Mechanism of reducing knee adduction moment by shortening of the knee lever arm via medio-lateral manipulation of foot center of pressure: A pilot study

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ABSTRACT

Prominent conservative treatment options for medial-compartment knee osteoarthritis include footwear that reduces knee adduction moment (KAM) correlated with detrimental loads in the medial compartment of the knee, thus providing clinical benefit. The proposed mechanism by which they reduce KAM is a lateral shift in foot center of pressure (COP) and a consequent shortening of the knee lever arm (KLA), thereby reducing KAM, which can be simply calculated as KLA multiplied by the frontal plane ground reaction force (FP-GRF). The present study investigated this mechanism for a unique biomechanical device capable of shifting COP by means of moveable convex elements attached to the shoe. Fourteen healthy young male subjects underwent gait analysis in two COP configurations of the device for comparison: (1) laterally and (2) medially deviated. Average midstance KLA and KAM were decreased by 8.2% and 8.7%, respectively, in the lateral COP compared to medial. Ground reaction force parameters, frontal plane knee angle (FP-KA), and spine lateral flexion angle (SLF) did not differ between COP configurations. No study parameters differed for terminal stance. Linear mixed effects models showed that COP and FP-GRF components, but not FP-KA and SLF, were significant predictors of KLA. In addition, KLA and FP-GRF were significant predictors of KAM; although, FP-GRF did not change significantly with medio-lateral COP shift, while KLA did. This suggests that the mechanism by which the study device reduces KAM is primarily through shortening of KLA brought on by a lateral shift in COP.

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1. Introduction

Knee osteoarthritis (OA) is a debilitating progressive degenerative disease characterized by loss of articular cartilage, or joint space narrowing, bony enlargement, osteophytosis, crepitus, joint stiffness, joint laxity, and pain, among others (Altman et al., 1986; Felson et al., 2000). It is one of the leading causes of global disability with estimated global prevalence of 3.8% of radiographically confirmed knee OA in 2010 (Cross et al., 2014). Mechanical stress is suggested to be the primary factor contributing to knee OA (Visser et al., 2015). Therefore, current nonsurgical nonpharmacological treatment recommendations include load reduction on the diseased joint (Zhang et al., 2008), which may slow or stop disease progression (Lafeber et al., 2006; Marijnissen et al., 2002; Pollo and Jackson, 2006; van Valburg et al., 1995; 1999), provide

relief of symptoms, and increase quality of life (Erhart-Hledik et al., 2012; Erhart et al., 2010b; Haim et al., 2012).

The knee adduction moment (KAM) has been shown to be significantly correlated with medial-compartment knee loads, and is commonly enhanced in knee OA patients (Erhart et al., 2010a; Zhao et al., 2007). Likewise, it has been shown that frontal plane knee lever arm (KLA), or the perpendicular distance from the knee joint center (KJC) to the frontal plane ground reaction force (FP-GRF), is enhanced in knees with OA (Hunt et al., 2006; Weidenhielm et al., 1995, 1994). Manipulation of KAM by means of foot orthoses is a widely accepted conservative treatment option for medial-compartment knee OA (Erhart et al., 2008; Haim et al., 2012; Hinman et al., 2009; McAlindon et al., 2014). Specifically, foot orthoses that shift the foot center of pressure (COP) work on the principle that a lateral shift in COP, along the medio-lateral foot axis, shifts the point at which the FP-GRF vector acts to a location closer to the KJC. This consequently shortens KLA, and hence reduces KAM, since KAM is the product of KLA and FP-GRF magnitude.

In theory, several parameters are responsible for KLA length. These include the COP, which determines the initial point of the FP-GRF vector, the components of the FP-GRF vector, which produce its angle, and the frontal plane knee angle (FP-KA), which if altered may act to displace KJC relative to the FP-GRF vector. Other changes in lower limb and trunk kinematics and kinetics may also contribute to KLA and KAM (Kean et al., 2013), as well as more complex compensatory mechanisms (Jenkyn et al., 2011).

The effect of COP shift on KLA, and consequently on KAM, until the last decade, has only been suggested in the literature. Hunt et al. (2006) was the first to describe KLA magnitude throughout the entire stance phase, and found that it remained relatively constant, and had a higher association with KAM than FP-GRF magnitude in knee OA patients, subsequently corroborated by other studies (Hinman et al., 2012; Jenkyn et al., 2011; Kean et al., 2013). Consequently, they suggested that studies reporting KAM as an outcome measure would benefit from reporting KLA in addition. Several studies using different foot orthoses have described the relationship between COP and KLA, and between KLA and KAM (Hinman et al., 2012; Jenkyn et al., 2011; Kean et al., 2013). These studies found a correlation between medio-lateral shift in COP and shortening of KLA, and between the latter and reduction in KAM, using modified shoes combining a variable stiffness sole and lateral wedging (Kean et al., 2013), variable stiffness shoes alone (Jenkyn et al., 2011), and lateral shoe wedges alone (Hinman et al., 2012). As opposed to shortening of KLA being associated with a lateral COP (Hinman et al., 2012; Kean et al., 2013), Jenkyn et al. (2011), surprisingly, found a medial shift in COP with variable stiffness shoes, which they interpreted as a product of a specific compensatory mechanism adopted in response to the study device.

Mechanism of reduction of KAM should be exhibited for different types of foot orthoses individually, as they are known to induce different gait patterns (Kean et al., 2013), as exemplified by the counter-intuitive medial COP shift seen in the study of Jenkyn et al. (2011). The objective of the present study was thus to exhibit similar outcome measures to previous studies using a foot-worn biomechanical device whose modus operandus is custom shift of COP to a desired location in the transverse plane of the foot. The therapeutic mechanism of the device is in contrast to those with a fixed COP (i.e., one that cannot be adjusted). Accordingly, the shoe sole was designed uniquely for this purpose and therefore differs from its counterparts in that it includes two convex elements attached to each shoe sole, one in the forefoot region and one in the hindfoot region. Significant KAM reduction with medio-lateral COP manipulation, using the study device, has been demonstrated in both healthy subjects (Haim et al., 2008) and knee OA patients (Haim et al., 2011). Additionally, a longitudinal study established clinical success in treatment of knee OA, associating the reductions in KAM with clinical improvement (Haim et al., 2012). However, an explicit investigation of the KAM reduction mechanism for the study device is lacking. Additionally, the aforementioned previous studies that assessed KLA did so in pathological (knee OA) or overweight populations. While these are among the target populations for such devices, the mechanism of KAM reduction may be unique to characteristics of the pathology or condition, as demonstrated by the different predictors of KLA and KAM, and hence different mechanism for KAM reduction, in overweight versus knee OA patients using the same device (Kean et al., 2013). Thus, it is of interest to assess the relationship in a healthy population with the present study device, as a foundation for future study.

The specific objective of the present study was to assess the direct role of a laterally deviated COP in shortening of KLA, and subsequently, the role of KLA in reducing KAM, using a device whose therapeutic mechanism is shift of COP. We hypothesized that a lateral deviation of COP, as compared to a medial deviation, would cause KLA to shorten, and thus be associated with a reduction in KAM.

2. Methods

2.1. Participants

The study group consisted of 14 healthy young male adults with similar characteristics and anthropometric measures (Table 1). Subject exclusion criteria were previous surgery, musculoskeletal injuries, or pathologies of the lower limbs or back, inability to cooperate with or understand instructions, problems with balance, or any neurological, orthopedic, or cardiovascular pathology that would cause pain, impact function, or prevent physical exertion ability. The Ethics Sub-Committee gave approval for the study and all subjects gave informed consent. The purpose and methods were explained to the participants.

2.2. 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 (Haim et al., 2008). 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. 1(a)). A medial or lateral shift of the elements along the medio-lateral foot axis causes a corresponding directional shift of COP (Haim et al., 2008; Khoury et al., 2013).

2.3. Experimental protocol

A neutral configuration (Fig. 1(b)) was custom-positioned by a single trained physiotherapist, and defined as the position of the device elements in which there was the least varus, valgus, dorsi-flexion, or plantar flexion torque about the ankle, as determined by observational gait analysis. The lateral and medial COP configurations (Fig. 1(c) & (d)) were defined as the most extreme deviations of the biomechanical elements from the neutral sagittal axis, along the medio-lateral foot axis, within the constraints of the device, such that there were no observable alterations in frontal plane ankle kinematics. The chosen deviations are therefore clinically implementable, as they do not induce abnormal biomechanics or awkward gait patterns.

Subjects were given a several-minute period prior to data acquisition to walk at a comfortable self-selected speed to become accustomed to the shoes. Subsequently, gait analyses were performed in the two COP configurations, medial and lateral, in random order.

2.4. 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 100 Hz sampling frequency. Ground reaction forces 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 1000 Hz sampling frequency. Six to 8 successful walking trials were performed in each COP configuration, and study parameters were averaged for each set of trials. A standard marker set was used to define joint centers and axes of rotation (Kadaba et al., 1990). Markers were placed by a single trained investigator according to the Vicon 'PlugInGait' marker placement protocol. Analyses were performed for the dominant leg.

Table 1
Subject Characteristics and Anthropometric Measures of the Dominant Limb (N = 14).

Sex	Male
Age [years]	26.5 ± 2.6
Body mass [kg]	71.1 ± 9.6
Height [cm]	174.6 ± 3.6
Leg length [mm]	910.0 ± 30.1
Knee width [mm]	102.4 ± 7.3
Ankle width [mm]	70.8 ± 3.2
Shoe size (French)	43

Values are mean ± standard deviation where relevant. Leg length was measured as the length between the anterior superior iliac spine (ASIS) and the medial malleolus. Knee width was measured as the medio-lateral width of the knee across the line of the knee axis. Ankle width was measured as the medio-lateral distance across the malleoli.
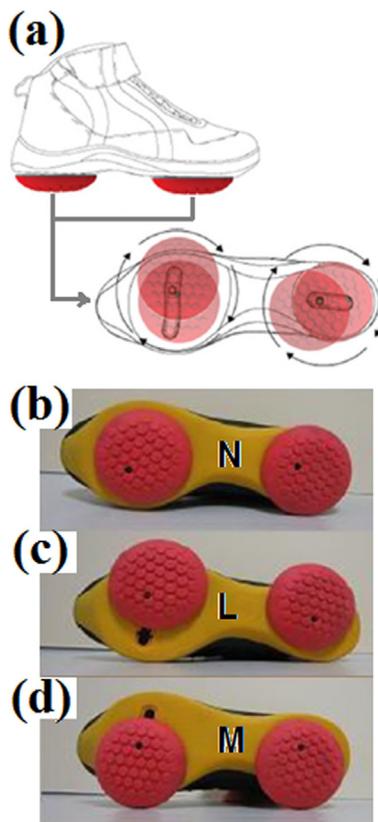

Fig. 1. (a) Biomechanical device with adjustable elements in (b) neutral, (c) lateral, and (d) medial foot center of pressure configurations. N = Neutral, L = Lateral, M = Medial.

Parameters calculated for each of the two device configurations were COP shift, KLA, FP-KA, spine lateral flexion angle (SLF), KAM, and FP-GRF magnitude. The KAM was normalized by body weight and height, and reported in units of Nm/BW*Ht% (Hinman et al., 2012; Jenkyn et al., 2011; Kean et al., 2013). The FP-GRF vector was also resolved into its two constituent components, the vertical GRF (V-GRF) and medio-lateral GRF (ML-GRF). The knee frontal plane was defined as the plane containing the KJC and lateral knee marker, with vertical axis perpendicular to the floor (same as the laboratory vertical axis). The medio-lateral axis was taken as the axis perpendicular to the vertical axis and lying within the knee frontal plane. This plane was chosen in order to control for knee rotation in parameter calculations, in that it rotates internally or externally about the vertical axis as the knee rotates. Table 2 describes the study parameters in detail and Fig. 2 depicts the lower-body parameters.

Table 2
Study parameter abbreviations and detailed descriptions.

| Study parameter	Description
COP	Foot center of pressure: COP shift along the medio-lateral foot axis calculated as the perpendicular distance from the force plate center of pressure, or the initial point of the GRF vector, to the line connecting the toe and heel markers [Haim et al., 2008; Solomonow-Avnon et al., 2016]; positive value indicates lateral direction
KLA	Knee lever arm: knee moment arm calculated as the perpendicular distance from KJC to FP-GRF
FP-KA	Frontal-plane knee angle: knee angle in the knee frontal plane defined as the angle formed by the hip joint center projected onto the knee frontal plane, KJC, and the ankle joint center projected onto the knee frontal plane; value <180° indicates varus alignment
KAM	Knee adduction moment: knee frontal-plane moment defined as the moment about KJC caused by FP-GRF and KLA (KAM = FP-GRF × KLA)
FP-GRF	Frontal plane ground reaction force: projection of the GRF vector onto the knee frontal plane
V-GRF	Vertical ground reaction force: vertical component of FP-GRF along the axis perpendicular to the floor
ML-GRF	Medio-lateral ground reaction force: medio-lateral component of FP-GRF along the axis perpendicular to the vertical axis, and lying in the plane containing the lateral knee marker and KJC; positive value indicates medial direction
SLF	Spine lateral flexion angle: angle between the long axis of the thorax and the long axis of the pelvis, around a floating transverse axis; positive value indicates flexion towards ipsilateral limb
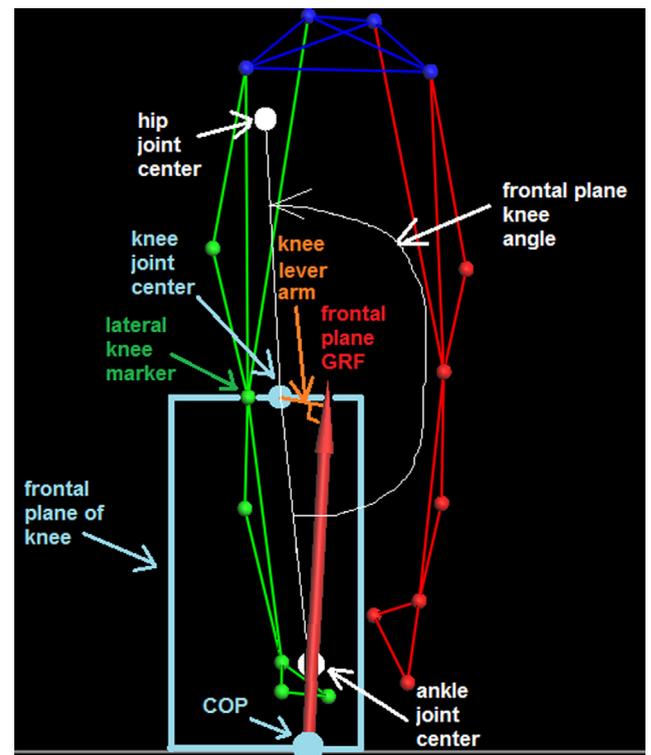
Fig. 2. Lower-body study parameters exhibited on a subject's midstance kinematic model (frontal view). GRF = Ground Reaction Force, COP = Center of Pressure.

Temporal parameters of speed and cadence were recorded. Hip, knee, and ankle joint centers were given automatically by the Vicon 'PlugInGait' algorithm. For reporting study results, the average of each parameter during midstance and terminal stance substages of the stance phase (Midstance: 16.7–50% stance phase,

Terminal Stance: 50–83.3% stance phase) (Perry, 1992) were calculated. Average parameters during loading response and pre-swing were not analyzed, as the curvature of the posterior aspect of the hindfoot element and anterior aspect of the forefoot element induces artifacts in the parameters during these substages of stance that are inconsistent across subjects and do not contribute to the study objective.

2.5. Statistical analysis

Data was analyzed by an experienced biostatistician (AH), using SPSS 23 ©. Data was checked for normal distribution by the Kolmogorov-Smirnov goodness-of-fit test. No variable was rejected for the normal distribution hypothesis. Parameters were compared between medial and lateral COP shift for both midstance and terminal stance by the paired t-test.

Change in each parameter was calculated as the parameter in the lateral configuration minus that in the medial. Two linear regression mixed effects (LME) models were fitted, using change in each parameter, to exhibit the mechanism by which COP shift affects KAM via change in length of KLA. Firstly, a LME model was fitted using KLA as the response variable, subject number (1–14) as a random effect, and COP deviation, ML-GRF, V-GRF, FP-KA, and SLF as fixed effects. Another LME model was fitted using KAM as the response variable, subject number (1–14) as a random effect, and KLA and FP-GRF as fixed effects. F-statistics were calculated and tested accordingly. Belsley collinearity diagnostics were used to test for multi-collinearity among fixed effects, and none was detected (condition indices < 5).

All reported p-values are two-sided. A p-value of <0.05 was considered statistically significant.

3. Results

3.1. Midstance and terminal stance average parameters

Mean and standard deviation for average midstance and terminal stance parameters for the 2 COP configurations, as well as mean difference in parameters between medial and lateral COP configurations, are exhibited in Table 3, along with paired t-test results. Speed and cadence for the total gait cycles are also pre-

sented in Table 3, and did not differ significantly between COP configurations.

During midstance, average COP, KLA, and KAM differed significantly between COP configurations. In the lateral configuration, COP was deviated 9.21 mm laterally ($p = 0.000$), on average, compared to the medial configuration. The KLA was 4.02 mm shorter (8.2% shorter; $p = 0.005$) and KAM was reduced by 8.7% ($p = 0.009$), on average, in the lateral compared to medial configuration. The FP-KA and GRF parameters did not differ significantly between COP configurations. Fig. 3 depicts a representative subject's graphs, averaged for all trials of each COP configuration, for COP, KLA, and KAM (Fig. 3(a), (b), & (c), respectively). Reduction in KLA and KAM is evident throughout midstance, while COP shift is seen for the entire stance phase.

During terminal stance, only COP differed significantly between configurations. In the lateral configuration compared to medial, COP was deviated 8.77 mm laterally ($p = 0.000$), on average.

3.2. Linear mixed effects regression models

Since no study parameters differed for terminal stance, LME models are presented for midstance only.

3.2.1. Knee lever arm model
The model for mean difference in midstance KLA is defined by Eq. (1) (Table 4).

$$KLA \sim 1 + COP + ML\text{-}GRF + V\text{-}GRF + FP\text{-}KA + SLF + (1|Subject) \quad (1)$$

The COP, ML-GRF, and V-GRF had a significant effect on KLA, while FP-KA and SLF did not. A laterally shifted COP, decrease in ML-GRF, and decrease in V-GRF predict shortening of KLA.

3.2.2. Knee adduction moment model
The model for mean difference in midstance KAM is defined by Eq. (2) (Table 4).

$$KAM \sim 1 + KLA + FP\text{-}GRF + (1|Subject) \quad (2)$$

Both KLA and FP-GRF had a significant effect on KAM. Shortening of KLA and decrease in FP-GRF magnitude predict decrease in KAM.

Table 3
Comparison of gait parameters in the lateral and medial foot center of pressure configurations.

Variable	Gait stage	Lateral COP	Medial COP	Difference	p- value
Speed [m/s]	Total	1.35 (0.12)	1.36 (0.12)	−0.01 (0.08)	0.825
Cadence [steps/min]	Total	105.80 (6.78)	105.91 (7.06)	−0.01 (3.51)	0.988
COP [mm]	MS	13.77 (4.06)	4.55 (5.87)	9.21 (2.64)	0.000*
KLA [mm]		49.04 (9.41)	53.07 (10.27)	−4.02 (4.43)	0.005*
FP-KA [°]		170.60 (2.41)	170.75 (2.52)	−0.15 (0.92)	0.550
KAM [Nm/BW*Ht%]		2.64 (0.42)	2.87 (0.51)	−0.23 (0.31)	0.009*
FP-GRF [N]		629.60 (95.55)	633.51 (90.16)	−3.21 (15.33)	0.357
ML-GRF [N]		48.43 (13.63)	50.15 (13.82)	−1.72 (6.59)	0.347
V-GRF [N]		627.42 (95.08)	631.16 (89.84)	−3.74 (15.25)	0.375
SLF [°]		3.12 (2.30)	3.46 (2.25)	−0.34 (0.59)	0.050
COP [mm]	TS	15.70 (2.44)	6.93 (4.63)	8.77 (3.46)	0.000*
KLA [mm]		20.84 (16.96)	21.42 (18.07)	−0.58 (2.90)	0.464
FP-KA [°]		175.21 (2.70)	175.41 (2.51)	−0.20 (0.71)	0.310
KAM [Nm/BW*Ht%]		1.14 (0.92)	1.15 (0.98)	−0.01 (0.15)	0.952
FP-GRF [N]		671.08 (82.56)	666.77 (87.20)	4.31 (15.43)	0.315
ML-GRF [N]		8.99 (8.32)	11.69 (10.73)	−2.70 (6.70)	0.156
V-GRF [N]		670.85 (82.55)	666.48 (87.17)	4.37 (15.37)	0.310
SLF [°]		−2.12 (2.98)	−2.22 (2.68)	0.10 (0.85)	0.668

Values are mean (standard deviation). MS = Midstance; TS = Terminal Stance; COP = foot Center of Pressure; KLA = Knee Lever Arm; FP-KA = Frontal Plane Knee Angle; KAM = Knee Adduction Moment; FP-GRF = Frontal Plane Ground Reaction Force; ML-GRF = Medio-Lateral Ground Reaction Force; V-GRF = Vertical Ground Reaction Force; SLF = Spine Lateral Flexion angle.
* =statistically significant result. Difference is calculated as parameter in lateral COP configuration minus that in medial.
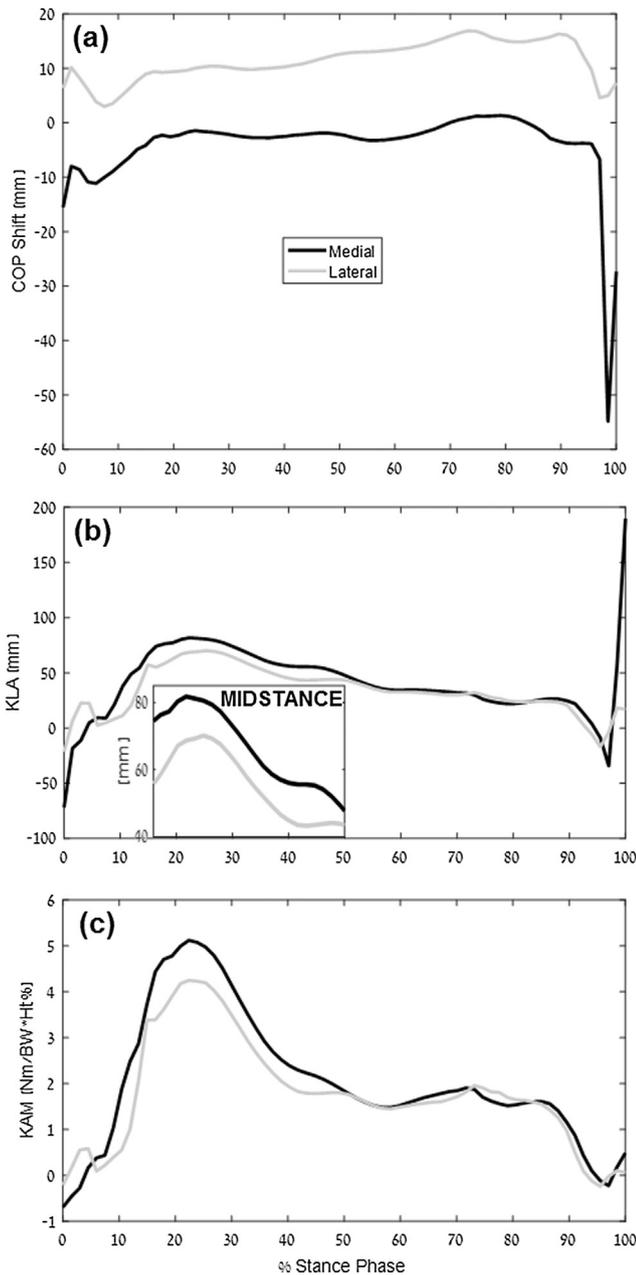
Fig. 3. Representative subject's graphs for (a) foot center of pressure (COP) shift (positive and negative directions represent lateral and medial shift, respectively), (b) knee lever arm (KLA) with inset that zooms in on midstance for visual clarity, and (c) knee adduction moment (KAM) versus percent stance phase for a lateral and medial shift of the biomechanical device elements. The graphs are the ensemble average for all gait trials of the particular subject in the particular device configurations. Positive or negative spikes at the end of the stance phase in several parameters are artifacts resulting from the structure of the shoe device that do not affect reported study results.

Table 4
Linear mixed effects models results.

	Coefficient (standard error)	*p*-value
Response variable: KLA (R² = 0.929 adjusted)		
Intercept	2.86 (1.98)	0.187
COP	−0.64 (0.20)	0.011[*]
ML-GRF	0.33 (0.08)	0.003[*]
V-GRF	0.24 (0.06)	0.003[*]
FP-KA	0.13 (0.81)	0.873
SLF	−1.51 (1.05)	0.188
Response variable: KAM (R² = 0.994 adjusted)		
Intercept	0.01 (0.02)	0.696
KLA	0.05 (0.00)	0.000[*]
FP-GRF	0.01 (0.00)	0.000[*]

COP = foot Center of Pressure; KLA = Knee Lever Arm; FP-KA = Frontal Plane Knee Angle; KAM = Knee Adduction Moment; FP-GRF = Frontal Plane Ground Reaction Force; ML-GRF = Medio-Lateral Ground Reaction Force; V-GRF = Vertical Ground Reaction Force; SLF = Spine Lateral Flexion angle.
[*] =statistically significant result.

2013). Additionally, in agreement with previous results on knee OA patients, while COP was significantly shifted, and KLA and KAM significantly reduced, there were no changes in SLF, FP-KA, or any GRF parameter (Jenkyn et al., 2011; Kean et al., 2013). Further, as FP-GRF did not change significantly with medio-lateral COP shift, it may be concluded that shortening of KLA is the primary mechanism of KAM reduction with the device, despite FP-GRF being a significant predictor of KAM.

Interestingly, although ML-GRF showed a positive correlation with KLA as expected, V-GRF also showed a positive correlation. Theoretically, increase in ML-GRF would cause FP-GRF to become less vertical, distancing it from KJC and increasing KLA. On the other hand, increase in V-GRF would cause FP-GRF to become more vertical, decreasing KLA. In our case, in light of the fact that multicollinearity among predictors was not found, we believe that V-GRF is likely correlated with another biomechanical change, possibly not accounted for in our model, that had a confounding effect to decrease KLA when V-GRF is decreased. Regardless, it should not be viewed as an error, but rather a suggestion that decrease in V-GRF may be associated with a biomechanical response that decreases KLA. Further in-depth analysis is needed to explain this phenomenon.

We elected to report only average midstance and terminal stance values for several reasons. Initially, we analyzed values for loading response and pre-swing, as we wished to exhibit the COP-KLA-KAM relationship throughout the entire stance phase. However, upon inspection of the parameters, unpredictable artifacts were discovered and it was decided that parameters during these periods would not contribute to the study objective. During early loading response and late pre-swing, only the posterior aspect of the hind element or anterior aspect of the forefoot element contacts the ground, respectively, possibly causing instability of the foot due to curvature of these aspects of the device. This may cause varying biomechanical responses across subjects in loading response and pre-swing. Further, large spikes (positive or negative) were observed at the end of stance phase in parameters of a portion of subjects (Fig. 3(a) & (b)) that are inconsistent across subjects. The spikes are artifacts of the shoe structure, stemming from a sudden jump in the initial point of the GRF vector (i.e., COP). We speculate that it may be a result of the point of contact of the shoe device with the floor moving from the convex element to the anterior tip of the shoe sole. Regardless, this does not affect the reported results, as we reported findings for midstance and terminal stance only.

We also note that we elected to use average midstance and terminal stance parameters rather than peak KAM values since not all

4. Discussion

In accordance with the hypothesis, a lateral deviation in COP was associated with shortening of KLA, and reduction in KAM. This specific phenomenon was observed only in midstance. Shift of COP, along with FP-GRF components, was a significant predictor of KLA. The KLA and FP-GRF were predictors of KAM. Results of the present study are in agreement with the previous studies in knee OA patients in that COP and ML-GRF were the significant predictors of KLA (Jenkyn et al., 2011; Kean et al., 2013), and KLA and GRF predictors of KAM (Hunt et al., 2006; Jenkyn et al., 2011; Kean et al.,

subjects had identifiable second peaks, and Hunt et al. (2006) found that peaks of KAM, KLA, and FP-GRF did not coincide, but rather occurred at significantly different times within the stance phase (Hunt et al., 2006). Thus we elected to use the less sensitive average measures. From the inspection of the data, the first peak of KAM (the entire "hump") was well-contained within midstance, while the terminal stance generally included all of the points leading up to and containing the peak (the first half of the "hump"). From Fig. 3 we see that average midstance parameters sufficiently represent the changes between COP configurations.

During terminal stance, although average COP was significantly more lateral in the lateral device configuration, the remaining study parameters were not affected. This finding is consistent with previous studies which find no effect of medio-lateral device deviations at the second peak of KAM on knee kinetics of healthy subjects (Haim et al., 2008), and further, minimal or no effect on hip kinetics and kinematics (Solomonow-Avnon et al., 2016, 2015). This may occur for several reasons. Previous studies show increased variability in gait at the second peak (Haim et al., 2008; Hurwitz et al., 2002). Secondly, during midstance both device elements are in contact with the ground, however COP is located under the posterior half of the foot in early midstance, and thus may be primarily determined by the hindfoot device element, which is in closer proximity to the ankle. In contrast, during late terminal stance, COP is located under the anterior half of the foot as the heel has already lost contact with the ground, and is thus primarily determined by the forefoot device element. Thus, impact of COP manipulation during this time may be mediated by more complex foot biomechanics. Additionally, during midstance the limb is in a relatively more passive state of having accepted the weight and maintaining forward progression, with the contralateral limb closer to the centerline of the body. Contrary to this, in terminal stance, the limb is in a more active state, initiating weight transfer to the contralateral limb, which is in front of the centerline. These factors may make limb biomechanics more conducive to change via COP shift during midstance. Although lack of change in parameters during terminal stance in healthy subjects is in agreement with Haim et al. (2008) Haim et al. (2011, 2012) later confirmed that KAM was reduced during terminal stance in knee OA patients using the device. They suggested that increased KAM at the second peak, commonly seen in knee OA patients (Mündermann et al., 2005), made it more susceptible to reduction.

Several study limitations must be acknowledged. We did not use a normal-shoe control configuration to which we compare medial and lateral configuration parameters, but rather used the two clinically implementable device extremes, and compared the lateral configuration to medial. These configurations are set relative to a neutral configuration, which we also did not use as a control because, upon inspection of the data, the positioning varied across subjects and was closer to the lateral in some subjects and closer to medial in others, and thus does not represent an adequate reference. Additionally, we did not use regular shoes (or the device with no elements attached) as a control because a flat sole presents a different walking condition than the unstable convex elements, inducing less neuromuscular demand, and thus likely affecting the study parameters in a way that could not be controlled or accounted for. Additionally, the unique structure of our study device allowed us to compare a test configuration (lateral COP) to a "control" (medial COP) without upsetting the kinematic model. Specifically, reflective markers were not moved throughout the entire experimentation period and each subject was compared to himself. The only change was shift in biomechanical elements. This provides superior accuracy. Thus, the methodology of the present study sufficiently serves the specific study objective, and exhibits the mechanism by which the device reduces KAM in a healthy population.

In order to ensure precise marker placement, for obtaining accurate parameter values, a single experienced investigator carefully placed markers according to the 'PlugInGait' protocol. Small errors could have occurred in marker positions; however, since markers were not moved between test conditions, and each subject served as his own reference, any error would be constant across test conditions, and thus would not affect the answer to the study question of whether or not the parameters differed across COP configurations, or if the hypothesized KAM reduction mechanism can be confirmed.

Also of note is that we took the line connecting the lateral knee marker and KJC to approximate the knee flexion axis. This may not be a highly accurate representation of the flexion axis. However, our intention was to control for knee rotation about the vertical axis by choosing a plane containing the knee flexion axis. We believe that the reported values for KLA and KAM are more accurate than those reported in the laboratory frontal plane.

We must acknowledge that frontal plane shifts in COP may have an effect on ankle moments. We attempted to minimize this by positioning the biomechanical elements in the most extreme medial or lateral configurations without observable changes in ankle kinematics. Although this may avoid abnormal gait, ankle moments may still have been affected, and thus would likely have a confounding effect on KAM, independent of the GRF and KLA. Since foot markers were placed on the shoe, accurate ankle kinetics cannot be acquired since the foot may move within the shoe and relative marker positions would not remain constant (Kean et al., 2013). Thus, we did not report ankle moments. We believe that the reported underlying KAM reduction mechanism holds true regardless of contributions from ankle moment. Nevertheless, the confounding effect of ankle moments on the KAM warrants further investigation.

Finally, the study results pertain only to populations with characteristics similar to those of our cohort (i.e., healthy young males). Extrapolation of the results to pathological populations, or those dissimilar to our cohort, requires further study. Additionally, the study may be limited by the relatively small sample size of 14 healthy subjects. Nevertheless, the protocol design, in which each subject served as his own reference, with no disruption of the kinematic model between test conditions, allows detection of small changes in parameters with reasonable confidence. However, further study should validate the pilot findings in a larger cohort.

The evidence in the present study suggests that the mechanism of reducing KAM with the study device is largely via shortening of the KLA during a critical time within the stance phase of gait. Further, the results suggest that the study device causes a direct shortening of the KLA as a result of lateral deviation of the COP. Subsequent study is needed to determine the change in KLA required for clinical benefit.

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Conflict of interest statement

The authors have no conflict of interest to declare.

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