Analysis of an Early Intervention Tibial Component for Medial Osteoarthritis

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Introduction: Tibial component loosening is an important failure mode in unicompartmental knee arthroplasty (UKA) which in some cases may be due to the 6-8 mm of bone resection required. A 6-8 mm resection removes the stiffest bone near the tibial surface. To address component loosening and fixation, a new Early Intervention (EI) design is proposed which consists of a plastic inlay for the distal femur and a thin metal plate for the proximal tibia. With this reversed materials scheme, the EI design requires minimal tibial bone resection compared to traditional UKA. This study investigated, by means of finite element (FE) simulations, the advantages of a thin metal tibial component compared with traditional UKA tibial components, such as an all-plastic inlay or a metal-backed onlay. We hypothesized that an EI component would produce comparable stress, strain, and strain energy density characteristics to an intact knee and more favorable values than current UKA components.

Methods: From a series of UKA cases, a typical preoperative CT scan of a knee joint was obtained from the left knee of a 60-year-old male patient who was later treated with UKA. A 3D bone model was generated from the CT scan using Mimics 16.0 software (Materialise, Leuven, Belgium).
Four tibial components were analyzed in this study (Figure1).
Two design variations of the EI tibial component were modeled (EI 2 mm thick and EI 3 mm thick). EI components were placed at a 2 mm resection depth below the bone surface and modeled in a cementless and a cemented mode (1 mm thick cement mantle). An all-plastic UKA inlay of 6 mm thickness was positioned at a 4 mm resection depth. A metal-backed UKA onlay consisting of a 6 mm plastic bearing and a 2 mm metal base was placed at a 6 mm resection depth. Both UKA models included a 1 mm cement layer. The cement layer was bonded to the tibial components and underlying bone surface. Cementless EI components were modeled with bonded contact between the metal component and supporting bone. Parts were assigned material properties listed in Table 1. 10-node tetrahedral elements of 2 mm were used in this study. Models were analyzed in Abaqus v6.12 (Dassault Systems, Providence RI). On the medial side, a downward force of 1,500 N was applied in the direction of the long axis of the tibia over an area of 40 mm². On the lateral side, a downward force of 750 N was applied over an area of 450 mm². An upward force of 1,500 N was applied over the tibial tubercle to simulate the force exerted by the patella tendon.
In all models, the base of the tibia was fully constrained.
The intact tibia bone was also modeled. The same axial forces were applied, however, the contact area on the medial side was 554 mm² to represent loading in an intact knee.
Stresses at the bone interface can be used to assess load distribution while the interface strains indicate possible loosening and bone failure. SED below the components can be used as an indication of bone remodeling and/or potential pain.

Results: Figure 2 compares maximum von Mises stresses at the bone interface between EI components, UKA components and an intact model. In the bar plot, stress results for an EI 3 mm and the metal-backed onlay were similar to an intact tibial model. Stresses for both the EI 2 mm and all-plastic inlay components were greater than an intact model. Figure 3 compares strains between tibial components and an intact tibia model. Strain values corresponding to an EI 2 mm component, EI 3 mm component, and metal-backed onlay were less than strain values calculated for an intact model (Figure 3). Conversely, strains for an all-plastic inlay were greater than the intact model and EI components. The strain values calculated for all models did not exceed the estimated yield strain of natural bone [7].
The cancellous bone beneath the surface in the central region of the medial condyle (Figure 4) was the region of interest when analyzing the strain energy density in both intact and implanted models. Compared to an intact model, the SED in an EI 2 mm with cement, EI 3 mm, EI 3 mm with cement, and a metal-backed UKA decreased by 18%, 53%, 57% and 41%, respectively.
Conversely, the SED in an EI 2 mm and an all-plastic inlay compared to an intact tibia increased by 53% and 300%, respectively.

Discussion: Indeed, FE results showed that an EI design reduced stresses, strains and strain energy density in the supporting bone compared to an all-plastic UKA component. Analyzed parameters were similar for an EI and a metal-backed onlay, but the EI component had the advantage of minimal resection of the stiffest bone.
The all-plastic inlay showed elevated stresses, strains, and SED compared to an intact reference model, EI 2 mm, EI 3 mm, and a
metal-backed onlay. For the inlay, the elevated SED, stress, and strain may be indicators of loosening, bone remodeling, and pain reported clinically caused in part by the low modulus of the plastic and its thickness. The bone strength and density has been found to diminish with depth below the surface, in both normal and osteoarthritic cases. An EI 3 mm and metal-backed onlay produced comparable stresses and strains; however, an EI 3 mm component only requires a 2 mm resection, whereas a metal-backed onlay requires a 6-8 mm resection. With less resection, the stronger and denser bone would be preserved providing stronger fixation of the component and ease of a future revision surgery since more bone stock is available.

**Significance:** The EI design will require small incisions and bone resections, which will allow for natural anatomy of the patient to be preserved and less invasive surgery techniques to be used. This study showed that an EI design has the potential to reduce the loosening rates seen in contemporary UKA designs.

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**References:**

**Figure 1.** Four components analyzed: EI 2 mm, EI 3 mm, all-plastic inlay, metal-backed onlay. Change in bone density and elastic modulus with resection depths ranging from 2 mm to 8 mm.
<table>
<thead>
<tr>
<th>Part</th>
<th>Material</th>
<th>Young's Modulus (MPa)</th>
<th>Poisson’s Ratio</th>
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<tr>
<td>Metal</td>
<td>CoCrMo alloy</td>
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<td>0.30(^1)</td>
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<td>634(^2)</td>
<td>0.45(^3)</td>
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<tr>
<td>Cement</td>
<td>PMMA</td>
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<td>0.40(^5)</td>
</tr>
<tr>
<td>Bone</td>
<td>Based on equations by Rho et al(^6)</td>
<td>0.2</td>
<td></td>
</tr>
</tbody>
</table>

Table 1. Material properties assigned to parts modeled

Figure 2. Maximum von Mises stresses at the surface of the resected bone. The horizontal lines represent stresses in an intact tibia corresponding to each component’s respective resection depth (2 mm, 4 mm, and 6 mm).
Figure 3. Maximum and minimum principal strains at the surface of the resected bone. The horizontal lines represent strains in an intact tibia corresponding to each component’s respective resection depth (2 mm, 4 mm, and 6 mm).
Figure 4. Strain energy densities beneath the components near the resected surface compared to an intact model.