FINITE ELEMENT EVALUATION OF THE MECHANICAL BEHAVIOUR OF A DETAILED FOOT/FOOTWEAR MODEL

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ABSTRACT
This work evaluates the mechanical behaviour of a foot bio-model using three-dimensional finite element analysis. Initially, CT scans and reverse engineering approaches are used to reconstruct detailed foot models in order to be able to achieve quality meshing and realistic results in the FE analysis. Following, two FE models are developed in order to examine the effect of modelling details and materials on the mechanical behaviour of the foot. Comparison between the two models with respect to stiffness and plantar stress distribution are made and the differences between the results for the various material models are discussed.

Keywords: biomechanics, foot bio-model, finite element analysis.

INTRODUCTION
During the last fifteen years, several researchers have proposed 3D foot models generated by CT or MRI scans for FE analysis. A preliminary 3D finite element model of a foot was developed by (Chen, 2001) where the metatarsals and tarsal bones were modeled with two rigid columns (medial and lateral) without segmenting each of the individual bone in order to reduce the complexity of the produced model. Similar simplifications of the foot model can be found in (Gefen, 2003). In (Camacho, 2002), a detailed model of the human foot and ankle, incorporating realistic geometrical properties of both bony and soft tissue components was proposed. Detailed models of the human foot have also been proposed in (Cheung, 2005). Another detailed FE model of a real human foot was developed in (Hsu, 2008) in order to examine the role of shoe insole shape in lowering plantar fascia pressures. A more realistic foot model was developed in (Chen, 2011) taking data from CT scans. A detailed FE model was developed consisting of 400000 three-dimensional elements for the bones, soft tissue and Achilles tendon. All the other parts were modeled with link elements. A fully parametric study revealed the effect of soft tissue material and insole material and shape on the normal and shear stress distribution. The results were supported by extended experimental data.

Although the evolution in the 3D modeling of human foot and the variety of the published works, there still exist fundamental open problems concerning the accuracy of the underlying 3D foot model. Efforts to reduce the complexity of the geometry in the 3D model lead to inaccurate representations that affect the estimations of finite element methods. The non-parametric form of the foot models results in a time-consuming process that reconstructs each 3D model from scratch. Despite the efforts for a detailed description of the human foot, current approaches do not allow high accuracy in the representation of the anatomic elements. Furthermore, all the missing parts of the human foot that occur during segmentation process are reconstructed by utilizing conventional CAD systems to non-standard mechanical shapes.
FOOT RECONSTRUCTION

The human foot consists of 26 bones, 33 joints and various muscles, ligaments and tendons along with soft tissues like vessels, nerves, and the skin. The construction of human soft and hard tissue bio-models is considered in a reverse engineering framework as a reconstruction process that is based on non-invasive imaging techniques. Computed Tomography (CT), Magnetic Resonance Imaging (MRI) and Ultrasonography are frequently used for this purpose since they provide structural information, physical properties and geometric data of human body through a series of tomographic images.

In this work, a complete reconstruction of foot bones from dense CT scans using two different approaches was used. The first approach is based on the commercial software MIMICS. The second follows a Reverse Engineering (RE) methodology where a point-cloud is extracted from CT scans for each bone. Then, basic RE techniques are applied to reconstruct the bone surface by using the commercial software Geomagic and the Poisson surface reconstruction method. The details of the reconstruction process can be found in an earlier work of the authors (Koutkalaki, 2014).

Foot data used in this research is based on a set of CT scans taken on a foot of a healthy – male subject with a resolution of 0.5mm. The reconstruction process provided two models to be used in the FE analyses. The first was taken using the MIMICS software and consisted of the soft tissue. The same software was used to discretize the model with three different mesh densities and then imported into the ANSYS commercial software. The three meshes, identified as Mesh 1, 2 and 3 consisted of 21000, 12000 and 8500 tetrahedral elements and are shown in Fig. 1.

The second model consisted of detailed reconstruction of the bone structure and was taken using the commercial software Geomagic and the Poisson surface reconstruction method (Fig. 2). Each bone geometry was imported into the ANSYS software, where a volume was created and meshing was achieved using tetrahedral elements.

![Fig. 1 - The three mesh densities from MIMICS software used in the analyses of the first FE model: from left to right, Mesh 1, Mesh 2 and Mesh 3](image)

FINITE ELEMENT MODELS

The first FE model consisted of the soft tissue, which was assumed to rest on a support pad, as shown in Fig. 3. Because of the large difference of stiffness between the bones and soft tissue, the bone structure was assumed rigid and the cavities in the soft tissue model were fixed. Two materials were considered for the support pad, a hard floor with Young's modulus of 17 GPa and a PU foam with Young's modulus of 15 MPa. Linear behaviour was assumed for both
materials. Concerning the material of the soft tissue, three different hyperelastic materials were considered, identified as F2, F3 and F5 and shown in Fig. 4. The 5 parameters Mooney-Rivlin hyperelastic material model of the ANSYS software was used with the parameters taken from the literature and shown in Fig. 4. For comparison, a linear material with Young's modulus of 1.15 MPa was also considered for the analyses. Contact elements were developed between the soft tissue and the support pad and the model was loaded with a step-wise normal displacement of the lower part of the pad. The reaction force, evaluated from the solution, was considered to be the force applied to the foot.

![Fig. 2 - Complete bio-model of the foot bones structure used in the second FE model](image)

![Fig. 3 - The foot model on the support pad](image)

![Fig. 4 - Stress-strain curves of soft tissue and corresponding coefficients of the hyperelastic material models (Cheung, 2005)](image)

<table>
<thead>
<tr>
<th>Coefficients</th>
<th>Normal</th>
<th>F2</th>
<th>F3</th>
<th>F5</th>
</tr>
</thead>
<tbody>
<tr>
<td>$C_{10}$</td>
<td>0.06556</td>
<td>0.17113</td>
<td>0.25669</td>
<td>0.42762</td>
</tr>
<tr>
<td>$C_{01}$</td>
<td>-0.05841</td>
<td>-0.11683</td>
<td>-0.17524</td>
<td>-0.29207</td>
</tr>
<tr>
<td>$C_{20}$</td>
<td>0.03900</td>
<td>0.07800</td>
<td>0.11700</td>
<td>0.19499</td>
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<tr>
<td>$C_{31}$</td>
<td>-0.02319</td>
<td>-0.04039</td>
<td>-0.06957</td>
<td>-0.11994</td>
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<tr>
<td>$C_{52}$</td>
<td>0.00531</td>
<td>0.01702</td>
<td>0.02335</td>
<td>0.04226</td>
</tr>
<tr>
<td>$D_1$</td>
<td>3.65273</td>
<td>1.23235</td>
<td>1.21758</td>
<td>0.73055</td>
</tr>
<tr>
<td>$D_2$</td>
<td>0.00000</td>
<td>0.00000</td>
<td>0.00000</td>
<td>0.00000</td>
</tr>
</tbody>
</table>
The second FE model consisted of the bone structure of the foot as taken from the reconstruction process (Fig. 5). The model also included all the ligaments, tendons (including Achilles tendon) and the plantar fascia. All bones were modeled using tetrahedral elements (SOLID285) and a total of 365000 elements were created. All the ligaments were modeled using a simple tension-only link element (LINK180) as their purpose is to transfer the force between the bones. Concerning the Achilles tendon and the plantar fascia, they were modeled with shell elements (SHELL181) because of the larger bond surface with the bone structure and in order to have more realistic load transfer. A total of 100 elements, simulating the soft tissues, were modeled. The element types and properties used are presented in Table 1.

The model was assumed to rest on a two-part pad. The upper part is considered to simulate the soft tissue and was assigned the same properties as for the first model. The lower part was assigned the properties of a hard floor or a PU foam. Contact element were developed between the three parts of the model and between neighbouring bones. A force of 250 N was applied to the Achilles tendon, an average value considered in the literature whereas the upper part of the bone structure was fixed, as shown in Fig. 5. The model was loaded with a step-wise normal displacement of the lower part of the pad and the reaction force, evaluated from the solution, was considered to be the force applied to the foot.

![Fig. 5 - Complete FE model](image)

<table>
<thead>
<tr>
<th>Component</th>
<th>Element type</th>
<th>Young's modulus (MPa)</th>
<th>Poisson ratio</th>
<th>Cross section area (mm²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bones</td>
<td>Solid 285</td>
<td>7300</td>
<td>0.3</td>
<td>-</td>
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<tr>
<td>Ligaments</td>
<td>Link 180</td>
<td>260</td>
<td>-</td>
<td>18.4</td>
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<tr>
<td>Plantar fascia</td>
<td>Shell 181</td>
<td>350</td>
<td>-</td>
<td>290.7</td>
</tr>
<tr>
<td>Tendons</td>
<td>Link 180</td>
<td>1200</td>
<td>-</td>
<td>58.6</td>
</tr>
<tr>
<td>Achilles tendon</td>
<td>Shell 181</td>
<td>1200</td>
<td>-</td>
<td>58.6</td>
</tr>
</tbody>
</table>

Table 1 - Element types and material properties
RESULTS

The results recorded from the post-processor of the ANSYS software were mainly the applied displacement, the force and the plantar pressure distribution. The first task was to check for convergence of the solution with regards to mesh density. Typical plantar pressure distributions (for the F2/floor model) are presented in Fig. 6 for the three meshes used (values in MPa). We notice a significant difference in the maximum pressure between the Mesh 3 and Mesh 2 (25%) and a very smaller difference between Mesh 2 and Mesh 1 (12%). However, the contact areas look very similar for all three meshes and the difference in the maximum pressure is clearly due to the local contact behaviour, which depends highly on element size. It should be noted that the results of the maximum plantar pressures are very close to the ones reported in the literature, both numerically and experimentally.

The force-displacement curves evaluated for the three meshes are almost identical, as shown in Fig. 7. This suggests that all three meshes evaluate accurately the global behaviour of the model. The effect of the material model used for the soft tissue is shown in Fig. 7 (for Mesh 2). We notice that the whole model is behaving in a "hyperelastic" fashion. Examining all the results reveals that this is not only due to the non-linear behaviour of the material but also due to the non-linear behaviour of the contact, since larger forces leads to larger contact area and higher stiffness.

![Fig. 6 - Distribution of plantar pressure (minimum principal stress) for Mesh 1, 2 and 3](image)

![Fig. 7 - Effect of mesh density and material type on force-displacement curves](image)
Typical distribution of the plantar pressure evaluated using the second FE model are presented in Fig. 8. Due to the shape of the soft tissue, the contact area predicted is smaller than that of the first model, however, the differences in the maximum plantar pressure values are very small. In Fig. 9, the force-displacement curves of three material sets are presented. The sets correspond to the materials of the two pads, where "rigid" refers to the floor and "hyper" to the hyperelastic material F2. The results reveal that the use of the hyperelastic material produces the lower stiffness, whereas the higher stiffness is observed when considering linear behaviour of the soft tissue on a hard floor. However, the values of the maximum plantar pressures (Fig. 9) are very similar ranging from 0.46 MPa for the linear/foam case to 0.56 for the linear/rigid case and for a force of 0.5 kN (50 kg).

If we compare the results between Fig. 7 and Fig. 9, we may assume that there is a considerable difference between the force-displacement curves between the two FE models. However, careful examination of the results shows that this is not the case. The comparison of the predicted stiffness for the two FE models should be made by taking into account the significant difference in modelling the soft tissue. Because of the fitting between the bones and soft tissue and the flatness of the soft tissue in the second model we can have differences in the displacements. If we modify the displacement results of the first model, and consider the same displacement for a force of 0.5 kN, then the modified curves are shown in Fig. 10. We see almost identical stiffness between the two models for the three material sets.
examined. This could suggest that we can use a simplified model to evaluate the foot/footwear stiffness, but this requires further study for all possible material models and also for the differences in plantar pressures observed between the two models.

CONCLUSIONS

This study demonstrated the effectiveness of the reconstruction process to produce models appropriate for FE analysis. The results of the analyses revealed that both FE models produced reliable results as compared to the literature. The research will be further extended to account for more realistic foot/footwear interactions with the goal to provide useful information for the design of footwear.

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