Inter-patient evaluation of stresses in proximal implanted tibiae

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Introduction

In biomechanics, finite element analysis (FEA) is still only a comparative tool. The predictive capabilities of FEA can only be evaluated by performing prospective clinical trials in conjunction with patient specific FE modelling. Previous 3D finite element studies of implanted tibiae (Taylor et al., 1998) have modelled a single tibia and have assumed that the results are generic and applicable to the whole patient population. To the authors’ knowledge, no study has examined multiple tibiae or included patient specific data, such as body weight, bone mechanical properties and component alignment. Only by constructing finite element models taking into account these parameters in combination with prospective clinical studies can the predictive power of FEA be assessed. The purpose of this study was to evaluate the differences in the predicted stresses and risk ratios observed on the resected surface of models of proximal implanted tibiae created from patient specific data.

Materials and Methods

Finite element models of four proximal implanted tibiae were analysed. The models were created from quantitative computed tomography (QCT) data of patients operated in University Hospital-Lüden. In all patients, the type of prosthesis implanted was a cementless, hydroxyapatite (HA) coated Howmedica-Crucifix knee replacement. The immediate post-operative situation was modelled by assuming frictionless contact between the tibia and the tibial plateau. Additionally, post-operative x-ray images in both the sagittal and frontal planes were used to determine the vertical tibial plateau position and alignment relative to the tibia in each patient.

Meshes of linear tetrahedral elements were created in I-DEAS for both bone and implant. Based on a previous mesh convergence study, it was concluded that an element edge length of 1.4 mm should be used on the contact surfaces of the tibia for all the models. The mesh of the implant was created in a way that it perfectly matched that of the bone, as it was noticed that this configuration produced more stable results. A program called Bonemat, developed at the Laboratorio di Tecnologia Medica – Istituti Ortopedici Rizzoli (Bologna-Italy), was used to assign the material properties on an element-by-element basis (Zannoni et al., 1998). This process is based on the correlation between QCT data and the material properties of the bone (McBroom et al., 1985; Carter et al., 1977; Linde et al., 1991). A linear correlation was defined (using calibration phantoms present in the CT images) to convert image intensity values in Hounsfield Units (HU) to apparent density (\(\rho\)) in g/cm\(^3\) (Equation 1). Since this equation did not work properly for small or negative values of HU, another linear equation was defined for this range, avoiding in this way negative values of bone apparent density.

\[
\begin{align*}
\rho &= 1.037 \cdot 10^{-4} \cdot (HU + 1000) \quad \text{for} \quad -1000 \leq HU \leq 138 \\
\rho &= 1.011 \cdot 10^{-3} \cdot (HU - 2000) + 2 \quad \text{for} \quad HU > 138
\end{align*}
\]  

(1)

In order to convert apparent density values (\(\rho\)) to material Young’s modulus (\(E\)), the equation reported by Carter and Hayes (1977) was found to be the most suitable for cortical bone. For trabecular bone, the correlation published by Linde et al (1991) was considered more accurate, as it was generated from experimental tests performed on specimens obtained from proximal tibiae. Therefore, two different equations were used depending on the bone apparent density values (Equation 2).

\[
\begin{align*}
E &= 2003 \cdot \rho^{1.56} \quad \text{for} \quad 0 \leq \rho \leq 0.778 \quad (\text{Linde et al., 1991}) \\
E &= 2875 \cdot \rho^{3.0} \quad \text{for} \quad \rho > 0.778 \quad (\text{Carter et al., 1977})
\end{align*}
\]  

(2)
The distal end of the tibia was rigidly constrained. The loads used in the models (Table 1) were equivalent to three times the weight of each patient (Morrison, 1970). A bi-condylar load case was used, in which 60% of the total force was applied on the medial side and 40% on the lateral side. The tibial polyethylene insert was not included in order to simplify the analysis; therefore, the forces were applied directly on the tibial plateau (figure 1).

Risk ratio values (defined as the von Mises stress divided by the ultimate compressive strength) in a 2-mm thick layer of the proximal tibia were examined and compared between all four models (figure 1). A value greater than 100% indicates that failure of the supporting bone is likely to occur.

All the analyses were carried out using MARC 2000.

<table>
<thead>
<tr>
<th>Patient No.</th>
<th>Weight (kg)</th>
<th>Medial Load (N)</th>
<th>Lateral Load (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>135</td>
<td>2381</td>
<td>1588</td>
</tr>
<tr>
<td>2</td>
<td>115</td>
<td>2029</td>
<td>1352</td>
</tr>
<tr>
<td>3</td>
<td>89</td>
<td>1570</td>
<td>1047</td>
</tr>
<tr>
<td>4</td>
<td>74</td>
<td>1305</td>
<td>870</td>
</tr>
</tbody>
</table>

Table 1: Weights and loads used for each patient.

Results & Discussion

The risk ratio distributions on the resected surface for the four tibiae examined are shown in figure 2. In all four models, a noticeable difference in the overall risk ratio distribution on the contact surface was observed. Table 2 shows the peak and mean values of Young’s modulus, von Mises stress and risk ratio for all patients.

In general, the peak values were found in the portion of cancellous bone supporting the posterior and frontal-lateral sides of the tibial plateau and (Figure 1). It was observed that, in some cases, a large portion of the resected surface of the tibia presented risk ratios greater than 100% (e.g. patient 2), whereas in patient 3, the percentage of volume with risk values exceeding 100% is significantly smaller (5.7%). This clearly shows how important it is to use patient specific data (in this case, mechanical properties of the bone).

The purpose of this study was to assess the risk of bone failure with four patient specific models. For all four tibiae, there were areas of bone, where the localised risk of failure was high. However, in general, the risk of failure was below 100% over the majority of the resected surface.

<table>
<thead>
<tr>
<th>Patient Number</th>
<th>Modulus (MPa)</th>
<th>Von Mises Stress (MPa)</th>
<th>Risk Ratio (%)</th>
<th>Risk&gt;100% (% Vol)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>mean</td>
<td>peak</td>
<td>mean</td>
<td>peak</td>
</tr>
<tr>
<td>1</td>
<td>267.8</td>
<td>12850</td>
<td>1.187</td>
<td>61.2</td>
</tr>
<tr>
<td>2</td>
<td>212.6</td>
<td>6655</td>
<td>1.237</td>
<td>25.9</td>
</tr>
<tr>
<td>3</td>
<td>241.7</td>
<td>10090</td>
<td>0.741</td>
<td>17.3</td>
</tr>
<tr>
<td>4</td>
<td>141.7</td>
<td>11640</td>
<td>0.623</td>
<td>18.1</td>
</tr>
</tbody>
</table>

Table 2: Peak and mean values of Young’s modulus, von Mises stress and risk ratio for the 2-mm layer of elements examined in each patient.
Roentgen stereophotogrammetric analysis (RSA) studies of tibial plateaus have shown that during the first six months after surgery, there is a period of rapid migration. This is widely thought to be due to the implant “bedding in” (Ryd et al., 1995; Nilsson et al., 1996). The findings of this study support this view, as some regions presented considerably high risk ratios. In these regions, the cancellous bone will be crushed and the load redistributed on the resected surface until an equilibrium position is reached. This study has shown that patient specific FE modelling does reveal significant differences in the predicted stress and risk ratio distributions. This emphasises the importance of moving away from the traditional generic modelling approach to patient specific modelling.

References


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