Prosthesis design and stress profile after hip resurfacing: a finite element analysis

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ABSTRACT

Purpose. To evaluate the effect of prosthesis design on stress profile in the proximal femur after hip resurfacing.

Methods. The von Mises stress profile of the native femur was simulated and compared with that of resurfaced femurs using various prosthetic materials (titanium, cobalt-chrome, ceramic), stem lengths (normal, half, short, and no stem), and femoral head coverage (shell size) [260º, 220º, 180º, and 140º].

Results. Hip resurfacing altered the stress profile of the cancellous (but not cortical) bone of the femoral neck. Maximal cortical stresses were observed at the posterior half of the medial femoral neck. The stress profile of the native femur was most similar to that of the resurfaced femur made of titanium, with a short or no stem and 260 degrees of femoral head coverage (shell size).

Conclusion. Optimising prosthesis design by minimising biomechanical alterations seems a valid approach to achieving favourable long-term outcomes. Cadaveric and in vivo studies are needed to confirm the clinical relevance and feasibility.

Key words: arthroplasty; biomechanics; finite element analysis; hip; prosthesis design

INTRODUCTION

Resurfacing of the femoral head is the first reported technique for hip arthroplasty.1 It conserves bone stock and the biomechanics of the proximal femur and facilitates revision total hip arthroplasty (THA), but long-term results are unpredictable,2-4 largely owing to polyethylene particulate wear–induced osteolysis. THA subsequently became the treatment of choice for advanced degenerative hip arthritis. Hip resurfacing has regained popularity owing to the availability of low-wear metal-on-metal articulations. It is not known to what extent biomechanical disadvantages and prosthesis design have attributed to the high failure rates.5-10 We evaluated the effect of prosthesis design on stress profile in the proximal femur after hip resurfacing, using a finite element analysis.

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MATERIALS AND METHODS

The stress profile of the proximal femur before and after hip resurfacing was analysed using a finite element model. A 3-dimensional proximal 3/4 of the ‘standardised femur solid model’ was obtained from the Biomechanics European Laboratory (Fig. 1).11

A generic hip resurfacing prosthesis design was simulated. It was hemispherical with an outer diameter of 50 mm and a uniform thickness of 2 mm. Three different prosthetic materials were simulated: titanium (elastic modulus=110 GPa), cobalt-chrome (elastic modulus=200 GPa), and ceramic (elastic modulus=350 GPa). The stem diameter was 3.8 mm and was aligned with the femoral neck axis, with 3 different lengths. The normal stem (most commonly used) extended to the intertrochanteric plane (i.e. the distal end of the femoral neck). The half stem extended to the transition between the femoral neck and head. The short stem extended to the centre of the femoral head. The femoral head coverage (shell size) was simulated as 260º, 220º, 180º, and 140º (Fig. 2).

A homogeneous cement layer with uniform thickness of 1 mm was simulated between the reamed femoral head and the prosthesis, but not around the stem. The interface between the stem and the surrounding bone within the femoral head and neck was simulated using a frictional surface contact algorithm, replicating the surgical procedure of an uncemented femoral stem. Owing to the relative motion between the stem and bone, a small sliding formulation was prescribed with a friction coefficient of 0.1 to normalise the primary contact force transmission to the implant surface. Because of differences in materials, the prosthesis surface was defined as the master and the surrounding bone as the slave. The bone surface deformation was dependent on the displacement of the implant surface.

Material properties of the cortical bone, cancellous bone, bone cement, and implant were assumed to be homogeneous, isotropic, and linear elastic (Table). To compare various permutations of designs, one loading condition was simulated. The resultant hip joint contact force was 2.8 times an assumed body weight of 72 kg (resultant force, 1978 N), and directed through the centre of the femoral head oriented 15º laterally in the frontal plane and 30º posteriorly in the sagittal plane. This load represented the range through which the hip contact force varied during normal to fast walking.12 The hip contact loading profile was parabolically shaped over an ellipsoidal load distribution of 49 nodes with the major axis in the anteroposterior direction (Fig. 1).13 Abductor muscle force was simulated as 1.285 times of the body weight (resultant force, 907 N), representing the magnitude of the resultant force of the gluteus medius and

<table>
<thead>
<tr>
<th>Material</th>
<th>Elastic modulus (GPa)</th>
<th>Poisson ratio</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cortical bone</td>
<td>17</td>
<td>0.3</td>
</tr>
<tr>
<td>Cancellous bone</td>
<td>0.7</td>
<td>0.3</td>
</tr>
<tr>
<td>Polymethyl-methacrylate</td>
<td>2</td>
<td>0.3</td>
</tr>
<tr>
<td>Titanium</td>
<td>110</td>
<td>0.3</td>
</tr>
<tr>
<td>Cobalt-chrome-molybdenum</td>
<td>200</td>
<td>0.3</td>
</tr>
<tr>
<td>Ceramic</td>
<td>350</td>
<td>0.3</td>
</tr>
</tbody>
</table>
The abductor muscle force was oriented in the frontal plane at an angle of 68° with the transverse plane. The load was applied over 9 nodes according to the insertion sites, 6 for the gluteus medius and 3 for the gluteus minimus. The most distal elements of the femoral shaft were constrained to zero displacement, simulating a distally potted femur, which is the standard set-up for *in vitro* testing.

The finite element model was developed using the Patran software (MSC Software, Santa Ana, CA, USA). The mesh consisted of 4838, mainly 8 node hexahedron elements and 5168 nodes. Finite element simulations were performed using the Abaqus software (Hibbit, Karlsson & Sorensen, Pawtucket, RI, USA). Von Mises stresses (a combination of compressive, tensile, and shear stresses) of various simulations were compared. The femoral neck was analysed at 10 transverse cross-sections perpendicular to its longitudinal axis. The von Mises stresses along the medial and lateral cancellous bone in the femoral neck were simulated (Fig. 3). The femoral head was analysed using frontal and sagittal cross-sections through the model.

The stress profile of the native femur was compared with that of resurfaced femurs using various prosthetic materials (titanium, cobalt-chrome, ceramic), stem lengths (normal, half, short, and no stem), and femoral head coverage (shell size) [260°, 220°, 180°, and 140°]. To reduce permutations, parametric tests for prosthetic materials were simulated with a normal stem and 260° of femoral head coverage, whereas those for stem length were simulated with a cobalt-chrome prosthesis and 260° of femoral head coverage and those for femoral head coverage (shell size) were simulated with a cobalt-chrome prosthesis with a normal stem.

**RESULTS**

**Validation**

The model was validated by comparing principal strain and stress values at the cortex of the femoral neck under identical loading conditions and normalised for the applied loading force ($\varepsilon/F$ and $\sigma/F$, respectively) using previous results from both analytical and experimental measurements. The normalised principal strain of 723 με/kN (range, 343–1800 με/kN) and the normalised principal stress of -1.3 cm$^{-2}$ at the medial neck and of +0.7 cm$^{-2}$ at the lateral neck (ranges, -0.3 to -2.2 and 0.1 to 2 cm$^{-2}$, respectively) were within the reported ranges. Hip resurfacing did not alter the stress profile of the cortical bone of the femoral neck, regardless of prosthesis design. Maximal cortical stresses were observed at the posterior half of the medial femoral neck, both before and after hip resurfacing. In contrast, hip resurfacing altered the stress profile of the cancellous bone of the femoral neck; the changes on the medial side were of greater magnitude than those on the lateral side (Figs. 4 and 5). In addition, there was stress concentration at the tip of the stem and stress shielding of the entire femoral head.

**Prosthetic material**

Compared to the native femur, there was stress concentration along the entire medial femoral neck and distal half of the lateral femoral neck, as well as stress shielding along the proximal half of the lateral femoral neck after hip resurfacing (Fig. 4a). The stiffer the prosthetic materials (ceramic>cobalt-chrome>titanium), the greater was the stress concentration and shielding of the femoral head and neck (Fig. 5a).

**Stem length**

Compared to the native femur, there was stress concentration along the entire medial femoral neck when a normal stem was used. It was along the proximal half of the medial femoral neck when a
A half stem was used. No stress concentration was observed when a short or no stem was used (Fig. 4b). Stress shielding along the proximal part of the lateral femoral neck was observed when a normal or half stem was used, but not with a short or no stem. Stress concentration along the distal half of the lateral femoral neck was observed when a normal or half stem was used, but not with a short or no stem.
femoral neck was only noted when a normal stem was used. Stem length did not greatly influence the stress shielding of the femoral head (Fig. 5b).

**Femoral head coverage (shell size)**

Compared to the native femur, after hip resurfacing there was stress concentration along the entire medial distal half of the lateral femoral neck, as well as stress shielding along the proximal half of the lateral femoral neck (Fig. 4c). The larger the femoral head coverage (shell size) [260°>220°>180°>140°], the greater was the stress concentration and shielding of the femoral head and neck (Fig. 5c).

**DISCUSSION**

The more physiological the stress profile after hip resurfacing, the better the performance of the prosthesis in terms of function and survival. Prosthesis design affects the stress profile of the proximal femur. A prosthesis made of titanium, with a short or no stem and 260° of femoral head coverage (shell size) is most similar to the native femur.

No significant change in the stress profile of the femur was reported after total hip articular replacement by internal eccentric shells (a prosthesis without a stem). In our study, the prosthesis without a stem resulted in minimal changes in stress profile of the femoral neck.

Compressive stress at the superolateral cement-bone interface, tensile stress at the inferomedial interface, and shear stress over the entire interface were reported after hip resurfacing with the Wagner prosthesis (another stem-less design). Severe bone resorption was observed at the peripheral head-neck region.

Stress concentration in the cortical bone adjacent to the rim of the prosthesis and stress shielding in the cancellous bone of the anterosuperior femoral neck were reported in a simulation of the McMinn prosthesis, which had a concentrically shaped shell, a normal stem, and normal anatomic anteversion of the femoral neck.

Failure of hip resurfacing has been associated with fracture of the femoral neck in the short term and aseptic loosening secondary to bone resorption of the femoral head in the long term. Stress shielding of the cancellous bone of the anterosuperior femoral neck was reported to be related to its fracture. However, at the 2-year follow-up bone mineral density in the femoral neck after resurfacing remained equal or even increased slightly. Similarly, the slight stress shielding of the cancellous bone at the lateral femoral neck is not attributable to its fractures. In our study, although there was stress concentration at the cancellous bone of the medial femoral neck, it was below the failure point (approximately 9 MPa), even with a peak load for activities of daily living.

Polyethylene particulate wear–induced osteolysis is the primary cause of bone resorption of the femoral head. The use of low-wear bearing couples has improved prosthesis survival. Bone resorption can also occur in areas of stress shielding or concentration, particularly the femoral head. Suboptimal fixation with subsequent micromotion can also lead to bone resorption secondary to the development of a fibrous membrane at the bone-prosthesis interface. High fluid pressure secondary to loading of such a membrane also causes bone resorption in vitro. All these mechanisms have contributed to failure of hip resurfacing secondary to aseptic loosening.

Our study had several limitations. All materials were assumed to be isotropic and homogeneous, however bone is an anisotropic, heterogeneous material. All analyses were performed using only one static loading condition. Although loading varies during different activities, the resultant hip contact force is not much different in most activities of daily living, with the exception of stair climbing. In addition, the prosthesis was simulated as a uniformly thick shell; the marketed designs have a cylindrical inner shape to enable fitting a prosthesis that is larger than a hemisphere. Some of the simulated prosthesis designs were less than optimal for stability and fixation. Despite difficulties in comparing studies due to the variability in prosthesis designs, an attempt to minimise biomechanical alteration of the proximal femur by optimising prosthesis design seems a valid approach to achieving favourable long-term outcomes. Cadaveric and in vivo studies are needed to confirm their clinical relevance and feasibility.

**ACKNOWLEDGEMENT**

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**REFERENCES**


