IN VIVO WEAR OF POLYETHYLENE ACETABULAR COMPONENTS

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Polyethylene acetabular cups retrieved at revision surgery were measured by a shadowgraph technique to determine linear wear, and the values were compared with those obtained from radiographs. There was a close correlation between them, although the radiographic measurements slightly underestimated the true wear. Average linear wear rates for surface-replacement components were much greater than those for conventional prostheses with femoral heads up to 32 mm in diameter.

Volumetric wear, calculated using a new formula, was found to be less than previously reported in vivo, and similar in magnitude to the results of experimental wear tests in vitro. The volumetric wear rates were greatest for the surface-replacement components and, for conventional components, were found to increase in a linear manner with component diameter.

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Wear of the polyethylene components of artificial joints is an important factor in their long-term durability (Walker 1977; Charnley 1979b; Rose and Radin 1982). Migration of the femoral head into the acetabular component may lead to loosening (Wroblewski 1988) and polyethylene wear debris may cause bone resorption at the bone-implant interfaces (Howie et al 1988).

Serial measurements of acetabular cup thickness are a convenient method of assessing linear migration of the femoral head through the polyethylene, but early work suffered from errors of technique which have since been improved. Only a few studies have correlated radiographic measurements with direct measurements of retrieved components and these have generally been performed for small subsets of the populations studied (Clarke et al 1976; Wroblewski 1985; Livermore, Ilstrup and Morrey 1990). It has not, therefore, been shown that the relationship between radiographic and true wear measurements holds for all sizes and configurations of implant (Hoeltzel et al 1989; Livermore et al 1990). From in vivo studies it has not been possible accurately to assess the volume of material released to the surrounding tissues. Simulator studies (Duff-Barclay and Spillman 1967; Scales, Kelly and Goddard 1969; Wright and Scales 1977; Wroblewski 1979; Wright and Scales 1980; McKellop and Rostlund 1990) have provided a quantifiable measure of volume change but they probably underestimate the volume produced.

Using the observed migration of the prosthetic femoral head as a measure of combined wear and creep, we have tried to determine: 1) the factors, in the patient or in the component, which correlate with the wear rate; 2) the accuracy of correlation between radiographic measurements and those made on retrieved components; and 3) the volume of wear material produced by various sizes of prosthesis.

MATERIALS AND METHODS

The shapes of 60 acetabular components retrieved during revision surgery for aseptic loosening were replicated with polymethylmethacrylate bone cement (PMMA) (Fig. 1). A double-casting technique was used as previously described to minimise the effect of shrinkage of the PMMA (Ma, Kabo and Amstutz 1983). There
were 20 surface-replacement components and 40 conventional components. The inside diameters of the cups ranged from 22 to 32 mm in the conventional components and from 36 to 54 mm in the surface replacements. The minimum thickness of the polyethylene was 8 mm in the conventional cups and 3 to 5 mm in the surface replacements. Casts were made of 30 more retrieved cups but these were excluded because component deformation, fracture, or damage to the articular surface during removal made them unsuitable for measurement.

**Measurement of retrieved components.** The wear of polyethylene components in vivo occurs in one direction, the femoral head cutting a cylindrically-shaped path through the material (Charnley and Halley 1975; Dowling et al 1978; Griffith et al 1978; Charnley 1979a; Wroblewski 1985; Gebhard, Kabo and Amstutz 1990). This is shown in Figures 1 and 2. The angle of the wear direction (β), was measured with respect to the plane of the mouth of the component. This angle did not necessarily represent the line of weight-bearing. The other parameters shown in Figure 2 were used to determine the volume of material displaced. It is to be noted that the wear track is not a complete cylinder (with volume = π linear wear). Since the acetabular component consists of only half a sphere, the direction of wear relative to its flat face must be taken into account. The correct expression for the volume of material lost (V) is given by the equation:

\[
V = \pi r^2 d - \pi \cos^{-1} \left( \frac{d \tan(\beta)}{r} \right) \left[ d^2 \tan^2(\beta) - d^2 + \frac{r}{3 \tan(\beta)} \right] + \pi \left[ \frac{d^2 \tan(\beta)}{r^2} \right] \frac{3}{2}
\]

The mathematical details of the measurement technique and volume calculation have been described elsewhere and are beyond the scope of this paper.

**Radiographic measurements.** Linear wear was measured, as reported previously, on anteroposterior radiographs taken shortly after the prosthesis had been implanted and again immediately before the revision operation (Charnley and Halley 1975; Clarke et al 1976; Charnley 1979a; Wroblewski 1985; Rimnac et al 1988). These were taken using standardised grid films and measurements were made in 55 cases using an overlay template with concentric rings and diameter lines and a digitising stylus. The tablet was calibrated for each radiograph from the known dimensions of the femoral component.

The patient's age, weight, and average hip rating score were obtained from the clinical notes. The lateral opening angle of the acetabular component was measured on the immediate postoperative radiograph and the direction of the weight-bearing axis was calculated as the sum of the lateral opening angle and the angle, β.

Linear and volumetric wear rates were calculated from the differences observed between the first operation and the revision procedure. Based on only two points in time the rates of wear appear to be linear but this does not necessarily mean that the actual wear process was constant between those dates.

Linear regression analyses, correlations, ANOVA, and Student's t-tests were used to determine factors which significantly increased polyethylene wear and to compare the differences between the surface and conventional replacement groups as well as to identify trends within the groups.

**RESULTS**

**Demographic data.** Table I gives the clinical details of the patients and their functional ratings based on the UCLA 10-point hip scale. The follow-up for the surface-replacement group was significantly shorter (p = 0.0001) and their average age was younger, although not
significantly (p = 0.103). Pain, walking, function and activity scores were not remarkably different between the two groups.

**Linear wear.** The actual diameters of the prosthetic femoral heads, which ranged from 22 to 54 mm in size, were found to be within 0.1% of the manufactured sizes. The true linear and volumetric wear values calculated from measurements of the retrieved components are components and only 42% for the surface replacements. The average wear rates for conventional components with femoral head diameters of 22 mm, 26 mm, 28 mm, and 32 mm, were 0.127 mm/yr, 0.229 mm/yr, 0.234 mm/yr, and 0.214 mm/yr respectively. It is interesting to note that the average wear rates for the Charnley components in this series, as determined from the radiographs and by direct measurement, were 0.14 mm/yr and 0.13 mm/yr, respectively, precisely the values reported by Wroblewski (1985) for his series of Charnley components. There was a general trend for the linear wear rate to increase with increasing cup diameter. The surface-replacement components of larger diameter had the highest wear rates (p < 0.001). This may be due either to the thinner layer of polyethylene, the greater frictional

### Table I. Details of the patients and their functional rating related to the diameter of the prosthetic femoral head

<table>
<thead>
<tr>
<th>Diameter (mm)</th>
<th>Age (yr)</th>
<th>Weight (kg)</th>
<th>Follow-up (yr)</th>
<th>Functional rating*</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Number</td>
<td>Mean Range</td>
<td>Mean Range</td>
<td>Mean Range</td>
</tr>
<tr>
<td>Conventional cups</td>
<td></td>
<td></td>
<td></td>
<td>Pain</td>
</tr>
<tr>
<td>22</td>
<td>5</td>
<td>51.8 (46 to 57)</td>
<td>80.6 (58 to 107)</td>
<td>14.4 (9.4 to 19.2)</td>
</tr>
<tr>
<td>26</td>
<td>3</td>
<td>44.3 (19 to 58)</td>
<td>58.3 (55 to 63)</td>
<td>13.0 (8.5 to 15.5)</td>
</tr>
<tr>
<td>28</td>
<td>23</td>
<td>49.7 (18 to 76)</td>
<td>64.8 (44 to 93)</td>
<td>10.4 (5.1 to 19.5)</td>
</tr>
<tr>
<td>32</td>
<td>9</td>
<td>54.7 (38 to 66)</td>
<td>72.0 (57 to 104)</td>
<td>9.8 (6.0 to 16.8)</td>
</tr>
<tr>
<td>All</td>
<td>40</td>
<td>50.7 (18 to 76)</td>
<td>67.7 (44 to 107)</td>
<td>10.9 (5.1 to 19.5)</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Surface replacements</th>
<th></th>
<th></th>
<th></th>
<th>Mean Range</th>
<th>Pain</th>
<th>Walking</th>
<th>Function</th>
<th>Activity</th>
</tr>
</thead>
<tbody>
<tr>
<td>36</td>
<td>2</td>
<td>23.0 (14 to 32)</td>
<td>58.0 (50 to 66)</td>
<td>5.7 (5.4 to 5.9)</td>
<td>8.3</td>
<td>7.5</td>
<td>8.3</td>
<td>6.5</td>
</tr>
<tr>
<td>39</td>
<td>3</td>
<td>20.3 (15 to 25)</td>
<td>51.5 (51 to 52)</td>
<td>6.6 (3.3 to 11.3)</td>
<td>8.8</td>
<td>7.7</td>
<td>7.7</td>
<td>6.0</td>
</tr>
<tr>
<td>43</td>
<td>4</td>
<td>50.8 (29 to 66)</td>
<td>71.0 (57 to 82)</td>
<td>3.4 (0.9 to 5.0)</td>
<td>7.8</td>
<td>7.6</td>
<td>7.1</td>
<td>5.8</td>
</tr>
<tr>
<td>47</td>
<td>7</td>
<td>52.4 (22 to 74)</td>
<td>76.3 (66 to 82)</td>
<td>5.3 (1.3 to 10.7)</td>
<td>8.1</td>
<td>7.3</td>
<td>6.7</td>
<td>6.0</td>
</tr>
<tr>
<td>51</td>
<td>2</td>
<td>61.0 (54 to 68)</td>
<td>90.0 (78 to 102)</td>
<td>7.5 (6.4 to 8.6)</td>
<td>7.5</td>
<td>7.8</td>
<td>8.0</td>
<td>6.8</td>
</tr>
<tr>
<td>54</td>
<td>2</td>
<td>47.5 (47 to 48)</td>
<td>93.0 (91 to 95)</td>
<td>9.1 (8.8 to 9.4)</td>
<td>9.8</td>
<td>9.3</td>
<td>8.3</td>
<td>8.3</td>
</tr>
<tr>
<td>All</td>
<td>20</td>
<td>44.7 (14 to 74)</td>
<td>73.3 (50 to 102)</td>
<td>5.7 (0.9 to 11.3)</td>
<td>8.3</td>
<td>7.7</td>
<td>7.4</td>
<td>6.3</td>
</tr>
<tr>
<td>Total</td>
<td>60</td>
<td>48.7 (14 to 76)</td>
<td>69.7 (44 to 107)</td>
<td>9.2 (0.9 to 19.5)</td>
<td>8.1</td>
<td>7.5</td>
<td>7.1</td>
<td>5.8</td>
</tr>
</tbody>
</table>

* UCLA 10-point hip scale

### Table II. The parameters of wear (mean ± SD) in surface replacements and conventional acetabular cups calculated from measurements of the retrieved components

<table>
<thead>
<tr>
<th>Surface replacement (n = 20)</th>
<th>Conventional cups (n = 40)</th>
<th>Total</th>
</tr>
</thead>
<tbody>
<tr>
<td>Linear wear (mm)</td>
<td>1.98 ± 0.92</td>
<td>2.24 ± 1.36</td>
</tr>
<tr>
<td>Linear wear rate (mm/yr)</td>
<td>0.375 ± 0.195</td>
<td>0.216 ± 0.126</td>
</tr>
<tr>
<td>Wear volume (mm³)</td>
<td>1726 ± 939</td>
<td>711 ± 424</td>
</tr>
<tr>
<td>Wear volume rate (mm³/yr)</td>
<td>313.5 ± 158.2</td>
<td>71.4 ± 46.5</td>
</tr>
<tr>
<td>Weight-bearing direction*</td>
<td>90.5 ± 14.9</td>
<td>81.9 ± 18.0</td>
</tr>
</tbody>
</table>

* degrees from the horizontal

![Fig. 3](image-url) Scattergram to show the relationship between radiographic and true linear wear measurements. Inverted triangles represent the data reported by Wroblewski (1985).
torque of the larger prosthetic head, or to the use of titanium alloy for some of the prosthetic heads (Ma et al. 1983).

**Volumetric wear.** The volumetric wear rates are shown in Figure 4 as a function of the femoral head diameter. For conventional cups of size 22 mm, 26 mm, 28 mm, and 32 mm the rates averaged 25.9, 63.4, 75.6, and 88.7 mm³/yr respectively. Linear regression showed a significant tendency (p = 0.015) for higher volumetric wear rates with larger femoral heads.

![Graph showing volumetric wear rates as a function of femoral head diameter.](image)

**Fig. 4**

Volumetric wear rates as a function of femoral head diameter.

**Wear direction.** The average direction of wear, the angle β in Figure 2, was 36.9° (0.14 to 69.4). The direction of wear is mainly influenced by the orientation of the acetabular component and in no case was it perpendicular to the mouth of the component. The average weight-bearing axis was 90.5° ± 14.9° for the surface-replacement components and 81.9° ± 18.0° for the conventional designs. In most cases the direction of wear was lateral to the vertical; in only 17 cases (28%) was it medial to the vertical.

The weight-bearing axis was significantly associated with linear and volumetric wear in the surface-replacement components (p < 0.04) but not in the conventional cups.

**Correlation with functional rating.** No significant relationships were observed between any of the measured or calculated wear parameters and pain, walking or function scores, or between the wear parameters and sex, weight or age, although there was a tendency for elderly patients to have the lowest rates. When the activity score was combined with component diameter in the regression model a much greater relationship to volumetric wear emerged than when either was used alone (R² = 0.56, p < 0.01) indicating that added activity does increase component wear. When age was included in the model there was no further change in this relationship, perhaps because many of the patients in this study maintained an active life-style in spite of advancing age.

**DISCUSSION**

There is increasing evidence that polyethylene debris is responsible for acceleration of the process of prosthetic loosening, and is the cause of osteolysis (Schmalzried et al. 1992). Controversy continues regarding the amount, size and shape of the particles which are the most detrimental, and an estimate of actual wear rates is of importance. Our study is the first to determine the relationship between radiographic and true wear measurements in a large sample and for the complete range of component sizes used in total hip replacement. Of particular significance is the mathematical formula for computing the volume of material lost. Previous estimates have generally depended upon an oversimplified formula, the cross-sectional area of the femoral head times the linear wear distance measured from radiographs. Such a calculation overestimates the volume of material by a factor of about two. Our estimates of volumetric wear rates are similar in magnitude to those that have been reported in wear simulation studies.

Most of the components which we studied had articular surfaces which were highly polished, with a well-formed demarcation between the worn and the original articular surfaces. The absence of gross surface defects suggests that most of the wear particles are microscopic in size and that wear is not the result of discrete fragments becoming dislodged from the surface.
If we assume that an average debris particle is \(1\mu m \times 1\mu m \times 10\mu m\), then the wear rates that we have observed in conventional components would produce \(35.4 \times 10^{18}\) to \(77.2 \times 10^{18}\) particles over a period of ten years.

The significantly greater volumetric wear of the surface-replacement components was probably the result of their larger diameter. Although contact pressures decrease with larger head diameters, the frictional torques increase substantially (Ma et al 1983) and those torques between the femoral head and the polyethylene cup are the most significant factor in the generation of polyethylene wear debris.

**Conclusions.** Radiographic measurements of linear wear give reasonable estimates of the true linear wear, although they slightly underestimate it. Volumetric wear, as correctly determined from retrieved specimens (or possibly from radiographic measurements), is substantially less than has been previously reported. Large femoral-head diameters and thin polyethylene shells result in high volumetric wear rates.

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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**


