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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.

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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 $\pm 10°$ 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.

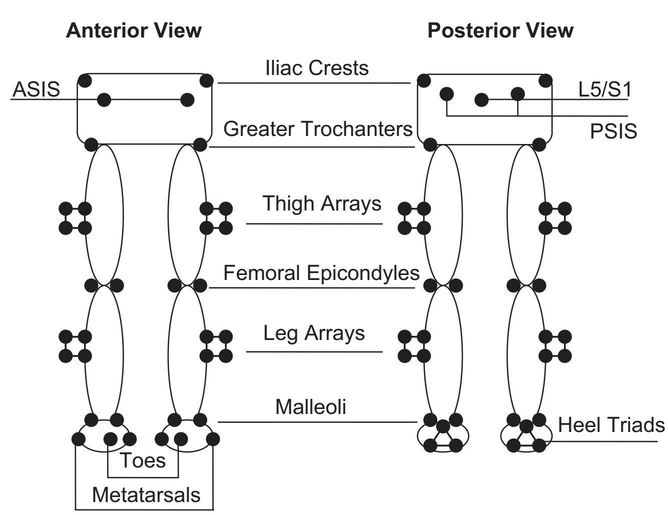

Fig. 2. Marker positions on the lower extremity.

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