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Redistribution of knee stress using laterally wedged insole intervention: Finite element analysis of knee–ankle–foot complex

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ABSTRACT

Background: Laterally wedged insoles are widely applied in the conservative treatment for medial knee osteoarthritis. Experimental studies have been conducted to understand the effectiveness of such an orthotic intervention. However, the information was limited to the joint external loading such as knee adduction moment. The internal stress distribution is difficult to be obtained from in vivo experiment alone. Thus, a three-dimensional finite element model of the human knee–ankle–foot complex, together with orthosis, was developed in this study and used to investigate the redistribution of knee stress using laterally wedged insole intervention.
Methods: Laterally wedged insoles with wedge angles of 0, 5, and 10° were fabricated for intervention. The subject-specific geometry of the lower extremity with details was characterized in the reconstruction of MR images. Motion analysis data and muscle forces were input to drive the model. The established finite element model was employed to investigate the loading responses of tibiofemoral articulation in three wedge angle conditions during simulated walking stance phase.
Findings: With either of the 5° or 10° laterally wedged insole, significant decreases in von Mises stress and contact force at the medial femur cartilage region and the medial meniscus were predicted comparing with the 0° insole.
Interpretation: The diminished stress and contact force at the medial compartment of the knee joint demonstrate the immediate effect of the laterally wedged insoles. The intervention may contribute to medial knee osteoarthritis rehabilitation.

1. Introduction

Knee pain and functional impairment are the most common complaints among patients with knee osteoarthritis (Brouwer et al., 2005). The nature of this loading could be altered by a number of conservative intervention strategies, such as foot orthoses, which are potentially capable of retarding the progression of knee osteoarthritis (Reeves and Bowling, 2011). The effects of foot orthoses on knee joint loading rely mainly on experimental measurements. Foot orthoses, such as laterally wedged insoles (LWIs), are assumed to shift the center of pressure (COP). This shift causes the ground reaction force (GRF) to pass closer to the knee joint center, consequently reducing the knee adduction moment, which are widely regarded as a surrogate index of medial knee compression (Yasuda and Sasaki, 1987).

Due to the experimental design diversity, subject individual differences, and relatively small changes introduced by the orthoses, consistent results have not been achieved (Abdallah and Radwan, 2011; Kakihana et al., 2005; Kerrigan et al., 2002; Maly et al., 2002). Currently, computer modeling, particularly finite element (FE) method gradually manifests its advantage of exploring the biomechanical responses of joint internal structures. Excessive stress on the cartilage layers and menisci predicted by FE model should be a direct index of knee loading. Reductions of the stress in any capacity would subsequently relieve the cumulative compressive loading on the knee joint. Thus, whether the LWIs reduce instant medial knee compartment loading could be deliberated through the FE analysis.

Our previous FE models of the foot and ankle (Cheung and Zhang, 2005, 2008; Cheung et al., 2005; Yu et al., 2008) have contributed to the understanding of the mechanical interaction between foot and foot supports. It has been shown that FE modeling could broaden our knowledge on foot biomechanics and improve foot support design with parametric information. Regarding the knee joint, FE modeling of the total knee joint or knee structures in clinical applications have also shown the potential of the FE method in investigating knee biomechanics with specified research interests (Beillas et al., 2007; Farrokhi et al., 2011; Peña et al., 2006; Ramaniraka et al., 2005; Shirazi and Shirazi-Adl, 2009). Though these works on modeling were encouraging, there is still a vacancy in FE studies of the lower extremity. To our knowledge, current FE models have not

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taken into account the lower extremity comprising knee joint, ankle joint and foot upon plantar loading.

The objective of our study was to develop a 3-dimensional FE model of the knee–ankle–foot complex, using detailed geometries of bony and soft structures together with orthosis support, to investigate the effects of LWIs on the internal loading distributions in the knee joint. An improved understanding of orthotic intervention through FE analysis can be helpful in knee pain prevention and rehabilitation.

2. Methods

The right lower extremity MR images of a normal male subject were scanned at 2 mm intervals from coronal plane in neutral unloaded position. The subject was 34 years old, with a height of 174 cm and a weight of 70 kg, free from any knee joint disease. The images were segmented in MIMICS v14 (Materialise, Leuven, Belgium) to reconstruct the geometry. The surface models of all the structures created from geometry reconstruction were converted to feature based solid models using Rapidform XOR3 (INUS Technology, Inc., Seoul, Korea). The generated solid models were then imported into ABAQUS v6.11 (Dassault Systems Simulia Corp., Providence, USA) for FE modeling and analysis.

The specific subject who volunteered the MR scan was then involved in the measurement of the GRF, COP, knee–ankle–foot position and plantar pressure. Vicon Motion System (Oxford Metrics, Oxford, UK), AMTI force platform (Advanced Mechanical Technology, Inc., Watertown, USA) and F-Scan (Tekscan, Inc., South Boston, USA) in-shoe pressure system were used in the experiments. The LWIs with wedge angles of 0, 5, and 10° were fabricated from high-density EVA material (A. Algeo Ltd., Liverpool, UK) with a shore hardness of 65. The insoles were trimmed to fit the subject's foot size and inserted into the shoes. Fig. 1a shows the subject standing on the pair of 5° LWIs. In the experiments, the subject went through several trials before data acquisition to get used with the foot supports and laboratory environments. The subject then performed three valid walking trials with each pair of LWIs.

A geometrical model of the LWI for each wedge angle condition was created in Rapidform XOR3 based on the shape of the insole used in experiment and then assembled with the knee–ankle–foot complex, separately (Fig. 1b). The walking position at single limb stance was simulated in a quasi-static manner. The alignment of the model from MR scan was modified to match the recorded kinematic location according to each wedge condition by altering knee, ankle and foot-ground angle in 3 anatomical planes. The ankle–foot structures were embedded in a volume of encapsulated foot soft tissue. An initial contact was first established between the orthosis and the plantar surface, with minimal contact pressure before loading. The Abaqus surface to surface contact relationship was assigned with a friction coefficient of 0.6 (Zhang and Mak, 1999) for the foot orthosis interface. GRF (Fig. 2c) was then applied at the location of COP underneath the ground support, which was allowed to translate and rotate in all degrees of freedom. As the boundary condition, the femur bone was cut approximately 10 cm above the femur condyles, and is fully constrained through local rigid body on distal femur (Fig. 2b). This setup allowed 6 degrees of freedom for the tibia.

The knee–ankle–foot complex, as shown in Fig. 2a, consisted of 30 bony segments, including the distal segment of the femur, patella, tibia, fibula, and 26 foot bones. The knee joint structures included the menisci, articular cartilages, and all the main ligaments of the knee joint, as shown in Fig. 2b. The cartilage-meniscus contact and interactions among bones were defined as surface to surface contact with frictionless tangential behavior. Foot and ankle ligaments and the plantar fascia were simulated as tension-only truss elements. The bony and other soft structures in the model were meshed with tetrahedral elements. The element size was determined to be approximately 1 mm in the knee ligaments, cartilages and menisci, 3 mm in bones, 5 mm in three insoles through convergence tests. The whole model contains approximately 155,615 nodes and 619,554 elements.

The material properties of the foot structures were assigned according to our previous FE model (Cheung et al., 2005). The encapsulated soft tissue and knee ligaments were defined as hyperelastic materials. The stress–strain data (Lemmon et al., 1997) on the plantar heel pad and polynomial expression (ABAQUS, 2011) were adopted for foot soft tissue assignment. The stress–strain relationships of knee ligaments were obtained from the model developed by Mesfar and Shirazi-Adl (2005). Strain-energy function of Mooney–Rivlin (ABAQUS, 2011) was employed for knee ligament assignment. All other tissues including bones were simplified as linearly elastic. The cartilage layers of the tibia, femur, and fibula were assigned a material property with Young's modulus of 12 MPa and Poisson's ratio of 0.45 (Moglo and Shirazi-Adl, 2003). The menisci were assigned with Young's modulus of 59 MPa and Poisson's ratio of 0.49 (LeRoux and Setton, 2002). The orthosis models were first created in Rapidform and then imported to Abaqus. They were meshed with 3-D brick elements and assigned material properties based on our previous work (Cheung and Zhang, 2008) using the EVA testing data.

The muscle forces were applied to the Abaqus connector elements (Fig. 2a), which modeled discrete physical connections. The insertion points of all the ligaments and muscles were determined according to MR images together with anatomy software (Interactive Series, Primal Picture Limited, London, UK) and tied to the bones. Muscle groups were chosen depending on their contributions in walking (Anderson and Pandy, 2001; Einhorn et al., 2000). Muscles playing minor roles around the knee joint and intrinsic muscles in the foot were ignored. The muscle forces during the simulated single-limb stance phase are tabulated in Table 1. The magnitudes of leg muscle forces were estimated from the physiological cross-sectional area of the muscles (Dul, 1983) and normalized EMG data during normal walking (Perry,

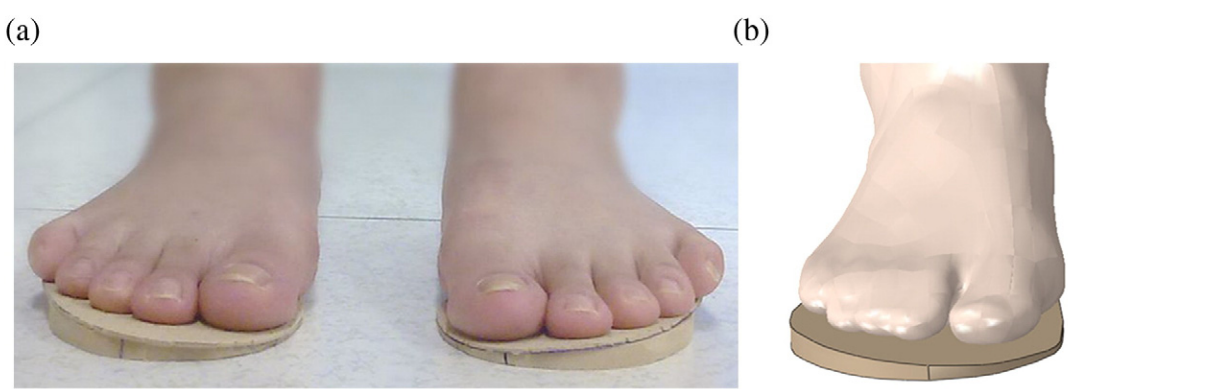

Fig. 1. The 5° LWIs investigated in this study. (a) The fabricated 5° LWIs used in motion analysis. (b) The developed 5° LWI in the FE model.