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Lateral wedges decrease biomechanical risk factors for knee osteoarthritis in obese women

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ABSTRACT

Obesity is the primary risk factor for the development and progression of medial compartment knee osteoarthritis. Laterally wedged insoles can reduce many of the biomechanical risk factors for disease development in osteoarthritis patients and lean individuals but their efficacy is unknown for at-risk, obese women. The purpose was to determine how an 8° laterally wedged insole influenced kinetic and kinematic gait parameters in obese women. Gait analysis was performed on fourteen obese (average 29.3 years; BMI 37.2 kg/m^2) and 14 lean control women (average 26.1 years; BMI 22.4 kg/m^2) with and without a full-length, wedged insole. Peak joint angles, the external knee adduction moment and its angular impulse were calculated during preferred and standard 1.24 m/s walking speeds. Statistical significance was assessed using a 2-way ANOVA (α=0.05). The insole significantly reduced the peak external knee adduction moment (mean decrease of 3.6 ± 3.9 Nm for obese and 1.9 ± 1.8 Nm for controls) and its angular impulse in both groups. The wedged insoles also produced small changes in ankle dorsiflexion (obese: $1.2 \pm 1.4^\circ$ increase; control: $1.5 \pm 1.4^\circ$ increase) and eversion range of motion (obese: $1.3 \pm 1.9^\circ$ decrease; control: $1.5 \pm 1.2^\circ$ decrease) but did not alter peak angles of superior joints. Although the majority of obese women may develop knee osteoarthritis during their lifetime, a prophylactic insole intervention could allow obese women with no severe knee malalignments to be active while preventing or delaying disease onset. However, the long-term effects of the insole have not yet been examined.

1. Introduction

Obesity is the primary, modifiable risk factor for both the development (Anderson and Felson, 1998; Felson et al., 1998) and the progression (Ledingham et al., 1995) of bilateral knee osteoarthritis (KOA). Obese individuals have a four-fold greater incidence of KOA than their healthy-weight counterparts because of increased loading on the joint resulting from greater body weights (Felson et al., 1998). Two-thirds of obese individuals will develop the disease throughout the course of a lifetime (Murphy et al., 2008) and the majority will be females (Felson et al., 1998).

KOA typically develops in the medial compartment because the internal knee joint contact loads are greater there than on the lateral side during the stance portion of gait (Andriacchi, 1994; Zhao et al., 2007). Although not a direct measure of loading, the external knee adduction moment (EKAM) is commonly measured due to its strong association with the medial compartment load (Hurwitz et al., 1998; Schipplein and Andriacchi, 1991; Shelburne et al., 2008; Zhao et al., 2007). Greater EKAM values may predispose individuals to KOA (Amin et al., 2004; Astephen and Deluzio, 2005; Baliunas et al., 2002; Miyazaki et al., 2002). The angular impulse of the EKAM gives an indication of loading across the stance phase instead of at a discrete instance. It is also greater in obese individuals (Russell et al., 2010) and in individuals with KOA (Thorp et al., 2006) compared to lean and asymptomatic individuals

During walking, forces acting on the lower extremity produce an EKAM due to the relatively large mediolateral moment arm between the knee joint center and the ground reaction force vector (Andriacchi, 1994; Schipplein and Andriacchi, 1991). The magnitude of the peak EKAM is also associated with the severity of KOA (Astephen et al., 2008) and how quickly it progresses (Miyazaki et al., 2002). Obese individuals with no KOA have greater peak EKAMs than healthy-weight individuals (Browning and Kram, 2007). Thus, even prior to diagnosis, obese individuals appear to be already at greater risk of developing KOA than lean individuals.

Orthotics and insoles are often used to modify lower extremity alignment and loading patterns. Laterally wedged insoles are designed to laterally shift the center of pressure under the foot

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and shift the mechanical axis of the limb (Toda et al., 2001; Yasuda and Sasaki, 1987) such that the moment arm of the EKAM decreases and the load distribution shifts towards the lateral compartment of the knee joint (Crenshaw et al., 2000; Shelburne et al., 2008). This center of pressure shift may decrease the peak medial compartment load and its surrogate measure, the peak EKAM (Erhart et al., 2008; Haim et al., 2008; Shelburne et al., 2008). Even small (1 mm) shifts in the center of pressure under the foot can significantly decrease the EKAM and peak medial compartment load (Shelburne et al., 2008). The majority of studies (Butler et al., 2007; Crenshaw et al., 2000; Erhart et al., 2008; Fisher et al., 2007; Kakihana et al., 2004; Kerrigan et al., 2002), but not all studies (Maly et al., 2002; Nester et al., 2003; Schmalz et al., 2006), have reported successes using lateral wedges to decrease the peak EKAM moment and its impulse (Haim et al., 2008).

When using lateral wedges, individuals with greater peak EKAMs at baseline experienced greater reductions in this measure than subjects with lower baseline values (Fisher et al., 2007). Obese individuals and KOA patients typically have greater EKAM values than their healthy-weight counterparts and it is possible that obese, but otherwise healthy, individuals could greatly benefit from a minimally intrusive wedged insole during walking.

The multiple joints within the foot and the ankle joint are the first joints within the linked chain of the lower extremity to be influenced by the wedges. Lateral wedges increase peak rearfoot eversion and eversion excursion (Butler et al., 2009). Obese women already have greater range of eversion motion than non-obese women (Messier, 1994) and how wedges affect the ankle joint and superior joint kinematics in this population is open to question. Numerous studies have shown the benefits of lateral wedges for treating KOA but no study has attempted a proactive approach and introduced these insoles to the most high-risk population: obese women.

Therefore, the purpose of this study was to analyze the influence of a laterally wedged insole on the lower extremity kinetics and kinematics of obese and healthy-weight women. We hypothesized that (1) the insole would significantly reduce the peak magnitude of the EKAM and its angular impulse in both groups; (2) the obese group would have higher baseline values and would therefore experience a greater reduction in the peak values than their lean counterparts; and (3) the insole intervention would not affect the kinematics of the knee and the hip but would affect the ankle kinematics in terms of increased rearfoot eversion and eversion range of motion.

2. Methods

An *a priori* sample size estimation was performed (Crenshaw et al., 2000) and 14 obese women (BMI $\geq$ 30 kg/m^2) and 14 healthy-weight control women (20 $\geq$ BMI > 25 kg/m^2) participated. Participants were free of lower extremity injuries affecting gait and had no knee pain. Although radiographs were not included, participants had never been previously diagnosed with KOA nor had they ever experienced or sought medical treatment for KOA-related symptoms. All participants completed an informed consent in accordance with the Human Subjects Research Policies set forth by the University of Massachusetts Institutional Review Board.

Static knee joint alignment was measured in all the subjects according to the procedures described by Vanwanseele et al. (2009) and calculated in visual 3D (C-Motion, Inc., Rockville, MD, USA) as the angle formed by the mechanical axes of the femur and tibia in the frontal plane. To remove the influence that severe frontal plane knee malalignments have on KOA risk, participants were excluded if they exhibited a varus or valgus knee alignment greater than 10°. A small range of values has been reported as "normal" frontal plane knee joint alignment (Moreland et al., 1987; Sogabe et al., 2009) and in obese individuals "normal" alignment tends towards abnormally large Q-angles and valgus alignment relative to controls (Messier et al., 2000), yet medial compartment KOA is still an issue. Therefore, a range of ±10° from 180° was used to account for the confounding factor of increased tissue thickness over the joints of the obese women while still excluding moderate-to-severe malalignments.

Body composition was measured using a dual X-ray absorptiometry scanner (Lunar Prodigy, GE Lunar Corp, Madison, WI). Three-dimensional kinetic and kinematic data were collected using an eight camera Qualisys Oqus 300 system (240 Hz; Qualysis, Inc., Gothenburg, Sweden) synchronized with, and set up around, a centrally located force platform (1200 Hz; AMTI, Inc., Watertown, Massachusetts, USA). Timing sensors (Lafayette Instrument Co., Lafayette, IN, USA) were used to determine walking speed through the capture area.

Data on each participant were collected in a single testing session. All participants wore the same make and model of shoe during data collection (New Balance RC 550). The laterally wedged insole consisted of ethylene vinyl acetate material with an 8° inclination along the entire lateral aspect of the insole (Fig. 1). The insole was worn during a 30-min habituation period prior to data collection.

Spherical retro-reflective markers were placed on anatomical landmarks and segments of the right lower extremity (Fig. 2). On the obese subjects, ASIS markers were often moved laterally to avoid excessive motion of the underlying tissue and PSIS markers were included to help track the motion of the pelvis.

Participants walked at a standard speed (SS) of 1.24 m/s. This speed was based on the mean preferred overground walking speed of 10 obese and 10 healthy-weight pilot participants who were within the ranges of ages included in this study. Participants also walked at their preferred speed (PS): the mean speed during six passes between the timing sensors. Participants walked in the insole and no-insole conditions, in both the PS and SS conditions, across the force platform 10 times each. The order of insole and speed conditions was randomized. Both speed conditions were used to allow for greater comparison opportunities with previously published studies.

Kinetic and kinematic data were imported into Visual 3D software for further analysis. Raw data were low-pass filtered using a fourth order, dual pass Butterworth low-pass filter with cutoff frequencies of 8 Hz for the kinematic data and 50 Hz for the ground reaction force data. A vertical ground reaction force threshold of 15 N identified initial foot contact and toe-off of the right foot.

Three-dimensional joint angles and range of motion were calculated using an *XYZ* rotation sequence. A Newton–Euler inverse dynamics method was used to calculate joint moments. The kinetic variables, peak EKAM and the angular impulse of the EKAM across the time series, were then determined. Data were calculated in absolute, un-scaled values because the absolute magnitudes of these variables are responsible for the medial compartment loading that can lead to KOA (Segal et al., 2009). Data during the stance phase were interpolated and

Fig. 1. Laterally wedged insoles used in this study. These ethylene vinyl acetate insoles had an 8° posting along the entire lateral border.

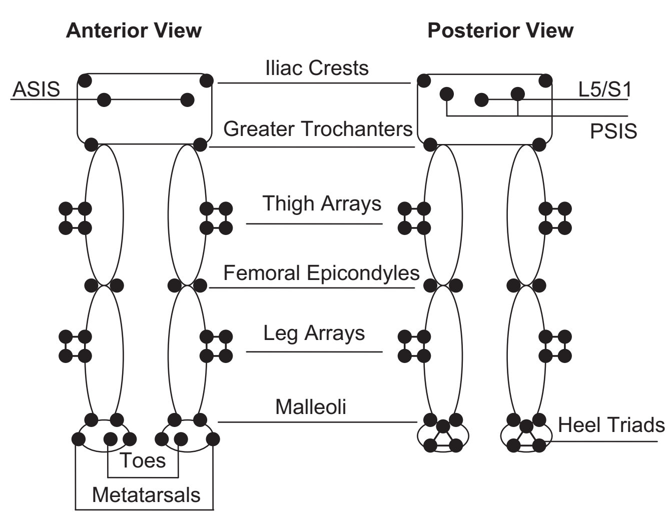

Fig. 2. Marker positions on the lower extremity.

standardized to 101 data points. Data from the 10 trials were then averaged within each subject. Group means and standard deviations within each condition were calculated.

T-tests assessed differences for subject characteristics of age, height, mass, BMI, percent body fat, knee alignment and preferred walking speed ($\alpha=0.05$). Each participant's mean data from the 10 trials were input into a two-way ANOVA (group—obese, control; insole condition—insole, no insole) to test statistical significance of the dependent variables.

3. Results

Participant characteristics are presented in Table 1. There were no significant differences between obese and control group's age, height, and PS; however, body mass ($p<0.01$), BMI ($p<0.01$), percent body fat ($p<0.01$), and knee alignment ($-$valgus/$+$varus) ($p<0.01$) were all significantly greater in the obese group. All obese participants and 10 control subjects exhibited some degree of valgus knee alignment. Results are presented in terms of the SS; PS values are presented when the results differ from the SS.

There were no significant interactions between group and insole condition such that the obese group did not respond with significantly greater reductions in the peak moment than the control group. However, walking with an 8° laterally wedged insole significantly reduced the peak EKAM in both groups and during both walking speeds ($p<0.01$) (Fig. 2). The control group reduced their peak moment by an average of 1.8 ± 2.2 Nm and the obese group reduced their peak moment by an average of 4.5 ± 4.8 Nm when walking at the PS. At the SS the control group reduced their peak moment by 1.9 ± 1.8 Nm and the obese group by 3.6 ± 3.9 Nm.

The effect of the group was also significant during both walking speeds ($p<0.01$) such that the obese group exhibited greater peak moments than the controls. During the SS the control group's average values were 26.6 ± 6.9 and 28.4 ± 7.0 Nm for the insole and no-insole conditions, respectively, and 43.4 ± 10.5 and 47.0 ± 13.1 Nm, respectively, for the obese group. During the PS, for the insole and no-insole conditions, the control group's average values were 29.3 ± 7.5 and 31.1 ± 7.9 Nm, respectively, and the obese group's were 45.3 ± 10.6 and 49.8 ± 11.9 Nm, respectively.

There were no significant interactions between group and insole conditions for the angular impulse values. The use of an 8° laterally wedged insole decreased the angular impulse of the EKAM in both groups and during both the PS and SS ($p<0.01$) (Fig. 3). The obese group also had significantly greater angular impulse values than the control group ($p<0.01$).

The influence of the insole on hip, knee, and ankle kinematics was compared to examine how the insole impacted surrounding joints. Fig. 4 presents joint angles across the stance phase of gait while walking at the SS. Table 2 presents selected joint angles and ranges of motion (ROM) during the SS walking condition. In both groups the insole had no significant effect on hip and knee joint kinematics. At the ankle, the insole significantly increased dorsiflexion in late stance. The insole did not produce greater peak eversion angles; however, the frontal plane profiles show that the ankle remained more everted throughout early stance in the insole condition. The insole also significantly increased the ankle eversion ROM (due to greater initial inversion). When comparing groups, the obese group walked with greater hip flexion and adduction and less knee adduction (Fig. 5).

4. Discussion

The purpose of this study was to determine if a laterally wedged insole could decrease kinetic risk factors for KOA while having a minimal impact on the joint kinematics in obese and healthy-weight women. In support of the first hypothesis, the addition of a laterally wedged insole reduced two of the primary biomechanical variables associated with KOA: the peak EKAM and its angular impulse. This peak moment represents a discrete instance of loading while the angular impulse of the adduction moment expresses loading across the stance phase.

Contrary to the second hypothesis that the obese group would experience greater reductions in the EKAM, both groups experienced similar reductions in EKAM with the use of a laterally wedged insole as participants in previous studies (Crenshaw et al., 2000; Fisher et al., 2007; Hinman et al., 2008). In the current study, the control subjects reduced their peak moment 6.6% and 5.2% for the SS and PS, respectively, and the obese reduced theirs by 7.7% and 9.1% for the SS and PS, respectively. These values are within the ranges reported in previous studies on subjects with no KOA. However, previous research has reported wide ranges of reductions in the peak moment for a variety of different degrees of wedging. Reductions ranging from 3.2% (Crenshaw et al., 2000)

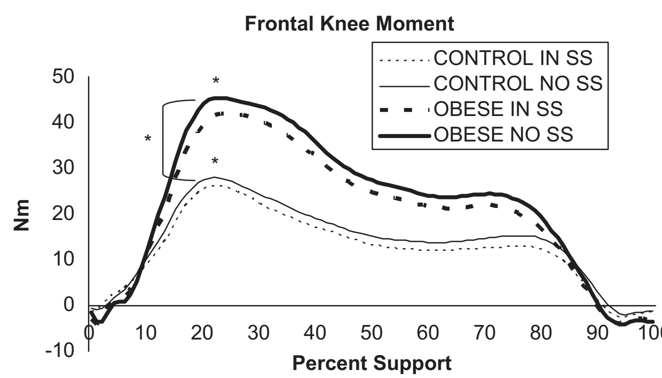

Fig. 3. Mean external knee adduction moment across the stance phase of walking for each group; * indicates statistical significance at peak values.

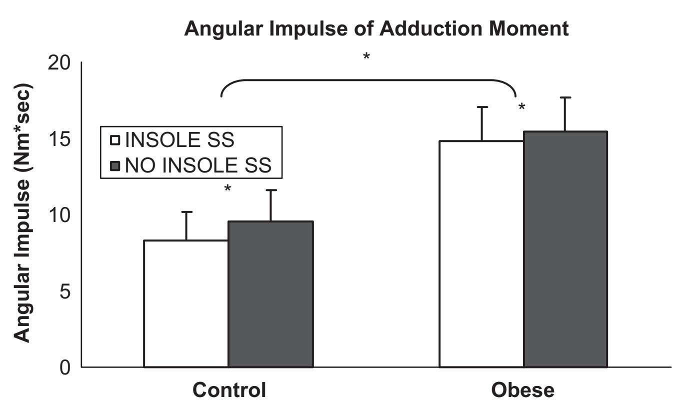

Fig. 4. Angular impulse of the external knee adduction moment for each group in both the insole and no-insole conditions. These data show the impulse from the standard speed (SS) condition; * indicates statistical significance.

Table 1
Subject characteristics; PS—preferred speed.

	Age (years)	Height (m)	Mass (kg)	BMI (kg/m^2)	Body fat (%)	Alignment (deg.)	PS (m/s)
Control	26.1 ± 6.9	1.64 ± 0.07	60.1 ± 6.5	22.4 ± 1.2	26.7 ± 6.1	-0.4 ± 2.9	1.46 ± 0.13
Obese	29.3 ± 10.4	1.66 ± 0.07	102.3 ± 17.8	37.2 ± 6.1	46.2 ± 4.9	-6.6 ± 2.1	1.36 ± 0.15

Table 2
Selected hip, knee, and ankle joint angles during walking at the set speed.

Joint motion	Control		Obese		*P* values	
	Insole	No insole	Insole	No insole	Effect of group	Effect of insole
Sagittal plane						
Hip flexion	23.4 ± 5.4	23.8 ± 5.6	28.6 ± 6.0	28.3 ± 6.1	0.034*	0.977
Hip extension	−16.3 ± 4.2	−16.3 ± 3.5	−12.8 ± 6.2	−13.4 ± 5.9	0.108	0.441
Hip ROM	39.8 ± 3.9	40.1 ± 4.0	41.5 ± 3.8	41.7 ± 2.9	0.232	0.306
Knee flexion	−15.5 ± 5.2	−15.8 ± 5.5	−17.0 ± 5.3	−17.5 ± 4.9	0.418	0.180
Ankle PF	−6.0 ± 3.8	−6.7 ± 4.1	−6.6 ± 3.1	−6.2 ± 2.8	0.957	0.586
Ankle DF	13.0 ± 3.1	11.5 ± 3.0	14.7 ± 3.5	13.5 ± 3.6	0.150	< 0.001*
Frontal plane						
Hip adduction	8.8 ± 2.4	8.4 ± 2.3	12.5 ± 3.4	12.3 ± 3.1	0.001*	0.258
Knee adduction	−0.4 ± 3.0	−0.2 ± 3.1	−3.4 ± 3.0	−3.2 ± 2.9	0.014*	0.050
Ankle eversion	−4.9 ± 2.4	−5.0 ± 2.1	−3.5 ± 3.9	−3.2 ± 3.1	0.144	0.850
Ankle ROM	12.4 ± 3.3	10.9 ± 3.0	12.4 ± 2.8	11.1 ± 2.3	0.965	< 0.001*

In bold are *P*-values for parameters with significant differences between the control and obese groups or the insole and no-insole conditions; DF—dorsiflexion, PF—plantar flexion, and ROM—range of motion.

* indicates statistical significance.

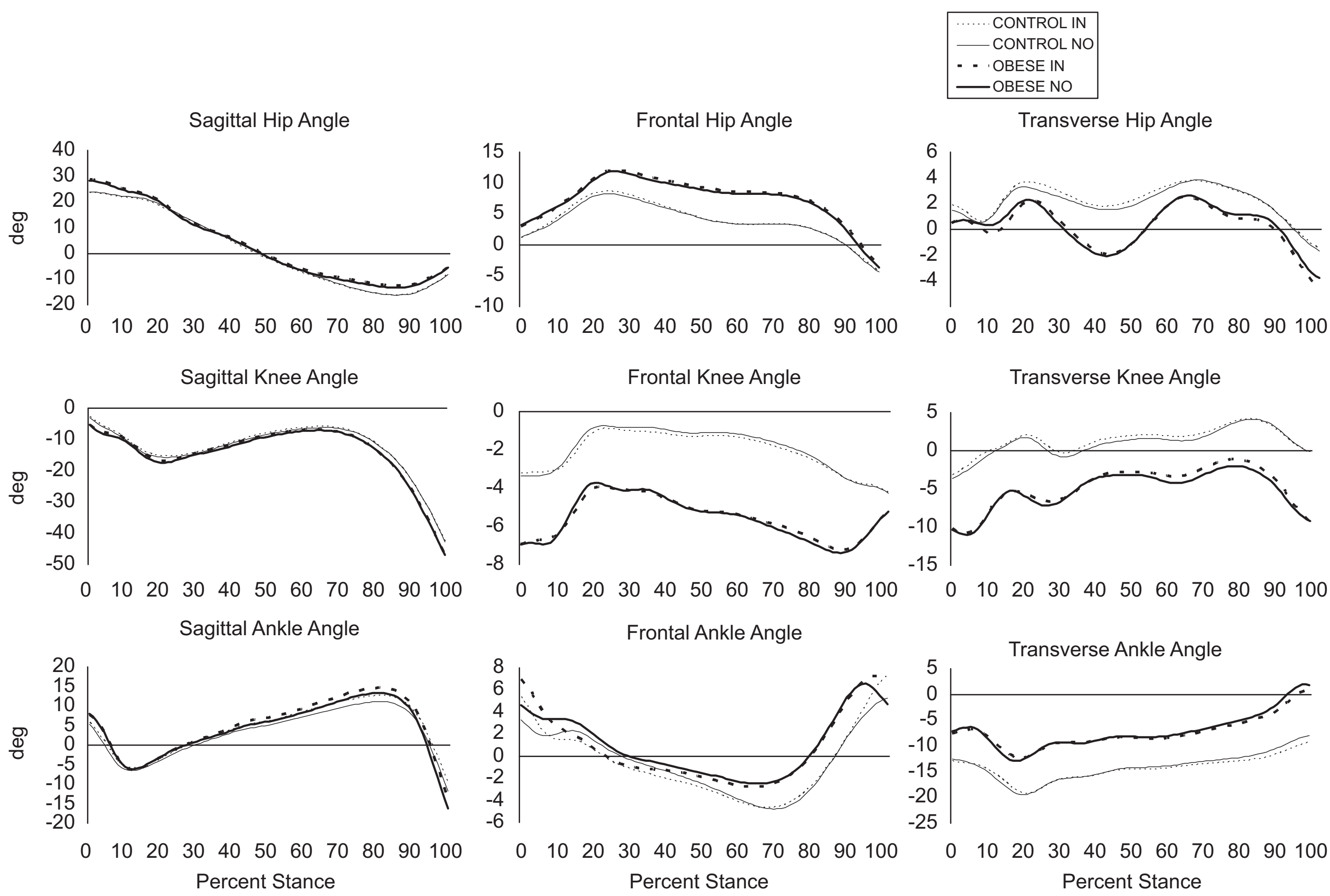

Fig. 5. Three-dimensional joint angles during walking at the standard speed (SS) for both groups in both the insole and no-insole conditions. The obese group is shown in thick lines and the control group is shown in thin lines. Solid lines indicate the no-insole (NO) condition while dashed lines indicate the insole (IN) condition.

to 11.9% (Hinman et al., 2008) were seen when using 5° wedges and Fisher et al. (2007) found reductions of almost 15% and 19% when using 4° and 8° full shoe wedges, respectively. Yet, other studies using larger angles of inclination of 10° (Nester et al., 2003) and 14° (Schmalz et al., 2006) found no reductions in the peak moment when using lateral wedges with healthy subjects. Thus, although previous research supports the efficacy of lateral wedges for reducing the EKAM, the results are varied and do not always correspond to the degree of wedging. Material makeup of different insoles may play a role (Fisher et al., 2007) and has consisted of materials from nickleplast (Crenshaw et al., 2000) to ethyl vinyl acetate (Hinman et al., 2008; Nester et al., 2003); however, the influence of different material properties was not studied.

In subjects with KOA, who tend to have greater peak EKAM and angular impulse values than asymptomatic lean individuals

(Thorp et al., 2006), insoles have been effective at reducing the EKAM between 6% and 10% at inclinations ranging from 5° to 10° (Kerrigan et al., 2002; Kakihana et al., 2004). Therefore, the obese group, who had a 7.7% reduction in peak moment with an 8° wedge at the SS, may behave and respond to the insole more similarly to KOA patients than the control group. Although the positive effects of the insole were not significantly greater in the obese group than the control group, an insole intervention to alleviate this biomechanical risk factor for KOA still may be particularly beneficial for this high-risk population.

Shelburne et al. (2008) used a musculoskeletal modeling approach to estimate how laterally shifting the center of pressure under the foot influenced the EKAM. These researchers reported that each 1 mm displacement of the center of pressure of the ground reaction force vector resulted in a 2% reduction in the medial load. Maly et al. (2002) also found a significant relationship between the experimentally measured center of pressure displacement and the EKAM. If the center of pressure/EKAM relationship from Shelburne et al. (2008) was applied to the current study, it could be estimated that the insole would laterally shift the center of pressure an average of 3.9 mm in the obese group and 3.3 mm in the control group.

The reduction in the EKAM was maintained across the stance phase in both groups. Angular impulse provides a quantitative, albeit surrogate, measure of how the medial compartment is loaded during the entire stance phase of gait. An 8° laterally wedged insole was effective at reducing the impulse of the EKAM in both groups. Haim et al. (2008) laterally shifted the center of pressure on the foot with specialized footwear and found a decrease in the impulse. Thorp et al. (2006) reported the impulse of the EKAM to be greater in individuals with KOA, even when they walked at slower speeds and peak moment values did not differ. Although this variable is not as widely reported as the peak EKAM, it is an important surrogate measure of the load borne by the medial compartment throughout stance that can be reduced with laterally wedged insoles.

The insole affected some aspects of joint kinematics and several of the components of the third hypothesis were supported. At both the SS and the PS, the insole increased ankle eversion ROM, as expected, but it did not increase the peak eversion value. This result signifies that peak inversion was greater in the insole condition to produce the greater ROM. The lack of significantly greater peak ankle eversion with the insole was unexpected since a consequence of the insole is to rotate the foot segment into a more everted position. However, the ankle everted earlier in support and to a greater degree up until the peak eversion value, but the wedge did not cause the ankle to experience greater peak eversion than walking without the wedge. The lack of greater peak eversion with the insole is an important result because the magnitude of peak ankle eversion has implications for soft tissue injury, particularly at the knee (Bates et al., 1979; Tiberio, 1987). It appears that the insole could be a successful strategy for reducing key biomechanical variables, such as the EKAM and its angular impulse, while having little impact on the peak motions of surrounding joints.

There were several limitations associated with this study. The aim of this study was to investigate the acute effects of the insole to determine if a long-term intervention may be warranted. Therefore, although the insole reduces several risk factors for disease development, it is currently unknown if these insoles may prevent or delay KOA. The length of time that the insole would need to be worn to experience the benefit is also unknown at this time. Additionally, it is unknown if all obese individuals would experience the same benefit from the insole. Two subjects in each group in this study experienced either no noticeable change with the addition of the insole or a minor increase in the EKAM. Therefore, it may be possible that not every individual experiences the same benefits of the insole.

The goal of this study was to provide a minimally intrusive intervention to reduce risk factors for KOA in obese women. Lateral wedges were effective at reducing the peak EKAM and its angular impulse during walking, while having minimal negative impacts on the kinematics of the superior joints. The majority of obese women will develop KOA throughout the course of a lifetime and this minimally intrusive intervention may be able to prevent or delay disease onset in obese women who do not have substantial varus/valgus knee malalignments. These results serve as a foundation for future research designed to use prophylactic insole interventions as preventative strategies for KOA instead of temporary treatments.

Conflict of interest statement

None.

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