IMPINGEMENT AND LOOSENING OF THE
LONG POSTERIOR WALL ACETABULAR IMPLANT

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Long posterior wall (LPW) Charnley acetabular implants are widely used as it is believed that the LPW helps to prevent dislocation. This has, however, not been proven statistically. In a preliminary study of these implants removed at revision marked erosion of the LPW was frequently seen, indicating that repetitive impingement may occur. The influence of the long posterior wall was therefore investigated mathematically.

LPW and standard sockets were found to be equally likely to dislocate provided that the standard socket was antverted 5° more than the LPW socket. With simulated external rotation, LPW sockets impinge 30% earlier than standard sockets. When impingement occurs a torque is applied to the components, which increases the shear stresses at the cement–bone interface. The torques, although not large enough to dislodge the socket immediately, are repetitive and so may contribute to loosening. The LPW socket can generate twice as much torque as the standard socket and therefore is more likely to loosen.

Charnley introduced the long posterior wall (LPW) acetabular socket to help prevent posterior dislocation of his total hip replacement (THR). It has, however, not been proven statistically that the LPW does prevent dislocation. Charnley observed that erosion of the long posterior wall occurred and was due to impingement of the neck of the femoral component (Charnley 1979). This impingement generates particulate polyethylene debris, and may place abnormal loads on the socket. A mathematical model of a THR was developed to investigate the effect of the LPW on dislocation and impingement and the resultant abnormal loading.

MATERIALS AND METHODS

Preliminary study. Ten LPW sockets retrieved at revision operations were examined. Four showed severe erosion of the LPW. This confirms that repeated impingement of the prosthetic neck occurs and generates large amounts of particulate debris (Fig. 1).

The mathematical model. The articular surface of a standard Charnley socket (Fig. 2) is hemispherical and is sunk 2 mm below the surface, at the base of a short cylinder. From the edge of the cylinder, the inner rim of the socket, the surface slopes out to the outer rim of the socket. The LPW socket is similar except that, posteriorly, the cylinder is continued out to the level of the outer rim. With the standard socket, neck impingement (I) occurs first on the inner rim. Subsequently it may impinge on the outer rim; this is known as secondary impingement (SI) (Nicholas et al 1990). With increasing external rotation of the hip, posterior impingement and anterior dislocation (D) occur; with flexion and internal

Fig. 1
Photograph of a socket showing severe erosion of the LPW.
rotation, there is anterior impingement and then posterior dislocation.

Once neck impingement has developed, any further angulation tends to lever the head out of the socket, applying a torque to the socket (Nicholas et al 1990). There is, of course, an equal and opposite torque applied to the stem. These impingement torques are resisted by shear forces at the cement–bone interfaces, and may therefore predispose to loosening.

Acetabular implants are usually inserted so that the outer rim is level with or sunk below the bony acetabular rim, and even if osteophytes are removed, they are trimmed only to the level of the implant. When secondary impingement occurs, therefore, it is usually against bone rather than against the rim of the acetabular cup; in this situation, no torque is applied to the socket.

Mathematical models were developed of the 47 mm standard and LPW sockets, articulating with standard narrow-necked femoral components, the dimensions having been taken from the components (Fig. 3). The models were used to calculate the angles at which impingement and dislocation occur, and the torques which develop. The equations used in the model are shown in the appendix.

RESULTS

The angles at which impingement (I), secondary impinge-

ment (SI) and dislocation (D) occur are shown graphically in Figure 4; they are measured from neutral, which is defined as the position at which the axis of the neck and the socket coincide. Also shown is the torque factor, which is the factor by which the joint load has to be multiplied to determine the torque.

DISCUSSION

LPW sockets were found to dislocate anteriorly earlier and posteriorly later than standard sockets. The range of rotation of the femoral component between anterior and posterior dislocation for the standard socket is 144°. The range for the LPW socket is the same but the arc of the movement is antverted by about 6°. A standard socket antverted 6° more than an LPW socket is therefore as likely to dislocate, both anteriorly and posteriorly, as the LPW socket.

With the standard socket, impingement occurs anteriorly and posteriorly at 53° from neutral. With the LPW socket, it occurs anteriorly at the same angle but at 38° posteriorly (see Fig. 2). If a standard socket is

THE JOURNAL OF BONE AND JOINT SURGERY
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antverted 6° more than an LPW socket, and the femoral component is in the standard position with some anteversion, then there is posterior impingement at about 30° of external rotation with the standard socket and 20° of external rotation (about 30° earlier) with the LPW socket. Posterior impingement is therefore much more likely to occur with an LPW than with a standard socket.

With anterior impingement the resultant torques are the same for both sockets but with posterior impingement the torque applied to the LPW socket is twice that for the standard socket. An LPW socket is therefore not only more likely to impinge posteriorly than a standard socket, but when it does so the torque generated is very much larger. For example, if impingement occurs when a standing patient weighing 75 kg rotates his body away from the weight-bearing side, a torque of about 30 Nm may be generated. Andersson, Freeman and Swanston (1972) found by experiment that to dislodge a cemented socket a torque of about 100 Nm was necessary. The impingement torques are therefore not quite large enough to dislodge a socket but they are repetitive and therefore must contribute to loosening. They are an order of magnitude greater than the frictional torques (the coefficient of friction being about 0.1).

Impingement torques tend to prevent further angulation and it might therefore be argued that they may stop dislocation. This, however, is not the case as dislocation usually occurs when the limb is not bearing weight and the joint is not heavily loaded; consequently, the torques are too low to prevent further angulation.

Nicholas et al (1990) using an experimental model of a Charnley THR investigated impingement and dislocation with acetabular augmentation. Ranges obtained between anterior and posterior impingement for both standard and LPW sockets from their experimental model were similar to those obtained in the mathematical model described above, thereby validating the latter. They did not, however, determine the torques generated by posterior impingement with LPW sockets.

Theoretically, LPW sockets are not less likely to dislocate than standard sockets but are more likely to impinge, and when they do so they generate larger shear stresses at the cement–bone interface. They, and the femoral components used with them, are therefore more likely to loosen than are standard sockets. I believe that LPW sockets should not be used. Instead standard sockets should be used and they should be anteverted about 5° more than the LPW socket. More importantly, every effort should be made to prevent THR from impinging, as impingement generates wear debris and shear stresses at the bone–implant interface both of which predispose to loosening.

APPENDIX

The following equations can be used to determine the angles of impingement and dislocation, and the impingement torques for any THR provided the relevant dimensions defined in Figure 4 are known:

**Impingement angle**

\[ F = 90 - \tan^{-1}\left(\frac{X}{R}\right) - \sin^{-1}\left(\frac{N}{\sqrt{X^2 + R^2}}\right) \]

**Secondary impingement**

\[ F = 90 - \tan^{-1}\left(\frac{W-X}{P-R}\right) - \sin^{-1}\left(\frac{M-N}{\sqrt{(W-X)^2 + (P-R)^2}}\right) \]

**Dislocation**

- **D before SI**
  \[ F = 90 - \tan^{-1}\left(\frac{X-Y}{2R}\right) - \sin^{-1}\left(\frac{N}{\sqrt{(X-Y)^2 + 4R^2}}\right) \]
- **D after SI**
  \[ F = 90 - \tan^{-1}\left(\frac{W-Y}{P+R}\right) - \sin^{-1}\left(\frac{M}{\sqrt{(W-Y)^2 + (P+R)^2}}\right) \]

**Torque = Torque factor T x Joint load, where**

\[ T = N + \left(\frac{X-L}{\sin F}\right) \text{ where } F = 90 - \tan^{-1}\left(\frac{L-X}{R}\right) - \sin^{-1}\left(\frac{N}{\sqrt{(L-X)^2 + R^2}}\right) \]

Between I and D there is a transition point at which the head begins to move over the inner rim. The equations governing the torques after this transition are not included because they are complex, and as after transition the torques are relatively low and therefore unimportant. The equations governing torque after SI are not included as they depend on whether bony or prosthetic impingement occurs. If there is bony impingement then no torques are generated; if prosthetic impingement occurs first then, although the torques are large, they occur only very late and are therefore relatively unimportant.

The author would like to thank Mr J. Kenwright for his help and advice.

No benefits in any form have been received or will be received from a commercial party related directly or indirectly to the subject of this article.

REFERENCES

