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Control of knee coronal plane moment via modulation of center of pressure: A prospective gait analysis study

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ABSTRACT

Objectives: Footwear-generated biomechanical manipulations (e.g., wedge insoles) have been shown to reduce the magnitude of adduction moment about the knee. The theory behind wedged insoles is that a more laterally shifted location of the center of pressure reduces the distance between the ground reaction force and the center of the knee joint, thereby reducing adduction moment during gait. However, the relationship between the center of pressure and the knee adduction moment has not been studied previously. The aim of this study was to examine the association between the location of the center of pressure and the relative magnitude of the knee adduction moment during gait in healthy men.
Methods: A novel foot-worn biomechanical device which allows controlled manipulation of the center of pressure location was utilized. Twelve healthy men underwent successive gait analysis testing in a controlled setting and with the device set to convey three different para-sagittal locations of the center of pressure: neutral, medial offset and lateral offset.
Results: The knee adduction moment during the stance phase significantly correlated with the shift of the center of pressure from the functional neutral sagittal axis in the coronal plane (i.e., from medial to lateral). The moment was reduced with the lateral sagittal axis configuration and augmented with the medial sagittal axis configuration.
Conclusions: The study results confirm the hypothesis of a direct correlation between the coronal location of the center of pressure and the magnitude of the knee adduction moment.

1. Introduction

Approximately 60–80% of the load across the knee is transmitted to the medial compartment (Andriacchi, 1994; Prodromos et al., 1985). The relatively high-medial compartment load is due to the fact that the line of force acting at the foot passes medial to the knee joint center during gait (Johnson et al., 1980), generating an adduction moment which is proportional to the combination of the ground reaction force (GRF) and the perpendicular distance of this force from the center of the joint (Schipplein and Andriacchi, 1991). This moment tends to adduct the tibiofemoral joint, providing a major contribution to the elevated medial compartment load. It has been proposed that the adduction moment plays a key role in the pathogenesis of osteoarthritis (OA) of the knee through greater compression of the medial side of the joint and through induction of lateral joint laxity via chronic stretching (Goh et al., 1993). An abnormally high-knee adduction moment has been reported to be characteristic of the gait patterns in people with knee OA (Andriacchi, 1994). Likewise, knee adduction moment was found to be an important factor regulating bone size and mineral content in healthy and arthritic subjects (Hurwitz et al., 1998; Jackson et al., 2004; Wada et al., 2001).

Footwear-generated biomechanical manipulations (e.g., wedge insoles, foot orthoses) are commonly used in clinical practice to counter the effect of elevated adduction moments. These interventions utilize the principle that parts of the body act as a system of chained links (joint and motors), whereby the whole limb is regarded as one kinetic functioning unit, starting from the foot proximally through the body segments (Zajac et al., 2002). The application of a laterally wedged shoe insole was first introduced in the 1980s (Yasuda and Sasaki, 1987). Gait analysis studies in healthy subjects showed that, under dynamic conditions, wearing laterally wedged insoles reduced the magnitude of adduction

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moments about the knee joint (Crenshaw et al., 2000). A beneficial effect of wearing a laterally wedged insole has been reported in patients suffering from OA of the knee; medial- and lateral-wedged insoles were found to increase and decrease lateral thrust at the knee during walking, respectively (Ogata et al., 1997). Similarly, wearing a laterally wedged insole reduced the knee joint adduction moment during gait (Shimada et al., 2006).

Previous studies have proposed biomechanical theories for wedged insoles. Maly et al. (2002) hypothesized that the mechanism of reduction of adduction moment of the knee with the use of insoles is a lateral shift in the center of pressure (COP) on the foot, reducing the distance between the GRF and the center of the knee joint, thereby reducing adduction moment during gait. Sasaki and Yasuda (1987) reported that lateral-wedged insoles did not alter the femorotibal angle but did increase the valgus position of the subtalar joint. Xu et al. (1999) confirmed that insole conditions caused a change in the location of the COP during gait. Kakihana et al. (2005) reported that wearing a laterally wedged insole significantly increased the valgus moment at the subtalar joint by creating a lateral shift in COP location. However, systematic correlation between the exact COP and the knee adduction moment has yet to be determined. The current study was devised, therefore, to examine the effect of COP location on the knee joint moment in the coronal plane during gait in normal healthy adults. We utilized a novel foot-worn biomechanical device comprised of two individually calibrated biomechanical elements, thus allowing controlled manipulation of the COP location. We hypothesized that translation of elements in the coronal plane (i.e., from medial to lateral) would result in matching COP displacement and would generate a matching response of the knee adduction moment during the stance phase.

2. Methods

2.1. Participants

The study cohort was comprised of 12 healthy male undergraduate students with equivalent shoe size (French 43) and a similar anthropometric profile (i.e., weight, height, dominant leg). The demographic data of the subjects are noted in Table 1. Exclusion criteria were any orthopedic musculoskeletal or neurological pathology. Approval of the Ethics Sub-Committee was obtained and informed consent was given by all participants. The purpose and methods of the study was explained to the subjects.

2.2. The biomechanical system

A novel biomechanical device comprising four modular elements attached onto foot-worn platforms was utilized (APOS System, APOS–Medical and Sports Technologies Ltd. Herzliya, Israel). The device consists of two convex shaped biomechanical elements attached to each of the feet (Fig. 1). One is located under the hind foot region and the other is located under the forefoot region. The elements are attached to the subject's foot using a platform in the form of a shoe. The platform is equipped with a specially designed sole, which consists of two mounting rails enabling flexible positioning of each element under each region. Each element position can be calibrated individually to convey specific biomechanical challenges in multiple planes. The biomechanical systems used in the study (3 pairs) were generously donated by the manufacturer prior to the study. A pilot trial conducted to assess the stability of the apparatus determined that, for healthy adults, satisfactory walking stability can be kept within the range of 1.5 cm medial and 2 cm lateral deviation of the biomechanical elements from the neutral sagittal axis.

Table 1
Demographic data of participants ($n = 12$)

Age (years)	Height (cm)	Weight (km)
25.7 ± 2.13	177 ± 3.8	73.3 ± 4.87

Note: Values are mean $\pm$ SD.

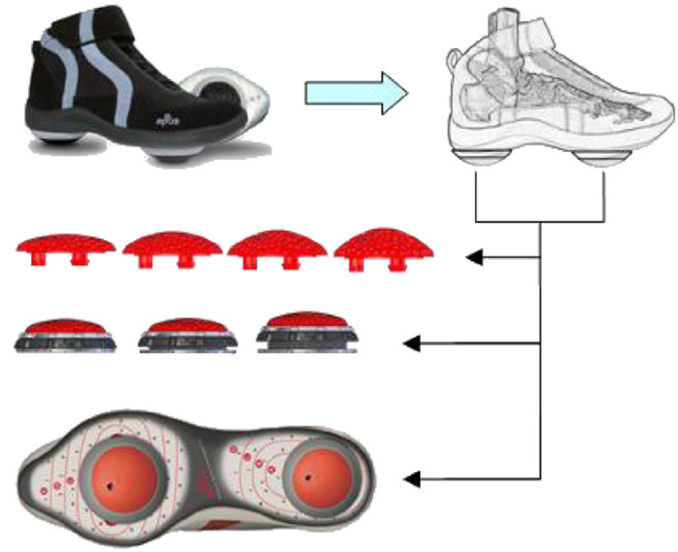

Fig. 1. Biomechanical platform and mobile elements.

2.3. Experimental protocol

Functional assessment of all subjects enrolled in the study was performed prior to testing by the same physician (HA). Calibration of the biomechanical device was performed by the same physiotherapist; first position of the elements for the "functional neutral sagittal axis" was determined and documented. The functional neutral axis was defined as the position in which the apparatus caused the least valgus or varus torque at the ankle to the individual being examined. Medial and lateral axes were then defined as 0.8 cm medial and 1.5 cm lateral deviation of the biomechanical elements from the neutral sagittal axis, respectively (Fig. 2).

Successive gait analysis testing each with singular calibration of the apparatus was performed in four conditions: foot-worn platform with no elements attached (control condition)—Fig. 2a, biomechanical elements placed at neutral axis—Fig. 2b, biomechanical elements placed at lateral sagittal axis—Fig. 2c, and biomechanical elements placed at medial sagittal axis—Fig. 2d. Subjects were asked to walk at a self-selected velocity, which was then indicated by a metronome to ensure consistent cadence throughout the trial. Six trials of each condition were collected per subject for averaging. All conditions were tested in random order on the same day.

2.4. Data acquisition and processing

Gait analysis of each subject was preformed at the Biorobotics and Biomechanics Lab (BRML) of the Faculty of Mechanical Engineering at Technion-Israel Institute of Technology. Three-dimensional motion analysis was performed using an 8-camera Vicon motion analysis system (Oxford Metrics Ltd., Oxford UK) for kinematic data capture. The GRF were recorded by two three-dimensional AMTI OR6-7-1000 force plates. Kinematic and kinetic data were collected simultaneously while the subjects walked over a 10 m walkway. Passive reflective markers were fixed with adhesive tape to anatomical landmarks identified by an experienced physician (HA). A standard marker set was used to define joint centers and axes of rotation (Kadaba et al., 1990). A knee alignment device (KAD; Motion Lab Systems Inc, Baton Rouge LA) was utilized to estimate the three-dimensional alignment of the knee flexion axis during the static trial. Knee joint moments in the coronal plane were calculated using inverse dynamic analyses from the kinematic data and force platform measures using 'PlugInGait' (Oxford Metrics, Oxford UK). Relative offset of the COP and corresponding values of the knee adduction moment (1st peak and 2nd peak) and the knee adduction impulse (the time integral of the knee adduction moment) were calculated for each trial and the average determined across trials for each subject. The latest value (adduction impulse) represents the cumulative magnitude of the knee adduction moment throughout the entire stance phase. Several recent studies utilized this value and reported it to be a useful gait parameter (Thorp et al., 2006a, b; Stefanyshyn et al., 2006).

All analyses were performed for the dominant leg. Joint moments were normalized for body mass and reported in SI units (Newton meters per kg). To examine the relationship between the different interventions on the outcome measures, a program was purposely written in MATLAB12 software. First, a single curve was plotted by averaging the six successive trials (in each configuration) and was normalized to stance phase time. The patterns of the knee joint moments in

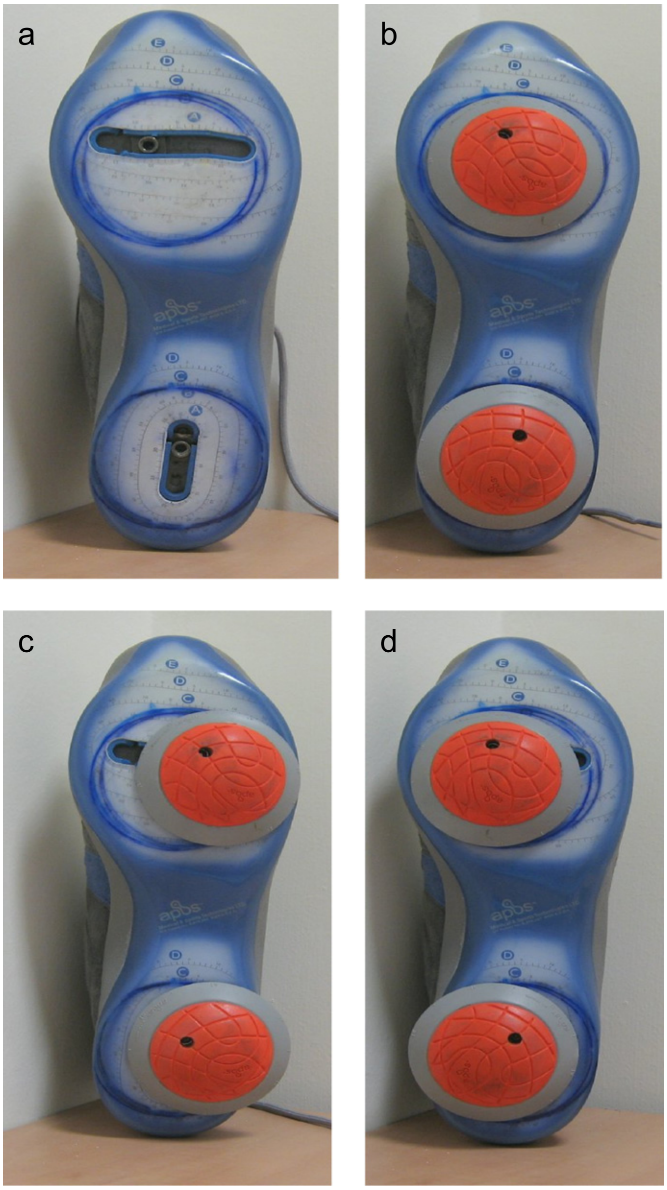

Fig. 2. (a) Biomechanical device with no elements attached; (b) at neutral sagittal axis; (c) at lateral sagittal axis; (d) at medial sagittal axis.

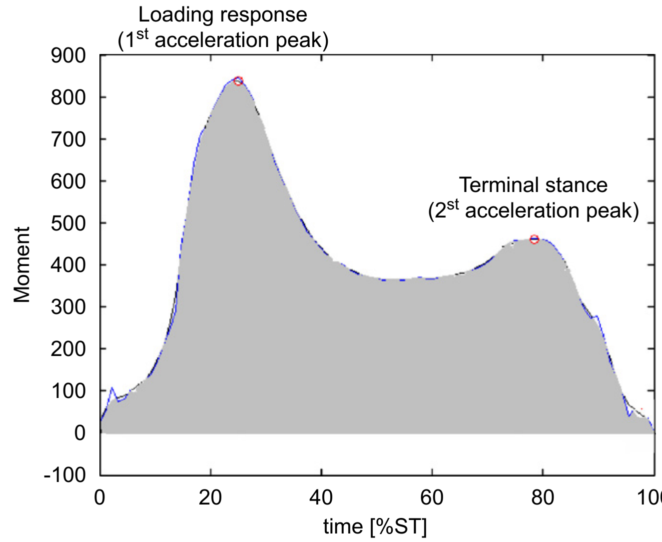

Fig. 3. Representative graph showing subject's adduction moment of the knee during stance (averaged from 6 trials) at loading response (1st peak) and at terminal stance (2nd acceleration peak). Shaded area represents knee adduction impulse.

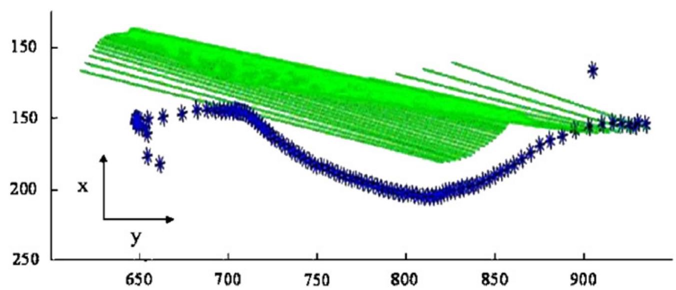

Fig. 4. Calculation of COP position during stance phase: Scatter plot of force plate recorded instantaneous COP coordinates (*) and matching instantaneous coordinates of foot segment sagittal axis (defined by heel and toe markers—green line). The relative COP position is defined as the distance from foot axis to its corresponding COP. (Negative values indicate lateral offset). *Y* and *X* are major axes of the lab's coordinate system.

the coronal plane during the stance phase are similar to those of the vertical GRF, which has two peaks separated by a valley (i.e., the loading response peak, the midstance valley, and the terminal stance peak). Hence, the program was devised to identify the two peaks and calculate their magnitude, as well as the time integral of the curve (Fig. 3). To obtain the COP position with each shoe modification, the instantaneous coordinates of the COP recorded by the force plate (i.e., the junction point of the GRF with the force plate surface) throughout the stance phase were extracted. Matching instantaneous coordinates of the heel and toe markers (defining a fixed sagittal axis reference in the foot segment) were then obtained (Fig. 4). The instantaneous distance in the horizontal plane from the foot axis to the COP was then calculated. Finally, the instantaneous relative COP offset in the medial and lateral configurations were assessed by examining the differences in COP-foot axes with respect to the neutral axis. The COP offset was calculated for loading response and at terminal stance period. Likewise, total COP offset was calculated by averaging the instantaneous values throughout the entire stance phase. Lateral and medial offset of the COP were defined as positive and negative values, respectively, and reported in mm.

2.5. Statistical analysis

Spearman's correlations were used to examine the relationship of spatio-temporal (cadence, step length, step width, gait velocity), kinetic (1st and 2nd acceleration peaks and the knee adduction impulse values) and COP offset parameters in the neutral axis, medial axis and lateral axis configuration of the apparatus. Non-parametric Friedman tests were used for compression of spatio-temporal, kinetic and COP offset parameters in the neutral medial lateral configurations of the apparatus. Wilcoxon tests were used to determine statistical significance differences between each configuration for each of the parameters. A probability of <0.05 was considered as statistically significant. All analyses were performed using SPSS (version 13.0).

3. Results

Mean values+standard deviation of spatial and temporal parameters are listed in Table 2. Cadence, step length and step width were similar for all configurations of the apparatus. The difference in average of walking velocity of 0.04 m/s for the lateral and the natural conditions was statistically significant ($p = 0.045$).

Fig. 5 illustrates the relative COP-foot axis distance during stance phase in a single subject at the four configurations tested (control, neutral sagittal axis, lateral sagittal axis, medial at sagittal axis). With no elements attached, the plot presents a typical COP pattern, curving laterally at midstance and progressing medially at terminal stance. With the convex elements attached in the neutral configuration, the COP displays a linear pattern progressing from lateral position at initial contact to medial position at terminal stance. Fig. 6 demonstrates a representative analysis of COP location in the medial and lateral configuration with respect to the neutral position. The COP shifts in accordance with the displacement of the biomechanical elements. Examining the COP curves, it is evident that COP relocation is minimal at initial contact. The COP shifts medially and latterly during the first 10% of stance phase, and displacement remains marked until final stance where the curves overlap again. Inter-subject analysis for the group mean values and standard deviation of COP displacement for control, medial and lateral configurations (relative to the neutral configuration) are presented in Table 3. COP offset significantly correlated with the medial and lateral translation of the biomechanical elements from the neutral axis. With reference to the control setting, the COP calculated for the neutral axis was located medially in all subjects but one (on average, the COP was located 6.27 mm medial to that of the control setting).

Fig. 7 illustrates adduction moment of the knee in a single subject in the four configurations tested (control, at neutral sagittal axis, at lateral sagittal axis, and medial at sagittal axis). Evidently, the external adduction moment about the knee during the stance phase was reduced with the lateral sagittal axis configuration and augmented with the medial sagittal axis configuration.

Group values for the knee adduction impulse and 1st and 2nd peaks of the moment during stance phase are presented in Figs. 8–10, respectively.

Table 4 presents mean values and standard deviation of adduction moment values. Translation of the biomechanical elements from neutral to lateral position significantly reduced the magnitude of the 1st peak of the knee adduction moment and the knee adduction impulse. The 1st peak was reduced in all subjects, (on average, the peak moment was reduced by 0.163 N-m/kg, a reduction of 20% from the natural configuration), and the adduction impulse was reduced in all subjects but one (on average, the peak moment was reduced by 0.05 N-m/kg*s, a reduction of 17% from the natural configuration). Translation of the biomechanical elements from neutral to medial positions augmented the magnitude of the knee adduction moment; however this was significant for 1st peak only. The 1st peak was increased in 10 subjects, decreased in one and unchanged in one (on average, the peak moment was increased by 0.05 N-m/kg, an increase of 6% from the natural configuration). The magnitude of the 2nd peak of the knee adduction moment was reduced by neutral to lateral translation of the biomechanical elements (this

Table 2
Spatio-temporal parameters, group values ($n = 12$)

Parameters	Control	Neutral axis	Lateral axis	Medial axis
Cadence (steps/min)	100.67 ± 11.78	100.38 ± 9.23	101.33 ± 10.2	100.33 ± 9.09
Step length (m)	0.67 ± 0.05	0. 67 ± 0.06	0.68 ± 0.06	0.67 ± 0.06
Step width (m)	0.16 ± 0.03	0.17 ± 0.03	0.17 ± 0.03	0.17 ± 0.03
Walking Speed (m/s)	1.13 ± 0.16	1.14 ± 0.17	1.17 ± 0.18	1.13 ± 0.15

Note: Values are mean ± SD.

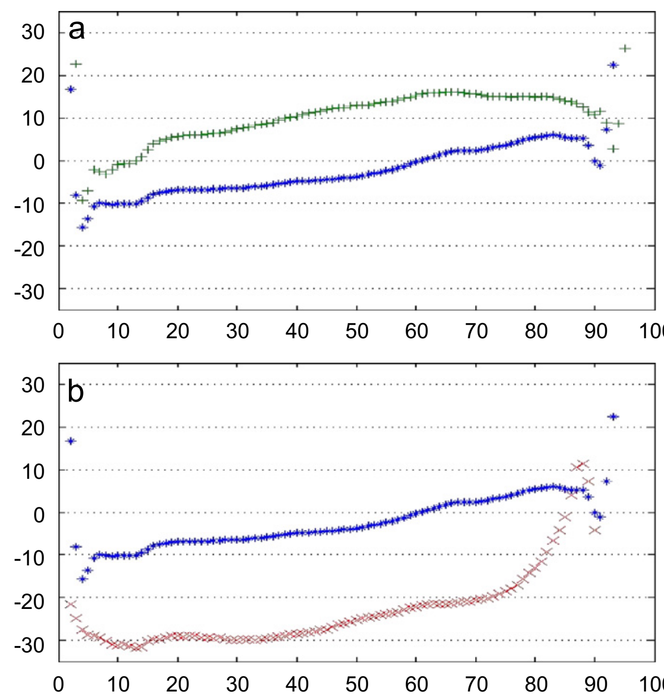

Fig. 6. Representative subject's COP relative offset at the medial (a) and lateral (b) sagittal axis-configurations. The vertical distance between the COP plot in the medial configuration (green) and lateral configuration (pink) from the neutral configuration (blue) represents the absolute COP offset. All values are reported in mm, negative values indicate lateral offset. The *X* axis represents 100% of stance phase time.

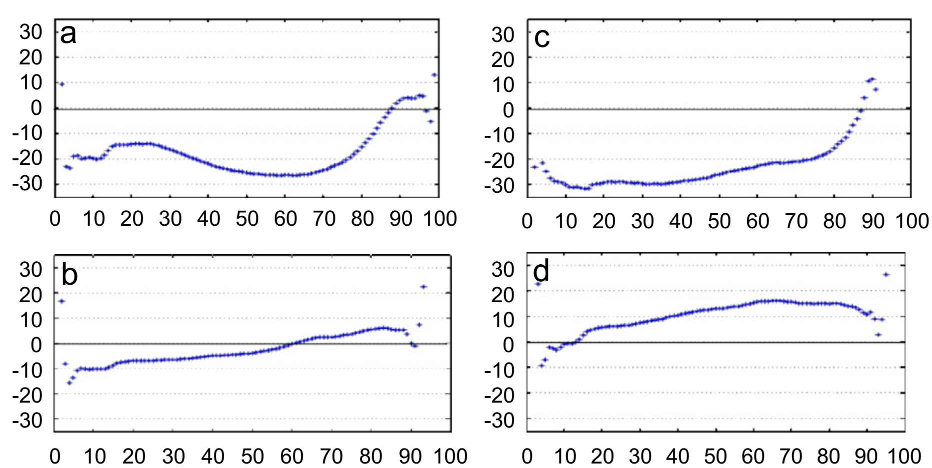

Fig. 5. Representative subject's COP in the Rt foot in the 4 configurations tested (biomechanical device with no elements attached-*a*; at neutral sagittal axis-*b*; at lateral sagittal axis-*c*; medial at sagittal axis-*d*). The *Y*-axis represents instantaneous distance (all values are reported in mm, negative values indicate lateral offset) of the COP to the foot segment sagittal axis, and the *X*-axis represents 100% of stance phase time.

Table 3
COP offset for control medial and lateral configurations (relative to the neutral configuration) ($n = 12$)

Configuration	Attribute		
	Loading response (1st) peak	Terminal stance (2nd) peak	Total COP offset
Medial	10.64 ± 3.27 ($p<0.01$)	10.28 ± 2.9 ($p<0.01$)	9.72 ± 3.38($p<0.01$)
Lateral	-13.8 ± 5.34($p<0.01$)	-15.35 ± 5.16 ($p<0.01$)	-14.37 ± 6($p<0.01$)
Control	-5.2 ± 3.52 ($p=0.01$)	-6.75 ± 5.63 ($p=0.01$)	-6.27 ± 3.09($p<0.01$)

Notes: Values are mean$\pm$SD; v alues are reported in mm; negative values indicate lateral offset.

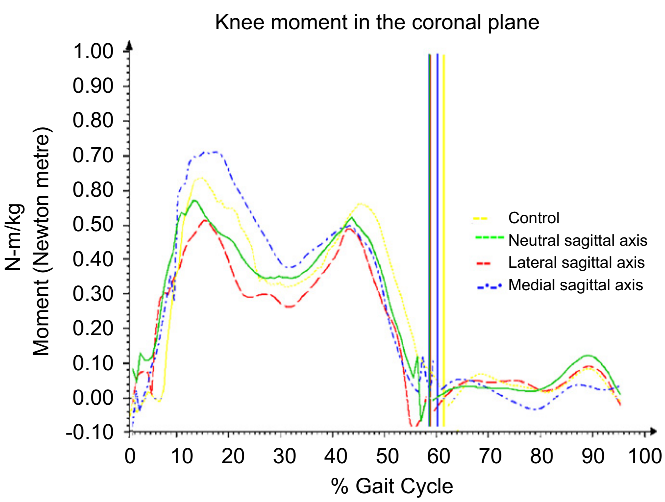

Fig. 7. Representative subject's adduction moment of the knee in the 4 configurations tested (biomechanical device with no elements attached, at neutral sagittal axis, at lateral sagittal axis, and medial at sagittal axis). The *Y*-axis represents moment (Newton-meters/kg (N-m), and the *X*-axis represents 100% of a single gait cycle. The vertical lines represent the end of the stance phase.

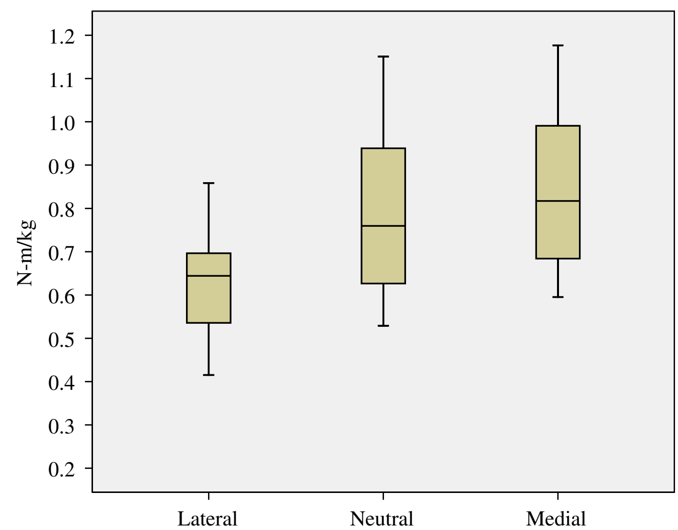

Fig. 9. Relationship between values of knee adduction moment at loading response peak (1st peak) with lateral, neutral and medial translation biomechanical elements. Data presented as box-plots-line in center of box represents the median value; the box represents the inter-quartile range, and the whiskers represent the range.

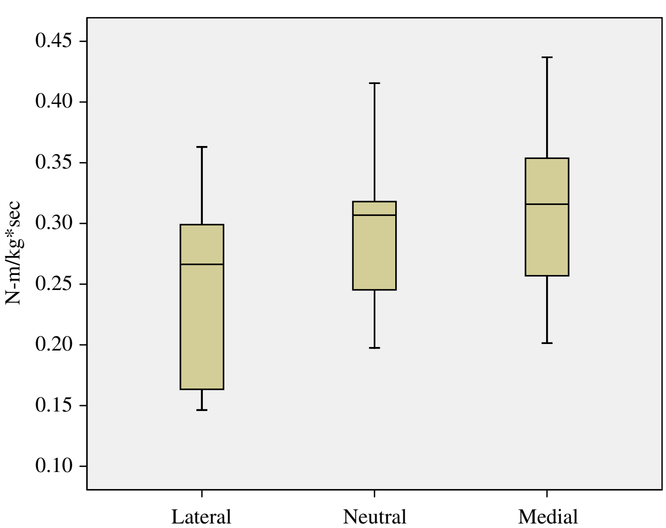

Fig. 8. Relationship between values of the knee adduction impulse during stance phase, with lateral, neutral and medial translation biomechanical elements. Data presented as box-plots-line in center of box represents the median value; the box represents the inter-quartile range, and the whiskers represent the range.

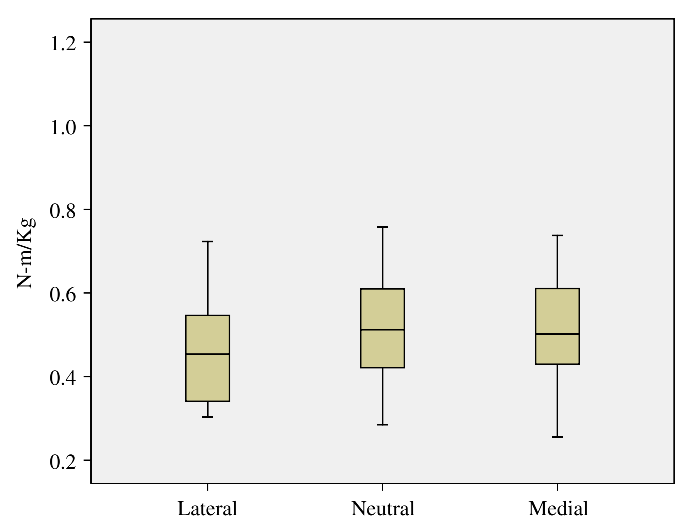

Fig. 10. Relationship between values of knee adduction moment at terminal stance peak (2nd peak) with lateral, neutral and medial translation biomechanical elements. Data presented as box-plots-line in center of box represents the median value; the box represents the inter-quartile range, and the whiskers represent the range.

change was not significant), and was relatively unchanged by neutral to medial translation. The knee adduction moment magnitude did not differ significantly between the neutral axis and the control (no elements) configuration.

Spearman's correlations analysis was performed with P1 values and knee adduction impulse of the moment with the three conditions obtained from the subjects (Table 5). There was a reverse correlation in moment values between the neutral and medial conditions, and a direct correlation between the lateral and neutral conditions.

Table 4
Knee adduction moment magnitude, group values ($n = 12$)

Parameters	Control	Neutral axis	Lateral axis	Medial axis	Medial-neutral difference	Lateral-neutral difference
Knee adduction impulse (N-m/kg*s)	0.27 ± 0.09	0.29 ± 0.06	0.24 ± 0.07	0.31 ± 0.07	0.017 ± 0.046 $p = 0.1$	0.051 ± 0.057 $p = 0.005$
Loading response (1st) peak (N-m/kg)	0.66 ± 0.22	0.8 ± 0.21	0.63 ± 0.13	0.85 ± 0.18	0.055 ± 0.066 $P = 0.019$	0.163 ± 0.09 $p = 0.003$
Terminal stance (2nd) peak (N-m/kg)	0.5 ± 0.22	0.51 ± 0.13	0.47 ± 0.13	0.51 ± 0.14	-0.02 ± 1.27 NS	0.05 ± 0.11 NS

Note: mean values+standard deviation.

Table 5
Spearman's correlations analysis of adduction moment magnitude at loading response and of knee adduction moment impulse

Spearman's ratio	Control-norm	Lateral-norm	Neutral-norm	Medial-norm
Adduction moment impulse				
Control-norm correlation coefficient	1.000	0.566	0.720**	0.524
Sig. (2-tailed)		0.055	0.055	0.080
N	12	12	12	12
Lateral-norm correlation coefficient	0.566	0.566	0.685*	0.874**
Sig. (2-tailed)	0.055	0.055	0.014	0.055
N	12	12	12	12
Neutral-norm correlation coefficient	0.566	0.566	1.000	0.685*
Sig. (2-tailed)	0.055	0.055	0.055	0.014
N	12	12	12	12
Medial-norm correlation coefficient	0.566	0.566	0.685*	1.000
Sig. (2-tailed)	0.055	0.055	0.014	
N	12	12	12	12
Adduction moment magnitude, at loading response				
Control-norm correlation coefficient	1.000	0.818**	0.895**	0.832
Sig. (2-tailed)		0.001	0.001	0.001
N	12	12	12	12
Lateral-norm correlation coefficient	0.818**	1.000	0.825**	0.804**
Sig. (2-tailed)	0.001		0.001	0.002
N	12	12	12	12
Neutral-norm correlation coefficient	0.895**	0.825**	1.000	0.895**
Sig. (2-tailed)	0.001	0.001		0.014
N	12	12	12	12
Medial-norm correlation coefficient	0.832**	0.804	0.895**	1.000
Sig. (2-tailed)	0.001	0.002	0.000	
N	12	12	12	12

* Correlation is significant at the 0.05 level (2-tailed).
** Correlation is significant at the 0.01 level (2-tailed).

4. Discussion

The results presented indicate that accurate management of adduction moment about the knee can be attained by controlled shift of the center of pressure. Our study examined the kinetic outcome of a novel biomechanical apparatus on the knee joint adduction moment in healthy subjects. Lateral-wedged insoles were previously reported to decrease load and adduction moment magnitude in the medial compartment of the knee joint in healthy subjects. It is suggested that this is due to the more laterally shifted location of the COP (Kakihana et al., 2005). To the best of our knowledge, this is the first study to utilize a biomechanical device which allows controlled modulation of the center of pressure.

We found that the magnitude of the knee adduction moment significantly correlated with the coronal orientation of the biomechanical elements; translation of the biomechanical elements from the medial to the neutral position and from the neutral to the lateral position significantly reduced the magnitude of the adductor moment's 1st peak. This finding confirms the study's hypothesis of a direct correlation between the coronal position of the biomechanical elements (location of the center of pressure) and the coronal kinetics of the knee. We speculate that the laterally shifted COP reduced the distance between the GRF and the center of the knee joint, resulting in reduced magnitude of the moment.

The magnitude of the terminal stance phase (2nd) peak of the knee adduction moment did not significantly correlate with the COP translation. This finding is in agreement with a previous study reporting increased variability of the 2nd peak (Hurwitz et al., 2002). Conversely, the adduction moment impulse (representing the total magnitude of the moment) displayed distinct correlation with the total COP offset. This highlights the value of utilizing this parameter for evaluating knee coronal moments throughout the stance phase.

Average walking velocity increased by 0.04 m/s at the lateral configuration. A positive correlation between walking velocity and the magnitude of the knee adduction moment was previously reported by Winter (1984). Reduction in moment values despite accelerated velocity in the lateral condition emphasizes the effect of COP orientation on the adduction moment.

Several studies have reported the biomechanical effects of lateral-wedged insoles in healthy individuals. Kakihana et al. (2004) reported a 10.4% reduction for healthy elders with a 6° lateral-wedged insole. Similarly, Crenshaw et al. (2000) reported

0.029 N-m/kg (6%) peak adduction moment reduction with a 5° lateral-wedged insole. In the present study, the average peak moment was reduced by 0.163 N-m/kg, a reduction of 20% from the neutral configuration. This suggests a superior mechanism with direct COP translation.

Several limitations arising from the current study should be noted. Firstly, the relative COP location was analyzed indirectly by calculating instantaneous force plate recorded COP and corresponding foot segment sagittal axis distance. While, this method offers reasonable evaluation of the COP offset, future studies incorporating direct COP measurement (e.g., pedobarograph analysis) could provide valuable data regarding shoe COP pattern modulation. Another limitation of this study was the employment of the apparatus with no elements attached as a control. This setting was elected to assure consistency of the kinematic model (biomechanical elements were attached and modulated without repositioning of the retro reflective markers). Clearly, 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 in respect to the control. Finally, 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. Moreover, it should be noted that individuals who exhibit reduced stability during walking (e.g., older individuals) may not be able to manage biomechanical challenges such as those generated by the apparatus utilized in the study. Further studies are therefore needed before these findings can be extended to other populations.

In conclusion, the present study indicates that COP modification enables controlled, customized manipulation of knee coronal kinetics in healthy subjects. The results confirm the hypothesis of a direct correlation between the coronal position of the COP and the coronal kinetics of the knee. These findings offer new understanding of lower limb biomechanics, and have implications in the field of biomechanical apparatus design and practice.

Conflict of interest statement

No author has any conflict of interest to declare.

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