FEMORAL COMPONENT REVISION USING IMPACTED MORSellISED CANCELLOUS GRAFT

A BIOMECHANICAL STUDY OF IMPLANT STABILITY

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We have tested the axial and torsional stability of femoral components after revision arthroplasty in a cadaver model, using impacted morsellised cancellous graft and cement. Each one of six matched pairs of fresh frozen human femora had either a primary or a revision prosthesis cemented in place. For the ‘revision’ experiments, all cancellous bone was removed from the proximal femur which was then over-reamed to create a smooth-walled cortical shell. An MTS servohydraulic test frame was used to apply axial and torsional loads to each specimen through the prosthetic femoral heads with the femur submerged in isotonic saline solution at 37°C.

The mean subsidence was 0.27 ± 0.17 mm for the primary and 0.52 ± 0.30 mm for the revision groups. The difference was statistically significant (p < 0.025), but the mean subsidence was <1 mm in both groups. The mean maximum torque before failure was 42.9 ± 26.9 N-m for the primary and 34.8± 20.7 N-m for the revision groups. This difference was not statistically significant (p > 0.015).

Based on our results we suggest that revision of the femoral component using morsellised cancellous graft followed by cementing with a collarless prosthesis with a polished tapered stem restores the integrity of the proximal femur and provides immediate stability of the implant.


Revision of the femoral component of a hip arthroplasty is a challenging problem. Immediate stability of the implant at the time of revision may be difficult to achieve due to underlying osteolysis. An enlarged sclerotic proximal femur may be deficient in both cancellous and cortical bone (Hungerford and Jones 1988; Poss et al 1988; D’Antonio et al 1989). The choice of implant depends on several factors including the extent of loss of proximal femoral bone, the quality of the remaining host bone, the activity level of the patient, and the experience of the surgeon (Callaghan 1992).

The implants have included uncemented, extensively porous-coated stems (Engh et al 1988; Paprosky, Bradford and Younger 1994), proximally-coated long-stem implants (Gustilo and Pasternak 1988; Hedley, Gruen and Ruoff 1988; Hungerford and Jones 1988; Mallory 1988; Malkani et al 1996) and cemented prostheses (Pellicci, Wilson and Sledge 1982; Callaghan et al 1985; Kavanagh, Ilstrup and Fitzgerald 1985; Rubash and Harris 1988; Marti et al 1990). The results of cemented revision have been variable, mainly due to the lack of interdigitation between the cement and the remaining sclerotic proximal femur (Dohmoe et al 1988). Proximally porous-coated stems have shown high rates of failure due to the incidence of intra-operative fractures and difficulty in obtaining proximal ingrowth from the poor host bone (Malkani et al 1996). Extensively coated, cementless long stems with stable distal fixation show good intermediate results at the cost of proximal stress shielding (Engh et al 1988; Paprosky et al 1994).

For massive proximal bone loss, methods of management have included the use of cortical strut grafts (Emerson et al 1992; Head et al 1993), allograft prosthetic composites (Gross et al 1993; Chandler et al 1994), and tumour-type prostheses (Malkani, Sim and Chao 1993). Cortical strut grafts have been useful in restoring structural bone loss around the proximal femur but their main function is to augment the existing bone stock and they cannot be used to provide implant stability.

The use of impacted morsellised cancellous allograft in the proximal femur followed by insertion of a cemented, polished, tapered femoral component has been advocated for proximal femoral revision (Gie et al 1993; Ling, Timperley and Linder 1993). The ‘Ling technique’ is believed...
to reconstitute proximal femoral bone gradually through substitution and incorporation (Ling et al 1993). This creates interfaces between the allograft and host bone, the cement and the allograft, and the implant and the cement. This raises concern about the ability to achieve immediate stability of the implant and the planning of postoperative rehabilitation. The immediate implant stability which is achieved has not been tested biomechanically.

Our aim was to test the axial and torsional stability of the femoral component after insertion of a cemented polished tapered stem into a bed of impacted morsellised cancellous graft in the cadaver femur.

MATERIALS AND METHODS

Specimen preparation. The proximal two-thirds of six matched pairs of fresh frozen human cadaver femora, from patients of mean age 73 years (62 to 88), were sealed in plastic and stored at a temperature of –20°C. Each specimen was radiographed to identify bony abnormality. All were thawed in a refrigerator at 4°C for 24 hours and then at room temperature until tested.

A primary and a revision arthroplasty was performed on each pair. On the primary side, a collarless polished tapered femoral component (CPT; Zimmer Inc, Warsaw, Indiana) was inserted using modern cementing techniques after standard reaming and broaching. The size of the implant was determined by using preoperative radiographs and templates.

In the opposite paired femur all cancellous bone was removed followed by over-reaming of the femoral canal to create a smooth cortical envelope. The heads from each pair of femora were used as cancellous grafts for restoration of the bone loss. A bone mill (Tracer Designs, Santa Paula, California) was used to grind the heads into chips and the resulting morsellised graft was impacted into the femur using CPT instrumentation as described by the manufacturer (Zimmer Inc, Warsaw, Indiana). In the final phase of impaction, smooth tamps were used to prepare the site for the introduction of the cement (Fig. 1). An identical prosthesis was then inserted using the same cementing technique. The amount of morsellised material obtained from the two heads was sufficient to create an adequate graft to fill the over-reamed canal and to accommodate an implant of appropriate size and a cement mantle 1 mm thick. The graft varied in thickness from approximately 3 mm at the corners of the implant to over 10 mm along its face. Radiological markers were inserted on the shaft of both femora at the level of the distal end of the implant to indicate the position of the implant before and after mechanical testing.

Mechanical testing. Each specimen was mounted on a stand on an MTS servohydraulic test frame (Model 858...
Bionix; MTS Systems Inc, Minneapolis, Minnesota). The cadaver femora were loaded through a 28 mm femoral head by direct contact with a concave indentation in a metal disc attached to the actuator of the MTS machine. The mounting stand was pinned at its base, allowing a rotational degree of freedom in the plane of the implant (Fig. 2). We performed axial cyclic loading and measurement of subsidence because the principal modes of failure in this technique are thought to be subsidence of the prosthesis and the inability of the graft to withstand repeated loading.

An axial load cell (Model 62.10A-05; MTS Systems Inc, Minneapolis, Minnesota) was used to record the compressive forces applied to the femur and also for feedback control of the cyclic testing. In each test the implant was slowly loaded through the prosthetic head to 440 N, simulating 70% of the body-weight. The position was recorded and an anteroposterior radiograph obtained. Cyclic testing was then performed by increasing the compressive force to 1320 N (approximately 200% of the body-weight during the stance phase of gait) and returning to 440 N (the swing phase of gait) at a frequency of 1 Hz over 5000 cycles to simulate the first few days to weeks of limited load-bearing. The axial displacement sensor for the linear actuator of the MTS machine was used to measure the position of the femoral head. The maximum and minimum positions were recorded for each cycle. The total axial displacement was defined as the difference in the position of the actuator between the first and last cycles under 440 N load. The axial subsidence was defined as the difference in position between the fifth and last cycles under 440 N load. The first five cycles were excluded to avoid the rapid initial settling of the specimen which occurs on initial loading. After cyclic testing another radiograph was obtained to check for visual evidence of subsidence as compared with that measured mechanically.

Figure 2a – Cyclic axial loading was applied through the prosthetic femoral head over 5000 cycles between 70% and 200% of average body-weight. A water chamber was used to keep the femur, graft, cement and prosthesis at 37°C to simulate clinical conditions. Figure 2b – Torsional loading was applied about the axis of the prosthesis and femoral shaft.

To test the ultimate axial stability the final 500 cycles in each test were used to compute a final rate of subsidence taken as the slope of a 'best-fit' line through the last 500 data points. This slope indicates the incremental distance moved with each cycle at the end of the loading period.

The femora were then loaded to failure in torsion using another rig (Fig. 2). We performed torsional tests as a definitive measure of the static stability. An axial load of 440 N was applied and the prosthesis then rotated at a constant angular rate of 2.0°/s. The angle and the applied torque were recorded simultaneously. The torsional strength (N-m) was determined as the maximum torque (N-m) applied to the specimen before failure. The torsional stiffness (N-m/°) was calculated as the slope of the linear portion of the curve generated by plotting applied torque (N-m) against the angle of deformation (°).

Data analysis. The total axial displacement, the axial subsidence, and the rate of final subsidence were analysed to determine the extent of subsidence of the implant and to ascertain if there was any statistically significant difference.

Table I. Results for cyclic loading in vitro of revision and primary hip prostheses in a cadaver model

<table>
<thead>
<tr>
<th>Specimen</th>
<th>Total axial displacement after 5000 cycles (mm)</th>
<th>Axial subsidence from 5 to 5000 cycles (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Primary</td>
<td>Revision</td>
</tr>
<tr>
<td>1</td>
<td>-0.44</td>
<td>-1.35</td>
</tr>
<tr>
<td>2</td>
<td>-0.36</td>
<td>-1.15</td>
</tr>
<tr>
<td>3</td>
<td>-1.19</td>
<td>-2.33</td>
</tr>
<tr>
<td>4</td>
<td>-0.66</td>
<td>-0.66</td>
</tr>
<tr>
<td>5</td>
<td>-0.49</td>
<td>-0.79</td>
</tr>
<tr>
<td>Mean ± SD</td>
<td>-0.63 ± 0.33*</td>
<td>-1.25 ± 0.66*</td>
</tr>
</tbody>
</table>

* p < 0.05
† p < 0.025

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between the primary and revision sides. The sinusoidal axial load varied between a maximum of 1320 N, representing the stance phase, and a minimum of 440 N, representing the swing phase of gait. Only measurements in the swing phase were analysed in order to minimise the influence of flexion of the femur and implant which is more pronounced in the stance phase. The axial displacement, the subsidence and the rate of subsidence on the primary side were compared with those on the revision side and tested for statistical significance using paired \( t \)-tests, as were the torsional strength and stiffness.

**RESULTS**

**Axial subsidence.** An error in collection of the data required elimination of one set of results of axial testing of the six paired femora. The results of axial loading on the remainder are shown in Figure 3 and Tables I and II. Both primary and revision specimens showed rapid subsidence in the first few cycles followed by a steadily decreasing rate with further loading. In every case, nearly all of the total displacement was seen during the first 1000 cycles of loading with relatively little occurring later. The total axial displacement by 5000 cycles was 0.63 ± 0.33 mm for the primary and 1.25 ± 0.66 mm for the revision prostheses. The axial subsidence from 5 to 5000 cycles was 0.27 ± 0.17 mm for the primary and 0.52 ± 0.30 mm for the revision prostheses. The results of the paired \( t \)-tests showed a statistically significant difference in both the axial displacement and subsidence between the primary and revision groups (\( p < 0.025 \)). The mean final rates of subsidence were 0.0108 ± 0.0082 \( \mu \)m/cycle and 0.0143 ± 0.0097 \( \mu \)m/cycle for the primary and revision prostheses respectively; these were not significantly different (\( p > 0.05 \)). Radiological analysis was inadequate since measurement was not able to distinguish the small displacements observed.

**Torsional testing.** The results of torsional testing are presented in Figure 4 and Table III. A testing error eliminated one pair of specimens. In every case failure was seen as a longitudinal or spiral crack in the femoral cortex. Failure or gross rotational displacement within the graft was not seen in the revision specimens. The torque at failure was 42.86 ± 26.9 N-m for the primary and 34.82 ± 20.7 N-m for the revision specimens. The torsional stiffness was 5.06 ± 3.9 N-m/° for the primary and 6.06 ± 3.3 N-m/° for the revision specimens. The strength and stiffness of the primary and revision specimens were not significantly different (\( p > 0.15 \)).

**DISCUSSION**

Concern about the stability of the implant after revision of the femoral prosthesis using impacted morsellised cancellous allograft has led to uncertainty about the immediate postoperative management. We have tested the axial and torsional stability of the femoral component after revision in a cadaver model, and although we could not reproduce the conditions pertaining in vivo we were able to compare...
the immediate stability of the prosthesis after the Ling technique with that of a standard primary cemented component.

The total axial subsidence was greater in the revision than in the primary group; most of the axial displacement in both occurred in the first few cycles. This probably represents settling and wedging of the polished implant into the cement mantle. In the revision group, rapid settling would also be expected in the morsellised bone which tended to expand slightly when left uncompressed for the period before the specimens were placed in the testing machine. Under clinical conditions, the hip is reduced intraoperatively and loaded immediately and continually by muscle contraction, so that there is no opportunity for graft expansion to take place. We therefore felt that it was more relevant to the clinical state to observe the subsidence beginning with the fifth cycle. The mean subsidence from 5 to 5000 cycles was 0.27 mm in the primary compared with 0.52 mm in the revision prostheses; neither would be considered clinically unstable. In both groups the subsidence steadily decreased with increasing loading cycles with no statistically significant difference in the final rate of subsidence (Table II), indicating increasing axial stability of the femoral component.

Analysis of the torsional data showed no statistically significant difference in strength or stiffness between the revision and primary groups. There was a large variation in the torsional findings (strength range 14.4 to 75.6 N-m; stiffness range 1.8 to 11.0 N-m/°), but this was primarily due to the variability in the quality of the bone in individual specimens. Torsional strength and stiffness were more dependent on the pair of femora being tested than on whether the procedure was of the primary or revision type (Table III). Statistically, the torsional stability was no different in the revision than in the primary group. Both failed by longitudinal or spiral cracking of the cortex, indicating that the implant/cement/graft/cortex interfaces were stronger in torsion than the bone itself. While increased axial stability from driving a wedge-shaped prosthesis into the graft material may be apparent, the excellent torsional characteristics strongly support the biomechanical rationale for this procedure.

Rotational moments have been reported about the hip during stair-climbing which are similar to the torques observed at failure in our experiment (Bergmann, Graichen and Rohlmann 1995). Our experimental method may have created a tendency towards lower torsional strength than would be expected in a whole healthy femur but failure occurred at the cortex and not in the prosthesis/cement/graft complex. The clinical implications of high torsional loading of the impacted allograft/host bone interface can only truly be addressed by further clinical investigation.

It is interesting to speculate where the slightly greater subsidence occurred in the revision group. The size of the prosthesis was the same in both. The wedge-shaped polished implant attempts to subside in the cement mantle causing an outward or radial compressive effect into the layers. The subsidence within the cement was probably the same in the two groups and the difference lay in the interposition of the impacted morsellised graft between the cement mantle and the solid host bone in the revision model. Compression will cause radial displacement of the graft which results in subsidence of the implant. Shear forces along the interfaces may allow slippage of the cement along the graft or of the graft along the smooth
cortical wall. Because of the wedge effect and interdigitation between the cement and the graft it is less likely that slippage will occur at the cement-graft interface. We therefore believe that the subsidence is due to a combination of slight slippage of the graft mantle against the smooth cortical wall of the host femur and compression as a result of the wedging of the implant and cement into it. The success of the technique is due to the increased friction developed by the high compressive forces; this results in progressively improved stability and reduced slippage.

We feel that revision of the femoral component using impacted morsellised cancellous allograft can provide immediate stability of the implant. Firm impaction of the graft material is essential. Potential fractures due to high stresses caused by graft impaction may be managed by supplementing the host cortical bone with allograft cortical struts. We were concerned about the immediate torsional stability under high torsional loading such as during stair-climbing and have allowed our patients to progress gradually to full weight-bearing by three months. Our experimental results are encouraging but must be confirmed by long-term clinical investigation.

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REFERENCES


