Stress analysis of cemented glenoid prostheses in Total Shoulder Arthroplasty

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Abstract

Glenoid component loosening is the most-frequently encountered problem in the total shoulder arthroplasty. The purpose of the study was to investigate whether failure of the glenoid component is caused by stresses generated within the cement mantle, implant materials and at the various interfaces during humeral abduction, using 3-D FE analyses of implanted glenoid structures. FE models, one total polyethylene and the other, metal backed polyethylene, were developed using CT-scan data and submodelling technique, which was based on an overall solution of a natural scapula model acted upon by all the muscles, ligaments and joint reaction forces. Material interfaces were assumed to be fully bonded. Based on the FE stress analysis, the following observations were made. (1) The submodelling technique, which required a large-size submodel and the use of prescribed displacements at cut-boundaries located far away from the glenoid, was crucial for evaluations on glenoid component. (2) Total polyethylene results in lower-peak stresses (tensile: 10 MPa, Von-Mises: 8.31 MPa) in the cement as compared to a metal-backed design (tensile: 11.5 MPa, Von-Mises: 9.81 MPa). The maximum principal (tensile) stresses generated in the cement mantle for both the designs were below its failure strength, but might evoke crack initiation. (3) The cement–bone interface adjacent to the tip of the keel seemed very likely to fail for both the designs. In case of metal-backed design, this interface adjacent to the tip of the keel appears even more likely to fail. (4) High metal–cement interface stresses for a moderate load might indicate failure at higher load. (5) It appears that both the designs were vulnerable to failure in some ways or the other. A part of the subchondral bone along the longitudinal axis of the glenoid cavity should be preserved to strengthen the glenoid structure and to reduce the use of cement.

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1. Introduction

Over the last 2 decades, clinical and radiographic feedback indicate glenoid component loosening as the most-frequently encountered problem in the total shoulder arthroplasty (TSA) (Cofield, 1984; Wirth and Rockwood, 1994; Torchia et al., 1997; Wallace et al., 1999). The cause of some cement–bone radiolucencies, in particular those seen in early post-operative radiographs, was attributed to poor cementing technique (Neer et al., 1982; Wilde et al., 1984; Barrett et al., 1987; Kelly et al., 1987). The appearance of radiolucent lines around the prosthesis keel increased over time. These findings suggest that the cement–bone radiolucencies and glenoid component loosening may be due to various reasons. Design of the implant, surgical technique, limitations in quantity and quality of bone for fixation, tissue reaction to particulate debris, rotator cuff deficiency with glenohumeral instability, and high-patient activity levels were proposed as potential causes of loosening (Boyd et al., 1990; Wilde et al., 1984; Barrett et al., 1987; Kelly et al., 1987).

The precise relationship between the cause and the effect, regarding aseptic loosening of the glenoid
prostheses and the extent to which the mechanical factors play a role in this process were not clearly understood yet. We presume that one of the major reasons for failure of the implanted glenoid structure is due to the excessive stresses generated in the implant materials or at various interfaces. Studies on stress analysis of glenoid components are few (Rohlmann et al., 1984; Orr et al., 1988; Friedman et al., 1992; Stone et al, 1999; Lacroix et al., 2000; Couteau et al., 2001). These 2-D or truncated 3-D models were mostly restricted to the glenoid area only, neglecting the effect of the forces acting on the bony structures connected to the glenoid, hence introducing large modelling errors that might lead to doubtful numerical results. Using a 3-D finite element (FE) model of implanted glenoid structure it might be possible to understand some mechanical factors responsible for probable failure mechanisms (Stone et al., 1999), and to suggest measures for improvement in the design of the prostheses.

The purpose of this study was to investigate whether failure is caused by stresses generated within the cement mantle, implant materials and at various interfaces, using 3-D FE analyses of two basic designs of cemented glenoid prostheses. These FE models, one total polyethylene and the other metal-backed polyethylene, were based on computed tomography (CT) data. A comparative analysis of these two basic models can reveal the biomechanical factors regarding choice of a prosthesis.

2. Materials and Method

2.1. Finite element model of cemented glenoid prostheses

A 3-D FE model of a glenoid (right scapula) was developed using CT-scan data. The glenoid prosthesis consists of an ultra high molecular weight polyethylene (UHMWPE) cup with a keel to anchor inside the scapula (Fig. 1). It consists of a centrally located tapered keel, which is symmetric with respect to the frontal plane. The implanted glenoid FE model was generated using ANSYS FE software. For metal-backed polyethylene design, a metal tray of 1 mm thickness, as used in practice, was specified and mesh generation was obtained utilising ANSYS FE software. The hard cortical outer shell was modelled as six-node triangular two-layered shell elements, with 0.5 mm thickness of each layer (Gupta and van der Helm, in press). A FE representation of the glenoid component with cement is shown in Fig. 2. A FE model of the implant–bone structure is shown in Fig. 3b. The FE model of the total polyethylene glenoid prosthesis contained 11,660 elements, 15,532 nodes and a total number of 56,871 active degrees of freedom (DOF). Whereas additional shell elements were used to represent the thin metallic tray, for metal-backed design. The FE model contained 12,053 elements, 15,532 nodes and a total number of 59,175 active DOF.

Bone was assumed to be a linear isotropic material. The material properties of bone elements were extracted from CT-scan image of a dry scapula. The apparent density ($\rho$) was linearly calibrated in term CT grey value ($H$), using the relationship $\rho = 527.47 + 0.44H$. Based on the proposed analytical model of cancellous bone (Gibson, 1985) and regression analysis of the experimental data of Frich (1994) power law relationships:

$$E = 0.00105\rho^2 \quad for \ \rho \leq 350 \ \text{kg m}^{-3}$$
$$E = 3.10^{-6}\rho^3 \quad for \ 350 \leq \rho \leq 1800 \ \text{kg m}^{-3}$$

were used allocate of Young’s modulus to each element. The Poissons’ ratio of the trabecular bone was assumed to be 0.3 (Dalstra et al., 1995).
2.2. Submodelling technique and applied loading conditions

The overall model of the natural scapula, which served as the reference solution included the effect of all muscles, ligaments and joint reaction forces during humeral abduction (Gupta and van der Helm, in press). The objective of using a submodel of glenoid prosthesis, with a fine mesh in the domain of inclusion of the prosthesis, was to focus our investigation in the domain of the prosthesis. The effect of the forces acting on bony structures connected to the glenoid, on the stresses generated in the implanted glenoid, cannot be neglected. In order to fulfil these two major considerations, a link between the submodel and the overall model was called for. This link can be established by transferring nodal displacements at the cut-boundary from the overall model to the submodel (Fig. 3).

The location of the cut-boundary is of crucial importance and will have predominant influence on the results. If the inclusion (prosthesis) is small as compared to the dimensions of the entire body, then it can be assumed that, sufficiently far from the inclusion, the differences between the two solutions (displacements) become zero. This procedure called for a large size of the submodel. The non-zero values of displacements at the nodes located along a cut-boundary, for each load step, were obtained from the FE solution of the overall model (Gupta and van der Helm, in press). These displacements were prescribed on the nodes located along the bottom (lateral border), the right (infraspinous and supraspinous fossa) and the posterior

Fig. 2. Finite element (FE) model of the glenoid prosthesis. (a) FE model of glenoid component; (b) FE model of prosthesis–cement configuration.

Fig. 3. Proposed submodel: (a) natural scapula model, and (b) submodel with the prosthesis.
(scapular spine) cut-boundaries in the submodel with the prosthesis (Fig. 3b). The validity of the submodelling technique was checked by comparing the values of imposed displacements at the nodes on the cut-boundaries, derived from the natural scapula, with those of a full model of the implanted scapula, having the same geometry, material properties and loading conditions. The full model of the implanted scapula contained a total number of 18,032 elements, 24,056 nodes and 96,876 DOF. The percentage difference between the imposed displacements on the submodel (derived from the natural scapula) and the nodal displacements of the full implanted scapula model, having similar nodal locations, varied between 2% and 6%. This proves that, sufficiently far from the inclusion, the differences between the two solutions (displacements) can be considered as negligibly small.

Based on the static musculoskeletal model of the shoulder (Van der Helm, 1994), the GlenoHumeral (GH) joint reaction force (393 N) was uniformly distributed on the face of elements, equivalent to the contact area (17.114 mm²) of the glenoid component. The contact area was estimated according to the Hertz theory of elastic contact for non-conforming surfaces in contact (Hertz, 1882).

### 2.3. Interface stress assessment

Since the interfaces are often the mechanically most critical zones in a bone–prosthesis structure, the distribution of normal and shear stresses are of crucial importance when evaluating an implant. The Hoffman’s failure criterion (Hoffman, 1967) was used in this study to evaluate bone–prosthesis interface failure, since it accounts for the multi-axial stresses (normal, \( \sigma_n \) and shear, \( \tau \)) in failure initiation and the gradually changing interface bone density and strength. Although experimental validations in support of the Hoffman’s criterion are very few (Stone et al., 1983; Kaplan et al., 1985), it works reasonably well for trabecular bone (Stone et al., 1983; Kaplan et al., 1985). The Hoffman number can provide for qualitative estimates (Huiskes and van Rietbergen, 1995). At a interface nodal point, the Hoffman number (\( FL \)) can be determined from the normal (\( \sigma_n \)) and shear stresses (\( \tau \)), using

\[
FL = \frac{1}{S_t S_c} \sigma_n^2 + \left( \frac{1}{S_t} - \frac{1}{S_c} \right) \sigma_n \sigma_c + \frac{1}{S_c} \tau^2
\]

with \( S_t \) and \( S_c \) being the uniaxial interface tensile and compressive strengths, respectively, and \( S_t \) the interface shear strength (Huiskes and Van Rietbergen, 1995). The interface strength was related to bone density and was assumed to be determined by density (\( \rho \)) of bone adjacent to the interface, according to

\[
S_t = 14.5 \rho^{1.71}, \quad S_c = 32.4 \rho^{1.85}, \quad S_n = 21.6 \rho^{1.65}
\]

These relationships between the density and the strength were derived from Kaplan et al. (1985) and Stone et al. (1983), who determined the static strength of trabecular for a range of bone densities. The use of interface strength data, given by Eq. (2), in the Hoffman’s failure criterion (Eq. (1)) is an attempt to account for the microstructure of bone. In the FE model, the range of bone density varied between 0.35 and 0.50 g cm⁻³; the superior regions having higher density values as compared to the central region where the keel is inserted (Batte et al., 1996; Müller-Gerbl et al., 1992). Using these data, we obtain a range of interface strength data varying between \( S_t = 2.408 \) MPa, \( S_c = 4.646 \) MPa, \( S_n = 3.821 \) MPa for \( \rho = 0.35 \) g cm⁻³ (glenoid central) and \( S_t = 4.43 \) MPa, \( S_c = 8.89 \) MPa, \( S_n = 6.88 \) MPa for \( \rho = 0.50 \) g cm⁻³ (glenoid superior). The state of interface stress can be substituted in the Eq. (1) to obtain a value, termed as the Hoffman number (\( FL \)). For \( FL \) less than 1, no interface failure is expected; for \( FL \) greater than 1, failure is expected.

### 3. Results

Results were obtained for six-load cases, from 30 to 180° abduction in steps of 30°. However, the results during 90° abduction are presented since the magnitude of forces was maximum for this load case. The maximum values of stress components are presented in Table 1.

#### 3.1. Total polyethylene glenoid component

During 60–120° abduction of the arm, the GH joint reaction force was relatively high (350–400 N) as compared to other load cases. As a result, high tensile stresses (1–10 MPa) were evoked in the cement mantle. Relatively higher stresses were observed in the cement layer immediately below the glenoid component and adjacent to the tip of the keel (Fig. 4). During 90° abduction, there was a tendency of the glenoid component along with the keel to bend in the super-

<table>
<thead>
<tr>
<th>Stress components (MPa)</th>
<th>Metal-backed PE</th>
<th>Total PE</th>
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</thead>
<tbody>
<tr>
<td>Von-Mises stress in PE cup</td>
<td>14.03</td>
<td>15.10</td>
</tr>
<tr>
<td>Normal stress, cement (ten.)</td>
<td>11.51</td>
<td>9.98</td>
</tr>
<tr>
<td>Normal stress, cement (comp.)</td>
<td>−10.15</td>
<td>−9.37</td>
</tr>
<tr>
<td>Von-Mises stress in cement</td>
<td>9.80</td>
<td>8.31</td>
</tr>
<tr>
<td>Normal stress, prosthesis-cement (ten.)</td>
<td>5.92</td>
<td>2.58</td>
</tr>
<tr>
<td>Shear stress, prosthesis-cement</td>
<td>1.22</td>
<td>1.33</td>
</tr>
<tr>
<td>Normal stress, cement-bone (ten.)</td>
<td>2.86</td>
<td>2.51</td>
</tr>
<tr>
<td>Shear stress, cement-bone</td>
<td>3.55</td>
<td>3.41</td>
</tr>
</tbody>
</table>
ior–inferior direction. This puts the superior part of the keel and the cement into higher tensile stresses (1–3 MPa) as compared to the inferior part. In some locations the tensile stresses varied between 3 and 4 MPa. Moreover, relatively higher tensile stresses (4–10 MPa) were evoked at locations adjacent to the tip of the keel. The maximum compressive and Von-Mises stresses generated in the cement mantle are –9.37 MPa and 8.31 MPa, respectively.

The cement–bone interface was subject to higher stresses as compared to the prosthesis–cement interface stresses (Table 1). The peak cement–bone interface stresses were generated at the interface adjacent to the tip of the keel (Fig. 4), whereas the peak prosthesis–cement interface stresses were generated around the superior edge of the prosthesis. The cement–bone interface around the superior edge of the prosthesis was subject to lower stresses ($\sigma_n = 1.14$ MPa, $\tau = 1.24$ MPa) as compared to the area adjacent to the tip of the keel.

The actual failure of a cement–bone interface depends on the strength of the interface bond. Using a higher strength data ($S_t = 4.43$ MPa, $S_c = 8.89$ MPa, $S_s = 6.88$ MPa for $\rho = 0.50$ g cm$^{-3}$) for interface stresses ($\sigma_n = 1.14$ MPa, $\tau = 1.24$ MPa) generated around the superior edge of the implant, the maximal $FL$ was calculated as 0.19. On the other hand, using the lower strength data ($S_t = 2.408$ MPa, $S_c = 4.646$ MPa, $S_s = 3.821$ MPa for $\rho = 0.35$ g cm$^{-3}$) for interface stresses ($\sigma_n = 2.51$ MPa, $\tau = 3.41$ MPa) generated around the glenoid central part, the $FL$ was calculated as 1.86.

A sectional view of the Von-Mises stresses in the glenoid, as indicated in Fig. 5, revealed that the stresses in the trabecular bone underlying the prosthesis, was slightly lower (0.05–3 MPa) as compared to that of the natural scapula (0.05–4 MPa). Maximum Von-Mises stress of 15 MPa was generated in the polyethylene (Table 1). The Von-Mises stresses (10–60 MPa) in the cortical bone were also reduced as compared to the natural glenoid, due to the inclusion of prosthesis (Fig. 6).

3.2. Metal-backed polyethylene component

The distribution of Von-Mises stresses in the cement mantle, varying between 1 and 9.80 MPa, during 90° abduction is presented in Fig. 7. The tensile stresses (1–2 MPa) in the cement mantle, largely, are reduced as compared to the total polyethylene component, except a few locations along the periphery where it varied between 2 and 5 MPa. Higher tensile stresses (5–11.5 MPa) were evoked at locations adjacent to the tip of the keel. The maximum value of compressive stress generated in the cement was –10.15 MPa (Table 1).

The implant–cement interface was subject to higher normal stresses, but lower shear stresses as compared to the total polyethylene design (Table 1). The maximum cement–bone interface stresses ($\sigma_n = 2.86$ MPa, $\tau = 3.55$ MPa) were generated in the area adjacent to the tip of the keel (Fig. 7). These stresses were higher than the stresses around the superior edge of the prosthesis. Considering similar data on bone density adjacent to the cement–bone interface, the $FL$ can be calculated for two different regions (Section 3.1). For higher interface bond strength ($S_t = 4.43$ MPa, $S_c = 4.98$ MPa, $S_s = 6.88$ MPa for $\rho = 0.5$ g cm$^{-3}$) in the superior side and for the peak multi-axial stresses ($\sigma_n = 3.39$ MPa, $\tau = 0.43$ MPa), the $FL$ was calculated as 0.68. Whereas, for lower strength ($S_t = 2.408$ MPa, $S_c = 4.646$ MPa, $S_s = 3.821$ MPa for $\rho = 0.35$ g cm$^{-3}$) of bone in the glenoid central region and for the peak multi-axial stresses ($\sigma_n = 2.86$ MPa, $\tau = 3.55$ MPa), the $FL$ was calculated as 2.16.

The influence of the metal-backed glenoid component on the bone is not entirely different from that of the total polyethylene component. The metal-backing and the metallic keel in the glenoid component carries the bulk of the stresses (Von-Mises), ranging from 1 to 20 MPa. Maximum Von-Mises stress of 14 MPa was generated in the polyethylene cup (Table 1). Higher Von-Mises stresses, ranging between 10 and 20 MPa, are observed in the metallic flange underlying the polyethylene cup. The metal-backing leads to tensile stresses (1–15 MPa) in the metallic flange, indicating bending of the metallic flange. The stresses in the underlying trabecular bone were lower (0.05–2 MPa) than that of the total polyethylene design (0.05–3 MPa), as indicated in Fig. 5c.
4. Discussion

The study is directed towards investigating some biomechanical factors related to the loosening of cemented glenoid prostheses, using a 3-D FE model of the implanted glenoid structure. Quantitative results of this study cannot be compared precisely with other studies, since these studies did not account for the effect of forces acting on bony structures connected to the glenoid. The structure of the scapula and the loading conditions call for the submodelling approach, so that the elastic behaviour of the overall model can be included in the analysis. The necessity of using the submodelling technique was assessed by comparing the results of the model using this technique with those where alternative solutions were obtained by prescribing artificial constraints to prevent rigid body motion. The maximum principal (tensile) stress generated in the cement mantle for total polyethylene design was 5.04 MPa, when artificial constraints were prescribed (alternative solution). Similar studies by Lacroix et al. (2000) and Murphy et al. (2001) predicted a higher value.
of 6 MPa for the same stress in the cement mantle. Whereas this stress increased to 9.98 MPa when the submodelling technique was used, thereby indicating a major difference in stress values (increase of 66–98%) as compared to alternative solution and other similar studies. Prescribing constraints to restrain rigid body motion also generates high stresses at the constraints, which seems to be inappropriate and would lead to doubtful evaluations (Lacroix et al., 2000). The results of this study therefore, seem to be more accurate.

The method employed in this study has a number of limitations. Only one particular keel shape was considered. As compared to the hip prosthesis, different stem shapes are known to generate different interface stress patterns (Huiskes, 1990). Hence, the shape will affect the evolution of the interface failure (debonding) process. Since material interfaces were assumed to be fully bonded, the calculated stresses only represent trends. In reality, however, the interfaces stresses will differ due to relative micromotions at the material interfaces (Verdonschot and Huiskes, 1997). This study, based on initial stress distribution, is a first step towards investigating the relationship between stresses generated and failure in an implanted glenoid. Only one loading case was studied and that in a FE analysis the maximum values of stress components (Table 1), especially at the interfaces of different materials, might not be necessarily calculated very precisely. Bone geometry and density distribution was based on CT-scan data of one representative scapula. Bone was assumed to be linear isotropic material. Whereas, in reality, bone is anisotropic. It should also be noted that the stresses in the low-density, open-cell trabecular bone structure were observed macroscopically, which will differ for a micromechanical model (Van Rietbergen et al., 1995). These assumptions were idealisations of the reality. Hence, the results of this study predicted certain qualitative trends.

4.1. The cement mantle

The cement mantle, presumably, is the most likely material from which the initiation of crack propagation occurs in the glenoid arthroplasty. The cement mantle, for total polyethylene design was subject to lower tensile (1–10 MPa) stresses as compared to the metal-backed design (tensile: 1–11.5 MPa). Cement is strong in compression, but weak in tension. The failure strength (tensile) of cement has been reported to be 25 MPa (Lee et al., 1977; Saha and Pal, 1984). Results indicated that the stresses generated in the cement mantle for both the designs during unloaded abduction were below its failure strength. However, these stresses might evoke crack initiation in the cement. A common value for crack initiation in cement under physiological conditions is about 5–7 MPa (Davis et al., 1987).

This study showed that for metal-backed design, the load was transferred from the polyethylene cup to the metal, which eventually carried the bulk of the load and generated lower stresses in the underlying material. As a result, stresses in the cement layer below the glenoid component were generally reduced, except some locations of high stress along the periphery. Stresses in the polyethylene were also reduced (by 8%) as compared to the total polyethylene design. But these reductions were minor, since a thin (1 mm) metallic flange was used in this model. A thicker metal backing would result in higher stress reduction in the polyethylene cup; thereby reducing the chances of polyethylene wear particle
formation. This behaviour would apply for any stiff material used in the prosthesis design. Whereas, for the total polyethylene implant the reinforcement function was offered only by the cement layer. The stresses in the cement layer below the polyethylene component were increased. The load passing through the comparatively less stiff total polyethylene component allowed for more uniform transfer of load to the underlying cement and bone. A total polyethylene design therefore appears to provide a stress pattern that was closer to a natural glenoid, compared to a metal-backed design, indicating lesser possibility of bone remodelling. These results were well supported by 2-D models on implanted glenoid (Friedman et al., 1992; Stone et al., 1999) and by 3-D model on acetabular prosthesis (Dalstra, 1993).

4.2. The cement–bone interface

The cement–bone interface adjacent to the tip of the keel, for total polyethylene design, appeared more likely to fail \( (FL = 1.86) \) as compared to the superior edge of the prosthesis \( (FL = 0.19) \). Whereas, for metal-backed design, this interface adjacent to the tip of the keel was more likely to fail \( (FL = 2.16) \) as compared to the superior edge of the prosthesis \( (FL = 0.68) \). Hence, the actual failure of an interface depends on the bone density and the bone strength of the interface bond, which was higher in the superior side of the glenoid than in the central region. These results were well supported by radiographic study of Torchia et al. (1997). They observed that the percentage of cement-bone radiolucent lines in any area of the glenoid component, excluding the keel, remained constant \( (84\%) \). The percentage of radiolucent lines, however, increased around the keel \( (42\% \) to 76\%) as did the percentage of components \( (10\% \) to 38\%) that shifted in position \( (Torchia et al., 1997) \). Probably, the most important parameter to prevent initial mechanical interface failure \( (‘primary stability’) \) is the quality \( (strength) \) of the interface bond. This factor is represented in the failure criterion, which is dependent, in reality, on the precision of the surgical procedure, the quality of the bone stock and the penetration of the bone cement for cemented implants \( (Krause et al., 1982) \) or the amount of bone ingrowth into the uncemented implants.

4.3. The prosthesis–cement interface

The prosthesis–cement interface stresses were lower than at the cement–bone interface stresses for the total polyethylene design \( (Table~1) \). However, the fixation strength of polyethylene with cement was reported to be poor \( (Fukuda et al., 1988) \), which increases the risk of a mobile prosthesis. On the other hand, for metal-backed design, the maximum metal–cement interface stresses were higher than the cement–bone interface stresses \( (Table~1) \). Considering the interface strength \( (S_t = 8\, MPa, S_s = 6\, MPa) \) it appears that the prosthesis–cement was secure against interface failure \( (Keller et al., 1980; Stone et al., 1989) \). However, failure might occur at higher load, which was comparable to similar behaviour in the cemented femoral hip components, where debonding occurred mainly at the prosthesis–cement interface \( (Harrigan and Harris, 1991) \).

4.4. The subchondral bone

The role of the subchondral bone might be crucial in the design of the implant. The subchondral bone with a higher density of about 1–1.2 g cm\(^{-3}\) as compared to the trabecular bone \( (0.1–0.5\, g\,cm^{-3}) \) offers higher interface bond strength. During surgery, this bone is usually removed and replaced by cement. A part of the subchondral bone along the longitudinal axis of the glenoid cavity should be preserved to strengthen the glenoid structure, since removal of bone and replacing it by cement means a substantial loss in the strength of the glenoid and an increase in stresses in the cement. A thin layer of cement could be used in between the implant and subchondral bone for secured primary fixation.

4.5. Comparison with other studies

Results of this study may be compared with similar studies using simplified 3-D models and loading conditions \( (Lacroix et al., 2000; Couteau et al., 2001; Murphy et al., 2001) \). These studies were mainly focussed on the calculating stress distributions in the implanted glenoid area due to the GH-joint reaction force only, omitting the effect of bony structures connected to the glenoid. The cement stresses \( (tensile, Von-Mises) \) mostly varied between 1 and 3 MPa, except some locations where it varied between 3 and 7 MPa \( (Lacroix et al., 2000; Couteau et al., 2001; Murphy et al., 2001) \). A maximum compressive stress of \( –11.8\, MPa \) was generated in the cement for total polyethylene design \( (Couteau et al., 2001) \). Qualitative agreement with these studies \( (Lacroix et al., 2000; Couteau et al., 2001; Murphy et al., 2001) \) notwithstanding, the results of this study were more accurate since it was based on the submodelling technique. Hence, this study is more useful to understand the load transfer mechanism in the implanted glenoid, with regard to some failure scenarios.

5. Conclusions

Using a detailed 3-D FE model of glenoid prosthesis based on CT-scan data and physiological loading
conditions the following qualitative conclusions can be made.

(1) A large size of the submodel and the use of prescribed displacements at cut-boundaries located far away from the glenoid would have considerable influence on the results.

(2) The maximum principal (tensile) stresses generated in the cement mantle for both the designs were below its failure strength, but might evoke crack initiation.

(3) For total polyethylene design, the cement–bone interface adjacent to the tip of the keel appeared very likely to fail as compared to the superior edge of the prosthesis. Whereas, this interface for metal–backed design appeared even more likely to fail as compared to the non-backed design.

(4) Higher implant–cement interface stresses were generated for metal-backed design as compared to total polyethylene design, which might indicate failure at higher loads.

(5) A total polyethylene design appears to provide a more physiological stress distributions than a metal backed design in the cancellous bone regions underlying the prosthesis.

(6) A part of the subchondral bone should be preserved to increase the strength of the implanted glenoid structure and to reduce the use of cement.

(7) It appears that the stresses generated within the cemented keel-type glenoid components is one of the reasons behind loosening and eventual failure of the implant.

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References


