Lateral Wedges Alter Mediolateral Load Distributions at the Knee Joint in Obese Individuals

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ABSTRACT: Obesity is the primary risk factor for knee osteoarthritis (OA). Greater external knee adduction moments, surrogate measures for medial compartment loading, are present in Obese individuals and may predispose them to knee OA. Laterally wedged insoles decrease the magnitude of the external adduction moment in Obese individuals but it is unknown how they alter the center of pressure on the tibial plateau. A gait analysis was performed on 14 Obese (avg. 29.3 years; BMI range: 30.3–51.6 kg/m^2) and 14 lean women (avg. 26.1 years; BMI range: 20.9–24.6 kg/m^2) with and without a full-length, wedged insole. Computed joint angles, joint moments, and knee extensor strength values were input into a musculoskeletal model to estimate center of pressure of the contact force on the tibial plateau. Statistical significance was assessed using a two-way ANOVA to compare the main effects of group and insole condition ($\alpha = 0.05$). The insole resulted in a significant ($p < 0.01$) lateral shift in the center of pressure location in both the Obese and Control groups (mean: 2.9 $\pm$ 0.7 and 1.5 $\pm$ 0.7 mm, respectively). The insole also significantly reduced the peak external knee adduction moment 1.88 $\pm$ 1.82 N m in the Control group ($p < 0.01$) and 3.62 $\pm$ 3.90 N m in the Obese group ($p < 0.01$). The results of this study indicate the effects of a prophylactic wedged insole for reducing the magnitude of the load on the knee's medial compartment in Obese women who are at risk for knee OA development. © 2012 Orthopaedic Research Society. Published by Wiley Periodicals, Inc. J Orthop Res

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Obesity is the primary modifiable risk factor for knee osteoarthritis (OA).[1] Over two-thirds of Obese adults will develop knee OA during the course of a lifetime,[2] the majority females.[1] Knee OA typically occurs on the medial compartment of the tibial plateau, and estimates and measurements of the joint load are typically greater in the medial compartment than the lateral compartment.[3–8] The external knee adduction moment,[8,9] a model-based surrogate measure of medial tibiofemoral loading, suggests that medial joint loads are greater in Obese than lean individuals.[10,11] Although the relationship between joint load and OA progression has not been widely studied,[12] if the consequence of walking while carrying large amounts of excess body mass is high-force, repetitive loading on the tibial plateau, Obese individuals may be at a greater risk for development and progression of knee OA than lean individuals.[1,2] From the perspective of clinical biomechanics, we are therefore interested in assessing simple and non-invasive interventions for offloading the medial aspect of the tibiofemoral joint.

Laterally wedged insoles are one potential intervention that may delay the progression of knee OA by decreasing the magnitudes of the adduction moment, which is associated with disease onset and progression.[13–22] Lateral wedges are designed to shift the center of pressure of the ground reaction force (GRF) under the foot such that the GRF vector is directed closer to the center of the knee joint. The adduction moment is thus reduced by decreasing the frontal plane lever arm of the GRF. The majority of studies comparing gait kinetics with and without lateral wedges have had success using various types of wedges to decrease the magnitude of the adduction moment,[13–21] which presumably reduces the internal joint load as well.[22]

Although lateral wedges decrease the adduction moment in Obese individuals,[21] it is unknown how lateral wedges affect the magnitude and location of the knee joint contact force on the tibial plateau in an Obese population. These variables are important when assessing joint loading interventions because reductions in the adduction moment do not always ensure that the magnitude of the medial tibiofemoral load is concurrently reduced.[23] Contact force characteristics usually cannot be measured in vivo, but they can be estimated from model-based biomechanical analyses (e.g., Refs. [22,24]). Therefore, the purpose of this study was to determine the extent to which laterally wedged insoles could reduce the peak knee joint contact force and the peak medial location of the joint contact force during walking, particularly in Obese women. It was hypothesized that (1) walking with a laterally wedged insole would result in a more laterally located joint contact force location than a no-wedge condition and (2) a reduction in the adduction moment when walking with lateral wedges would be associated with a lateral shift in the predicted location of the joint contact force on the tibial plateau. It was also hypothesized that (3)

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both groups would experience similar changes to the insole intervention.

METHODS

Fourteen Obese women ("Obese" group; body mass index (BMI) value >30 kg m^{-2} and body fat percentages greater than 30%, as calculated from DEXA scans, to ensure the "Obese" criteria was based on both BMI and excess body fat) and 14 healthy-weight women ("Control" group; $20 \leq \text{BMI} < 25$ kg m^{-2}) Control women, as determined by an a priori sample size estimation,[13] gave written consent to participate. All participants were free of lower extremity injuries affecting gait and had no history of chronic knee pain or medical diagnosis of knee OA. Knee joint alignment was measured non-radiographically in all participants according to Vanwanseele et al.[25] and those exhibiting obvious varus or valgus malalignment were excluded.

Three-dimensional kinetic and kinematic data were collected during overground walking using an eight-camera Qualisys motion capture system (240 Hz; Oqus 300, Gothenburg, Sweden) set up around a centrally located force platform (1,200 Hz; AMTI, Inc., Watertown, MA). Timing sensors located 6 m apart on either side of the force platform were used to calculate walking speed to ensure that all subjects maintained a standard speed of 1.24 m/s. This speed was based on pilot data of the average preferred over ground walking speed of 10 Obese and 10 lean women within the age and BMI ranges included in this study.

Participants performed trials of overground walking with and without laterally wedged insoles in their shoes. The order of insole conditions was randomized. Spherical retro-reflective markers were placed on the pelvis and right lower extremity. Locations on the pelvis included the iliac crests, greater trochanters, anterior superior iliac spines, and the space between the fifth lumbar and first sacral vertebrae. Posterior superior iliac spine markers were used to help track the motion of the pelvis in the Obese group as markers on the anterior superior iliac spines can experience excessive movement in this population. Other markers were secured to the medial and lateral femoral epicondyles and malleoli. Locations on the foot were palpated through the shoe and included the first and fifth metatarsal heads and the distal toe. Rigid arrays of markers were secured to the lateral thigh, lateral leg, and posterior heel.

The insole was constructed of ethyl vinyl acetate material with an eight-degree inclination along the entire lateral aspect of the insole. The insole was worn for 30 min prior to data collection for familiarization. All participants wore the same make and model of shoe (New Balance RC 550) for all trials.

Kinematic data were digitized using Qualisys Track Manager software (Qualysis, Inc., Gothenburg, Sweden) and kinematic and analog data were input to Visual3D (C-Motion, Rockville, MD) for further analysis. Raw data were filtered using a Butterworth low pass filter (8 Hz for kinematic data; 50 Hz for analog data). A vertical GRF threshold of 15 N identified the start and end of the stance phase of gait, which was normalized to 101 data points. Kinetic and kinematic measures were calculated using a Cardan *XYZ* sequence of rotations (sagittal, fontal, transverse).[26] Three-dimensional hip, knee, and ankle joint angles were calculated. Joint reaction forces and moments were calculated from inverse dynamics analysis of a linked-segment model built from marker positions during upright quiet stance, with the moment acting on the distal segment resolved in the coordinate system of the proximal segment. Data for each participant were averaged over trials within each condition and used in all subsequent analyses. Only the clinically relevant frontal plane knee moments are reported in the results section; however, all of the joint moment data utilized in the optimization problem described in the next section are presented in Supplementary Material A.

In order to calculate the knee joint contact force and its center of pressure on the tibial plateau during walking, the forces in 34 lower limb muscles (Supplementary Material B) were estimated using static optimization. Muscle moment arms were defined as polynomial functions of the joint angles based on the equations reported by Menegaldo et al.[27] for the musculoskeletal model developed by Delp et al.[28]. Muscle moment arms were assumed to be dependent on joint angle and were not calculated differently between the Obese and lean subjects.

For each participant, the static optimization problem was to find the muscle forces at each 1% of stance that minimized the sum of the cubed muscle stresses, with each muscle treated as an ideal force generator.[29] Minimization of this quantity has been regarded as a general principle for predicting force patterns of synergistic muscles in human movement,[30] and the rise and fall of the predicted muscle forces is frequently similar to the timing of electromyograms measured during walking.[29,31] Muscle forces were constrained to replicate six resultant joint moments from the inverse dynamics, and to be non-negative. Three-dimensional hip moments, the knee flexion/extension moment, and the ankle subtalar and talocrural moments were included. The frontal plane knee moment was not included as a constraint in the optimization problem. A genetic algorithm (Genetic Algorithm & Direct Search Toolbox, Matlab, version 7.0.1, The MathWorks, Natick, MA) was used to find numerical solutions to the optimization problems.[32]

The line of action for each of the 12 muscles crossing the knee joint was estimated from the musculoskeletal model[28] across the stance phase in 1% increments. At each time point, the force in each knee joint muscle (Fm) was multiplied by its mediolateral moment arm to the knee joint center (Rm_x). The sum of all of these components was the muscular contribution to the frontal plane knee joint moment (M_{muscle}):

$$M_{\text{muscle}} = \sum_{m=1}^{12} (Fm \times \mathrm{Rm}_x) \tag{1}$$

A muscle's axial force component (Fm_{axial}) was calculated using the muscle orientation vectors along the axial and mediolateral axes of the tibia:

$$\theta = \tan^{-1}\left(\frac{\hat{Y}}{\hat{X}}\right) \tag{2}$$

$$Fm_{axial} = Fm \times \sin\theta \tag{3}$$

where θ is the angle of the actuator relative to the medial–lateral plane of the tibial plateau and X and Y are muscle orientation vectors along the mediolateral and axial axes of the tibia, respectively. The axial knee joint contact force ($Fcon_{axial}$) was calculated as:

$$Fcon_{axial} = JRF_{axial} - \sum_{m=1}^{12} Fm_{axial} \tag{4}$$

where JRF_{axial} is the axial component of the knee joint reaction force from the inverse dynamics analysis. The resultant frontal plane knee joint moment (M_{exp}) calculated from inverse dynamics is due to the combination of muscle forces, ligament forces, and joint contact forces. To calculate the location of the axial knee joint contact force, we assumed ligament forces were negligibly small during walking compared to muscle and joint contact forces (e.g., Refs. [14,24,29]). The moment due to joint contact forces (M_{con}) was calculated as:

$$M_{con} = M_{exp} - M_{muscle} \tag{5}$$

Finally, the location of the axial knee joint contact force was calculated by dividing the joint contact moment by the joint contact force:

$$COP = \frac{M_{con}}{Fcon_{axial}} \tag{6}$$

Center of pressure (COP) = 0 indicates a joint contact force acting directly through the knee joint center; a positive/negative COP indicates a medially/laterally located contact force, respectively. Figure 1 details the loads acting about the knee joint.

Peak values of the dependent measures were defined for each subject then overall averages and standard deviations were computed. Student's unpaired, two-tailed, t-tests assessed differences between the two groups for participant characteristics. A two-way (within- and between-factors) mixed-model ANOVA with factors for group (Obese, Control) and footwear (insole, no insole) tested statistical significance ($p < 0.05$) of the dependent variables (peak joint contact forces and the peak medial location of the COP of the knee joint contact force). To compare the strength of the relationship of selected dependent measures (COP shift and peak external adduction moment), effect sizes (d) were calculated to compare between conditions (insole, no insole) as per Cohen.[33]

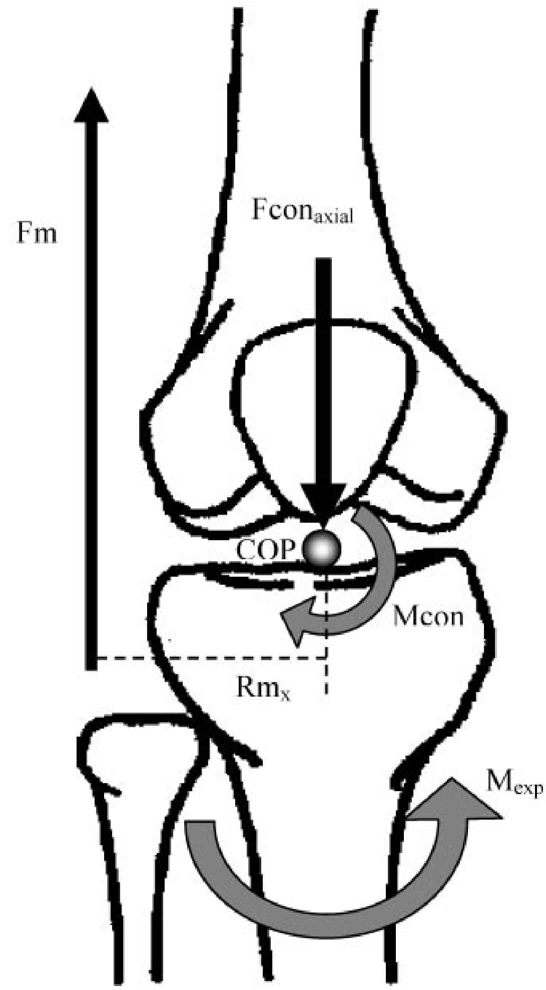

Figure 1. Loads acting about the knee joint. The external adduction moment acting about the knee joint must be balanced internally by the moments created by the muscle forces (Fm), their respective moment arms (Rm), and bone contact forces (Fcon).

RESULTS

Participant characteristics are presented in Table 1. Wearing the laterally wedged insole resulted in a significantly lower peak external knee adduction moment ($p < 0.01$; Fig. 2). This moment was reduced in the Control group by 6.6% (1.88 ± 1.82 N m, $d = 0.270$) and by 7.7% in the Obese group (3.62 ± 3.90 N m, $d = 0.305$; Fig. 2).

For both groups the peak medial position of the COP on the tibial plateau was significantly shifted laterally in the insole condition ($p < 0.01$) versus the no insole condition. Relative to the initial position of the COP, the Control group experienced an average absolute shift of 1.5 ± 0.7 mm, or 4.9% ($d = 0.211$), and the Obese group experienced an average absolute shift of 2.9 ± 0.7 mm, or 9.1% ($d = 0.454$; Fig. 3, Table 2). The Obese group had greater un-scaled peak contact forces than the Control group ($p < 0.01$) but wearing the insole did not significantly affect the peak values (Figure 4). There were no group-by-insole interactions

Table 1. Subject Characteristics (Mean ± Standard Deviation, [range])

	Age (years)	Height (m)	Mass (kg)	BMI (kg m^{-2})	Alignment (deg)
Control	26.1 ± 6.9	1.64 ± 0.07	60.1 ± 6.5	22.4 ± 1.2	−0.4 ± 2.9
	[20–48]	[1.50–1.75]	[47.7–67.7]	[20.8–24.6]	[−5.5–4.6]
Obese	29.3 ± 10.4	1.66 ± 0.07	102.3 ± 17.8*	37.2 ± 6.1*	−6.6 ± 2.1*
	[18–50]	[1.54–1.78]	[79.6–151.0]	[30.3–51.6]	[−8.9–4.6]

Negative alignment values indicate knee joint varus.
*Significant differences between groups.

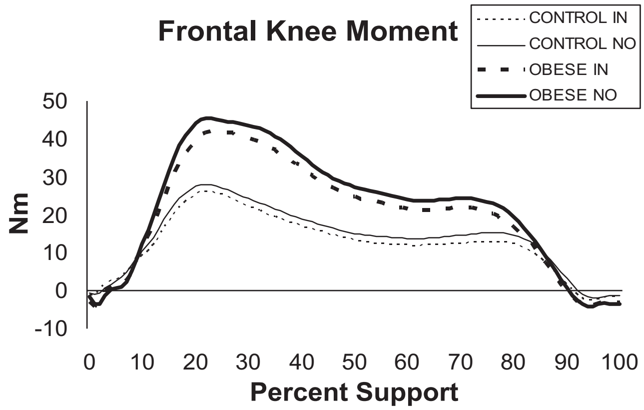

Figure 2. Ensemble averaged external knee adduction moments shown for the Obese and Control groups in the insole (IN) and no insole (NO) conditions. A significant decrease in the peak moment was found with the laterally wedged insole.

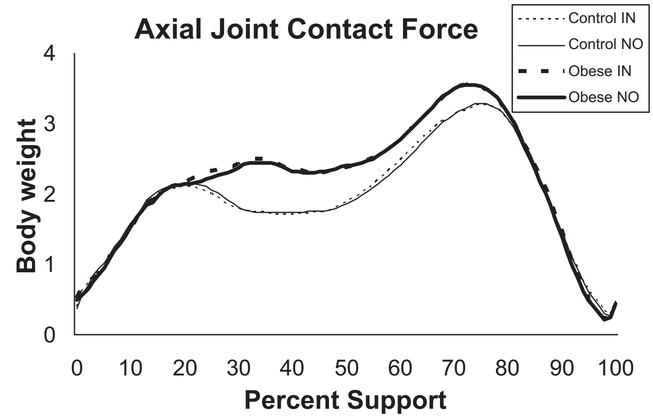

Figure 4. Ensemble averaged axial joint contact force in multiples of body weight. Values are shown for the Obese and Control groups in the insole (IN) and no insole (NO) conditions.

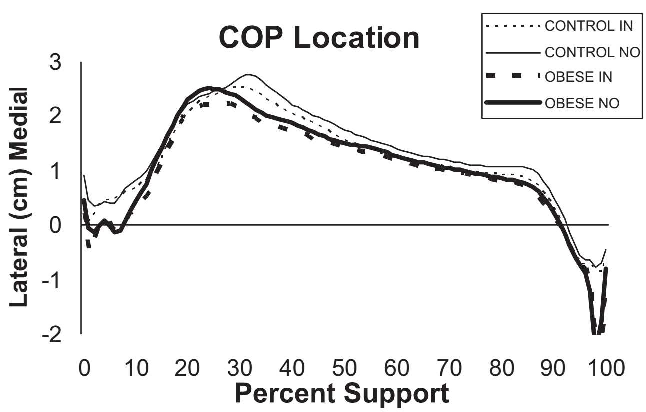

Figure 3. Ensemble averaged COP location of the contract force on the tibial plateau. A significant decrease in the peak medial COP location was found with the laterally wedged insole.

for any of the dependent measures and all pairwise comparisons not included above did not reach statistical significance ($p > 0.05$).

DISCUSSION

The present study is the first to our knowledge to investigate the effects of laterally wedged insoles on characteristics of the knee joint contact force during walking in Obese women. The insole intervention is of interest in this population, as they are prone to the development of knee OA.[1,2] Additionally, the intervention is low-cost and requires no surgery or medication. Previous studies have suggested that lateral wedges can reduce external knee joint loading during walking in lean individuals.[14,22] The present study corroborates this finding in Obese individuals, and adds evidence for the effect of wedged insoles on internal joint loading as well.

In support of the first and third hypotheses, an 8° lateral wedge resulted in a significant lateral shift of the COP of the joint contact force location than when walking with no-wedge in both the Obese group (2.9 ± 0.7 mm) and the Control group (1.5 ± 0.7 mm). This shift equates to approximately 2–4% of the width of the tibial plateau (assuming an 80 mm total width). This effect was only analyzed at the peak medial location but the contact force remains more laterally located throughout the gait cycle, which may be clinically relevant for reducing medial compartment loading over the course of thousands of strides each day. In support of hypothesis 2, the lateral wedge also reduced the peak adduction moment during walking as it concurrently resulted in a less medially located predicted peak location of the knee joint contact force. Effect

Table 2. Kinetic Variables

	COP Location (cm)		Axial Contact Force (N)		Scaled Axial Contact Force (Multiples of Body Weight)	
	No Insole	Insole	No Insole	Insole	No Insole	Insole
Control	2.98 ± 0.75	2.82 ± 0.70	2,000.7 ± 622.5	2,001.4 ± 634.1	3.33 ± 0.79	3.33 ± 0.81
Obese	2.94 ± 0.68	2.65 ± 0.58	3,703.0 ± 1,481.6*	3,699.8 ± 1,448.8*	3.61 ± 0.90	3.60 ± 0.91

*Significant differences between groups.

sizes for the change in the peak adduction moment with lateral wedges indicated that there was a small effect present.

Although the un-scaled peak adduction moment was greater in the Obese group, the medial location of the COP was not significantly different between groups and the groups responded similarly to the insole condition. Therefore, if we presume the Obese subjects are at a greater risk of developing knee OA than the lean subjects, the greater magnitude of loading, rather than the location of that load, may be more responsible for cartilage deterioration and early knee OA onset in Obese populations.

Despite being statistically significant, the magnitude of the shift in COP was relatively small. Given these small magnitudes of distance, and the fact that they were predicted using a model-based analysis, there are several limitations that affect the confidence level of the predictions. Soft tissue artifacts in the motion capture data, errors in defining the anatomical knee joint rotational axes, lack of subject-specific muscle geometry and force production dynamics, and neglecting possible ligament contributions could all potentially influence the results. However, some of these limitations are of minor concern. For example, previous studies have suggested that muscle contractile dynamics have a minimal effect on model-based muscle force predictions,[31] and that knee muscle and ligament forces are not simultaneously large during the stance phase of walking.[6] Thus, we would speculate that the general conclusions of this study should be robust to changes in the details of the musculoskeletal model, even if the exact numerical results are subject to a degree of uncertainty.

A specific modeling limitation was the knee and contact force models, which considered only the sagittal plane of the knee in computing muscle force and only the frontal plane of the knee when computing contact force magnitude and location. The former assumption was necessary because we cannot assume the knee joint moments outside the sagittal plane are due primarily to muscles, and because modeling the muscle and contact forces simultaneously would require a more detailed model with more assumptions (e.g., material properties)[34] than there was sufficient data for. The statistical design of the study was within-subjects (i.e., knee joint parameters were not different between conditions), so we would not expect different conclusions if the study was repeated with more detailed or patient-specific knee models. However, we can only speculate on this topic as we have only used the model described in the present study. The latter assumption (contact force computed from the frontal plane only) neglects contributions to the contact force magnitude and location outside the frontal plane, but does not affect our primarily outcome variable (the mediolateral shift in the contact force location).

We also assumed that the relationships between muscle moment arms and joint positions were the same for all subjects. Comparisons of moment arms in the lower limb between lean and Obese individuals are scarce in the literature. Wood et al.[35] reported a small difference (<0.5 cm) in the psoas moment arm between lean and Obese males. Generic moment arm models can misrepresent the offset of the moment arm—joint angle relationship compared to MRI-based models,[36] but given the within-subjects study design, qualitatively different results from using personalized moment arm models would not be expected.

However, given these limitations of this study, we suggest that the results are most appropriately interpreted as "proof of concept" the ability of lateral wedges to alter the location of the knee joint contact force during walking in a manner that may reduce the loading of the medial tibiofemoral compartment. Although a cross-sectional, not longitudinal, investigation was performed, Keating et al.[37] reported that lateral wedges improved functional pain scores over a 1-year follow-up in patients with knee OA, particularly when they were implemented prior to severe joint deterioration. Decreasing the magnitude of the load on the medial compartment of the knee may enable Obese individuals to exercise more effectively for weight loss, while concurrently reducing the primary biomechanical risk factors for knee OA development.

This study was performed in an Obese, but otherwise healthy, population with no history of knee OA. The results are also limited to individuals with no substantial valgus knee alignments. A valgus knee alignment creates an anatomical predisposition for lateral compartment knee OA, which would not be aided with the insoles used in the present study, although they may be beneficial for neutral or varus knee alignments. It remains to be seen what degrees of change in joint contact force characteristics are needed for positive clinical outcomes on the treatment and prevention of knee OA, or similarly, what long-term effects are induced by unloading the medial cartilage (and presumably loading the lateral cartilage) in populations at risk for knee OA. Methods that reduce or alter biomechanical risk factors for medial compartment loading may potentially be used to delay or prevent disease onset in this high-risk populations for knee OA, instead of being used as temporary treatments after the disease develops.

The adduction moment is often used to indirectly assess knee joint loading during walking in knee OA patients or to assess risk for knee OA (e.g., Refs. [3,12,37,38]). In agreement with previous studies,[22] it was not expected that the overall contact force would change with the insole (given that, e.g., subjects had the same mass, walked at the same velocity) but the insole was able to alter the *location* of the knee joint contact force. Laterally wedged insoles may reduce the peak adduction moment during stance without altering the total knee joint contact force. Although we did not model separate medial and lateral tibiofemoral contact, the lateral shift in the location of the net joint contact force that we found would be associated with some of the net load being transferred from the medial side to the lateral side of the knee. Lateral wedges may therefore be beneficial for reducing medial tibiofemoral loading during walking. Data from instrumented joint replacements have indicated that changes in the medial tibiofemoral contact force are not always reflected by changes in the adduction moment.[23] In agreement with Walter et al.,[23] we suggest that clinical gait analyses of osteoarthritic gait should consider a variety of variables when assessing knee joint loading, rather than focusing on only the adduction moment. Model-based predictions of joint contact force characteristics can be useful complimentary variables for this purpose.

CONCLUSION

An 8° laterally wedged insole was effective in decreasing the medial displacement of the COP of the contact force on the tibial plateau across the stance phase of walking in groups of both lean and Obese women. The results of this study indicate the effects of a prophylactic wedged insole for reducing a portion of the load born by the medial compartment in Obese women who are at risk for knee OA development. The benefits of laterally wedged insoles have been shown in knee OA populations but they may also be effective for preventing or delaying disease development when used over thousands of strides each day in high-risk individuals. A specific goal in Obese populations is to reduce the overall risk of knee OA by encouraging weight loss. This may be accomplished via the most popular form of exercise available (walking) while shifting a portion of the load away from the medial compartment of the knee joint and potentially limiting the development of knee pain due to cartilage damage. Future longitudinal research studies are needed to determine if wedged insoles can delay disease onset in Obese women.

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