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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.

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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
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, 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.](image-url)
1992), assuming a linear EMG–force relationship (Kim et al., 2001). The magnitudes of quadriceps and hamstring muscle forces were obtained from literature (Anderson and Pandy, 2001).

To validate the model, the predicted plantar pressures in three LWI conditions were first compared with the F-Scan results measured from experiments. After that, the boundary conditions of the model were set up similar to a cadaveric experiment in the literature (Poh et al., 2011). An 1800 N compression force was loaded on the FE model in knee extension position as the literature prescribed to compare the model predicted knee contact pressures and areas with cadaveric data from the literature.

3. Results

Fig. 3 depicts the plantar pressure distribution obtained from F-Scan measurements and FE simulation of three LWI cases during the single-limb stance phase. The peak plantar pressure magnitudes were comparable between FE predictions and F-Scan results in three wedge angle conditions, respectively. The peak pressure tended to intensify at the forefoot pressure crest value region with increasing insole wedge angle. The contact areas predicted by the FE model were approximately 68, 49, and 43 cm² for 0, 5, and 10° conditions, compared with 70, 51, and 44 cm², respectively, from the F-Scan measurement.

The peak knee contact pressure and contact areas obtained from the FE analysis were compared with that in the cadaveric experiment from the literature (Poh et al., 2011), as shown in Table 2. The calculated contact pressures in the medial and lateral compartments were 12.62 and 8.97 MPa, respectively, compared with the mean values of 11.7 and 6.89 MPa, respectively, in the literature. The contact areas in the medial and lateral compartments were 315 and 363 mm², respectively, compared with the mean values of 374 and 403 mm², respectively, in the literature.

The predicted contact and distortion features between cartilages in the knee joint during the simulated single-limb stance phase were compared among three foot support conditions. Fig. 4 depicts the von Mises stress distribution patterns in the menisci and femoral cartilage with 0, 5, and 10° LWIs, separately. The peak pressures at the femur cartilage and menisci among the three wedge conditions were all predicted to occur in the medial side. A decrease in the peak von Mises stress at the medial femur cartilage region was revealed to be approximately 7.9% using a 5° LWI, and 12.7% using a 10° LWI.

Meanwhile, a decrease in the peak von Mises stress at the medial meniscus was revealed to be approximately 12.6% using a 5° LWI, and 15.4% using a 10° LWI.

Fig. 5 shows the effects of the wedge angle on the average von Mises stress in menisci (Fig. 5a) and the total contact force between the femur cartilage and meniscus (Fig. 5b). The trend in the stress reduction was more pronounced at the medial meniscus (15.4%) than that at the lateral meniscus (10.2%) with application of the 5° LWI, compared with the 0° LWI (Fig. 5a). The average von Mises stresses of both the medial meniscus and lateral meniscus with 5° wedge angle increments from the 5° LWI to 10° LWI were less sensitive than that with the same increments from the 0° LWI to 5° LWI.

<table>
<thead>
<tr>
<th>Muscles</th>
<th>Abbreviations</th>
<th>Force magnitudes (N)</th>
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<tbody>
<tr>
<td>Quadriceps</td>
<td>QUADS</td>
<td>200</td>
</tr>
<tr>
<td>Hamstring</td>
<td>HAM</td>
<td>80</td>
</tr>
<tr>
<td>Gastrocnemius</td>
<td>GAS</td>
<td>300</td>
</tr>
<tr>
<td>Soleus</td>
<td>SOL</td>
<td>400</td>
</tr>
<tr>
<td>Tibialis posterior</td>
<td>TP</td>
<td>70</td>
</tr>
<tr>
<td>Flexor hallucis longus</td>
<td>FHL</td>
<td>40</td>
</tr>
<tr>
<td>Flexor digitorum longus</td>
<td>FDL</td>
<td>30</td>
</tr>
<tr>
<td>Peroneus brevis</td>
<td>PB</td>
<td>30</td>
</tr>
<tr>
<td>Peroneus longus</td>
<td>PL</td>
<td>40</td>
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decrease in the average von Mises stress was found at the medial meniscus at approximately 19.2%, whereas an increase was found at approximately 4.5% at the lateral meniscus using the 10° LWI, compared with the 0° LWI (Fig. 5a). During single-limb stance, the medial meniscus sustained a higher total contact force than the lateral meniscus in all wedge angle conditions (Fig. 5b). At the medial meniscus, the total contact force decreased by approximately 16.7% using the 5° LWI, and by 26.2% using the 10° LWI. By contrast, the lateral meniscus experienced an increase in the total contact force by approximately 53.8% using the 5° LWI, and by 84.6% using the 10° LWI (Fig. 5b).

4. Discussion

In this study, we developed a 3-dimensional FE model of the human knee–ankle–foot complex assembled with the LWI. Wedge angles in 5° and 10° were chosen since inclinations ranging from 5°
to 10° were reported to prominently alter the knee adduction moment through experimental studies (Crenshaw et al., 2000; Hinman et al., 2012; Kerrigan et al., 2002). An angle around 10° was regarded as a high inclination which could lead to discomfort of the subject (Kerrigan et al., 2002). The 0° LWI was served as the control group. Our preliminary investigation focused on predicting the knee internal loading responses to the LWIs during specific phase of walking. Stress and pressure (surface stress) were predicted as direct parameters for loading responses.

Comparative tests were conducted before predictions to inspect the contact pressure and area of both the foot and the knee. The patterns of the predicted plantar pressure were comparable with the F-Scan measurements (Fig. 3). The magnitude of the peak plantar pressure in each condition measured by the F-Scan was smaller than that predicted by FE mainly due to differences in resolution between the F-Scan measurement and the FE prediction (Yu et al., 2008). The FE predicted peak knee contact pressures and knee contact areas in both medial and lateral compartments were in the standard deviation range of the cadaveric results in the literature (Poh et al., 2011). However, the predicted contact areas were less than the mean values in the cadaveric experiment (Table 2). Accordingly, the peak contact pressures were larger in a narrow area. These differences were reasonable due to the inconformity in the knee geometries between our subject and the cadaveric samples in the literature. Thus, the developed FE model was partially validated for successive stress predictions.

The model was then used to investigate the knee internal loading with the LWIs in three wedge angles. The predicted peak von Mises stresses in the medial condyle of the femur cartilage were greater than that in the lateral condyle in the three LWI conditions (Fig. 4). This phenomenon agreed with other FE studies (Shim et al., 2009) which investigated contact force distributions in the knee joint during gait. Nevertheless, the literature did not approach a quantitative analysis. It only provided information on the variation tendency of the contact force distributions. Again, there was a paucity of information in the knee joint internal loading during walking using FE analysis. The medial component dominance was also apparent in the peak von Mises stresses of the menisci. The peak von Mises stress occurred at the center region of the medial meniscus with a small leaning to the posterior region (Fig. 4), which indicated an internal rotation of the tibia during the single-limb stance.

| Table 2 |
|-------------------------|------------------------|
| **FE predicted peak contact pressure and contact area in the medial compartment and lateral compartment of the knee compared with the mean peak contact pressure and contact area (SD) obtained from the cadaveric experiment (Poh et al., 2011).** |
| **Medial compartment** | **Lateral compartment** |
| **Peak contact pressure (MPa)** | **Experiment** | **Mean 11.7 (SD 1.20)** | **Mean 6.89 (SD 3.47)** |
| **Contact area (mm²)** | **FE simulation** | **315** | **363** |
| **Experiment** | **Mean 374 (SD 87)** | **Mean 403 (SD 120)** |

![Fig. 4. von Mises stress in femur cartilage with (a) 0° LWI, (c) 5° LWI and (e) 10° LWI and von Mises stress in meniscus with (b) 0° LWI, (d) 5° LWI and (f) 10° LWI.](image-url)
The prediction showed that the percentage change in the total contact force on the lateral meniscus was larger than that on the medial meniscus (Fig. 5b), because the knee loading transmitted by the lateral compartment was much smaller than the medial compartment in the single-limb stance phase. The predicted trends of contact force were comparable with other computational model predictions. Shelburne et al. (2008) reported that a 5 mm displacement of COP led to a 0.1 body weight decrease in the peak medial contact force. As the FE model input, the lateral drift of COP from the 0° to 5° LWI was approximately 5 mm, and approximately 3.5 mm from the 5° to 10° LWI. In the FE results, the 5° LWI generated a 0.1 body weight decrease compared with the 0° LWI. The 10° LWI generated a 0.06 body weight decrease compared with the 5° LWI in the total contact force. The lateral drifts of plantar COP and medial movements of the knee joint center in wedge insole conditions observed from motion analysis implicated gait adaptation response which produced diminution of the GRF lever arm.

The developed model was built from a representative healthy subject with normal knee alignment and without knee pain. A systematic understanding about the biomechanical effects of orthotic intervention for knee OA requires research on the joint behavior, not only for patients, but also for healthy individuals. Several studies (Crenshaw et al., 2000; Fisher et al., 2007; Kakihana et al., 2004; Schmalz et al., 2006) have investigated the effects of foot orthoses on knee loading in healthy subjects using experimental methods. For instance, Fisher et al. (2007) indicated that altered angle shoe soles reduce the peak knee adduction moment in healthy subjects. These effects are better for subjects with greater peak knee adduction moments prior to the intervention. Despite the fact that the results between patients and healthy people could be diverse, the investigation could be launched in a less susceptible population for intervention. Regarding the intersubject variability, individual changes except geometrical diversity of structures could be easily added as parameters to the analysis in future modeling consideration.

The loading and boundary conditions together with muscle forces satisfied the equilibrium both in the knee joint and ankle joint in the FE solutions. It should be noticed that muscle force inputs among the three LWI conditions in the predictions were assumed consistent. It complied with the finding (Mills et al., 2012) that orthoses did not result in immediate changes to lower extremity EMG activities. However, divergences may exist for our practical cases and among subjects. In addition, the muscle force values in the present model were calculated from literatures. Individual EMG data and muscle forces predicted by musculoskeletal model should be synthesized for our specific subject as more realistic model input data.

It should be noticed that the model input data were obtained from experiments and literatures together through strict selection to approximate the authentic physical problem. Though the model was partially validated using contact pressure, similarly to a theory, it is in general impossible to prove the validity of a numerical model completely (Viceconti et al., 2005). Thus, it was demonstrated that the FE model could simulate the lateral wedged insole intervention to a suitable degree of accuracy.

Although the present study has yielded some preliminary findings in the single-limb stance phase, different walking phases with various joint positions and foot deformation will be taken into account in future study. Furthermore, subtalar joint and hip joint could be involved in the analysis to disclose the multi-joint effects of orthoses. The FE model we developed would be extended in validation and prediction of kinematic and kinetic responses of the joint structures including ligaments of the knee. In addition, better expression of the material properties should be pursued in the FE modeling to approach the authentic performance of the human tissue.

5. Conclusions

In conclusion, the FE predictions suggested that the LWIs could redistribute the knee internal loading by significantly relieving the stress and contact force at the medial compartment of the knee. Using the LWIs as intervention may prevent the onset and progression of the medial knee osteoarthritis. With further improvement of the FE model, together with experimental studies will provide a useful platform for understanding the biomechanical effects of foot orthoses. With an implementation of parametric analyses, we could alter any features of the orthoses through the FE simulation, for example various material and structural characteristics, to examine the joint biomechanical responses. Thus, this model will also provide scientific fundamentals for optimal orthoses design.

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