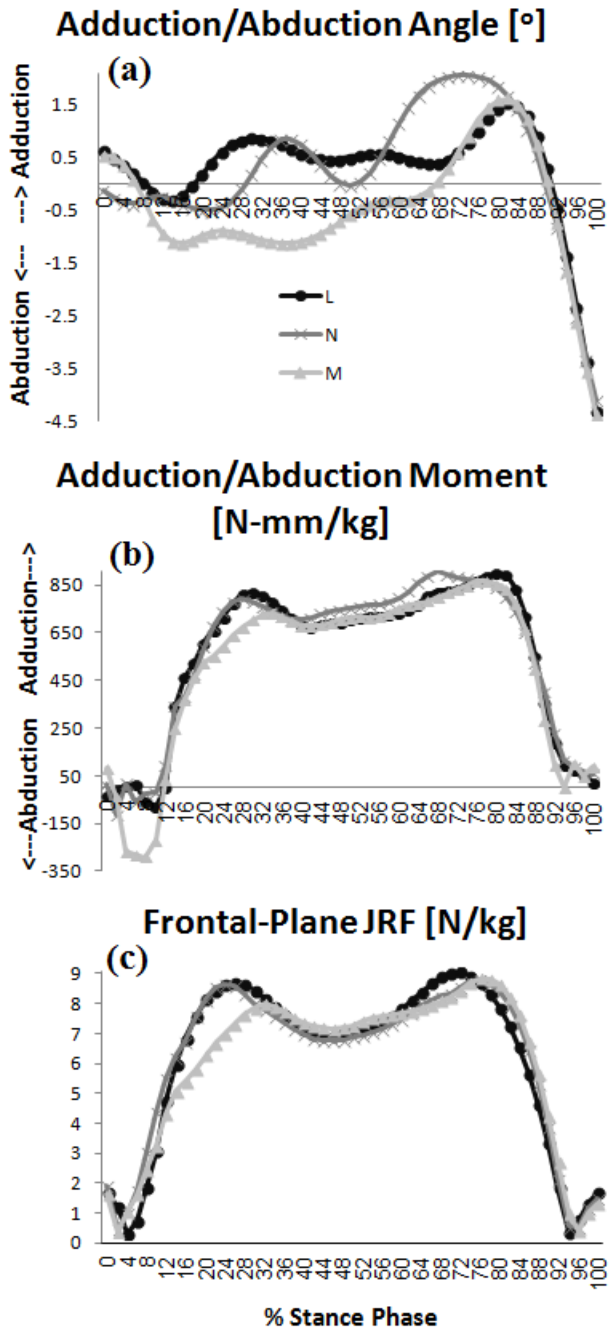


Figure 2



Adduction/Abduction Angle [°]
(a)
Abduction <--- ---> Adduction
L
N
M
Adduction/Abduction Moment [N-mm/kg]
(b)
<---Abduction Adduction--->
Frontal-Plane JRF [N/kg]
(c)
% Stance Phase

Figure 3