POLYETHYLENE WEAR FROM RETRIEVED ACETABULAR CUPS

B. WEIGHTMAN, S. A. V. SWANSON, G. H. ISAAC, B. M. WROBLEWSKI

From the Imperial College, London and Wrightington Hospital, Wigan

Laboratory wear testing of ultra high molecular weight polyethylene from 12 Charnley acetabular cups, removed after periods of up to 17.5 years showed that the large patient-to-patient variations in clinical penetration rate cannot be explained by batch-to-batch variation in the wear resistance of the material. Nor was there any evidence of a time-dependent degradation in wear resistance of the material.

The use of the Charnley hip prosthesis in younger, more active patients has led to an increase in the number of revision operations, with component loosening and tissue reaction to wear debris being two of the most common causes of failure (Isaac et al 1986). Furthermore, femoral head penetration into the acetabular cup in revised prostheses is found to be three times that of joints which are still in service (Isaac et al 1987) suggesting a causal relationship between wear and loosening. Although the detailed mechanism of loosening is as yet unclear, there is evidence that the accumulation of polyethylene wear debris at bone-cement interfaces produces osteoclasia and secondary fibroplasia (Black 1978; Revell et al 1978; Mirra, Marder and Amstutz 1982). In addition, excessive penetration can reduce the range of motion of a hip prosthesis to such an extent that impingement of the neck of the femoral component against the rim of the acetabular cup can produce increased stresses in the cement fixation (Weightman 1977; Isaac et al 1990).

Whatever the detailed mechanism of failure, the recognition of polyethylene wear as a significant factor indicates the need for continued study of the extremely complex phenomenon of prosthetic wear. In particular, researchers must explain why polyethylene wears at a much higher rate in some patients than in others, and also provide reassurance about the long-term stability of ultra high molecular weight polyethylene in the human body, so that surgeons can be confident about extending the treatment to younger patients.

The clinical wear rate of polyethylene has been monitored at Wrightington since the material was first introduced by Charnley in 1962 and has always shown a wide patient-to-patient variation (Charnley and Cupic 1973; Charnley and Halley 1975; Griffith et al 1978). The most recent study (Isaac et al 1990) found the penetration rate in 59 retrieved Charnley cups to vary between 0.005 mm per year and 0.6 mm per year; that is, by a factor of more than 100. In none of these studies was it possible to correlate penetration rate with patient weight, patient activity, or time of implantation. More recent work (Atkinson et al 1985a,b; Isaac et al 1990) has concentrated on the effect of femoral head surface finish and the possible detrimental effect of cement particles becoming embedded in the bearing surface of the cups, but again no direct correlation has been established.

In the absence of a convincing explanation of the variation in penetration rates, the primary aim of the present work was to study another possible factor, namely batch-to-batch variation in the wear resistance of polyethylene. Charnley and Cupic (1973) were the first to suggest this as a possible explanation, but, probably due to the practical difficulties involved, the hypothesis has not yet been tested.

A secondary aim was to investigate the possibility of degradation of polyethylene in the body. The topic has been discussed in the past, but with the extension of joint replacement to younger patients, it becomes even more important to know if the material will experience an accelerating wear rate due to degradation with time.

Rose et al (1977) found that the usual dose of ionising radiation, used during sterilisation of the implant,
generated a significant quantity of low molecular weight polyethylene which is known to suffer certain types of environmental embrittlement and degradation. On the other hand, in a later study Rose et al (1980) found no increase in the wear rate of two Charnley prostheses during 60-week runs in a wear-simulator, although considerable discoloration, indicative of oxidation degradation, was observed.

Dowling et al (1978) suggested that fluctuating stress in the presence of body fluids may produce stress corrosion or corrosion fatigue in ultra high molecular weight polyethylene, and Weightman, Isherwood and Swanson (1979) came to a similar conclusion after examining a fractured polyethylene tibial component.

Roe et al (1981) studied the effects of ageing of irradiated polyethylene in both serum and argon environments and concluded that a combination of increased density, higher stiffness and reduced molecular weight might eventually lead to brittle fatigue failure.

The published clinical data on the wear rate of polyethylene with time suggests a decreasing rather than an increasing wear rate. In one study, for example, the average wear rate during the second five-year period was found to be approximately 40% lower than during the first five years (Charnley and Halley 1975). However, since the measurements of 'wear' (that is, of penetration) included polyethylene creep, which has been shown to account for between 70% and 99% of early dimensional changes (Rose et al 1980), and to decrease with time (Nairn 1987), these findings can be criticised on the ground that any changes in the true wear rate (the rate of production of wear debris) may have been obscured.

We now report the results of laboratory wear tests on specimens prepared from acetabular cups retrieved after periods of implantation. The tests were designed to study the relationship between possible batch-to-batch variations and clinical wear rates, and to measure any reduction in wear resistance of the material with increasing time of implantation.

MATERIALS AND METHODS

A total of 12 Charnley acetabular cups from the Wrightington collection of retrieved prostheses were used. Measurements of the penetration of the femoral heads into the cups had been made using a shadowgraph technique on casts of the articular surface, but these were not made available until the laboratory wear testing had been completed. The cups were divided into three categories of clinical penetration rate: low (less than 0.05 mm per year), medium (0.1 to 0.2 mm per year) and high (greater than 0.5 mm per year), and there was a wide range of implantation times (10 months to 17.5 years; mean 8.5 years). The mean age at arthroplasty was 59 years (range 49 to 73).

As part of earlier studies at Wrightington, the cups had been soaked in dichloromethane to remove any loose organic material and to dissolve the acrylic cement still attached to the cups.

Cylindrical polyethylene specimens, 9 mm in diameter, were trepanned from the cups and then wear tested. A special jig was made to hold the cups so that trepanning could be carried out in a radial direction. This left a concave portion of the articular surface of the cup normal to the axis of the specimen. In an attempt to separate the effects on wear resistance of degradation and fatigue, two specimens were prepared from each cup, one from the worn region and one from the unworn region; the former may have experienced degradation and fatigue, the latter degradation only.

The six-station pin-on-plate wear screening machine, one station of which is shown schematically in Figure 1, was based on the machine developed by McKellop and Clarke (McKellop et al 1978; McKellop et al 1981). When mounted in the machine, each specimen was loaded by a pneumatic cylinder against a flat metal counterface with a constant axial load of 223 N.

Individual wear chambers were clamped to an oscillating table which was driven through a 25 mm stroke at 60 cycles per minute (maximum sliding speed 78.5 mm/s). The chambers were made of Perspex and each contained 25 ml of bovine serum (Koch-Light Laboratories Ltd) plus 3 ml of 1% sodium azide solution to inhibit bacteriological degradation. Each chamber was connected to a reservoir of de-ionised water through
siphon tubes which automatically compensated for evaporation.

Surgical grade (316S12) stainless steel counterface discs were machined from 51 mm diameter bar stock. Each disc was ground on 220- and then 600-grit silicon carbide paper, and then polished with 6 μm followed by 1 μm diamond lapping compound. Finally they were cleaned and degreased ultrasonically and passivated in 30% nitric acid. The surface finish was typical of the most highly polished prosthetic components at better than 0.025 μm.

Wear was measured by loss in weight of the polymer specimens. Each group of specimens had its own control which was mounted in a holder in a bath of bovine serum but was neither loaded nor worn. Before starting the tests, and then at approximately 0.5 million cycle intervals, the specimens were washed, rinsed, dried with alcohol, and weighed. Any gain (or loss) in weight of the control specimen was added to (or subtracted from) the apparent loss in weight of the wear specimens to correct for fluid absorption. The wear chambers and counterfaces were cleaned and fresh serum added after each weighing.

The overall wear rates, in milligrams per million cycles, were taken as the slopes of the best-fit straight lines through the wear data, calculated by the method of least mean squares regression. These were then converted to wear coefficients (mm³/Nm) using the known density of the material, the load on the specimens, and the stroke of the machine.

RESULTS AND DISCUSSION

Figure 2 shows the raw wear data for specimens from one acetabular cup and is representative of most of the groups. Table I lists the wear coefficients calculated from the slopes of the individual wear graphs, together with the period of implantation, the weight of the patient and the clinical penetration rate.

In Figure 3, the laboratory wear coefficients are plotted against the clinical penetration rate. The low, medium and high penetration rate grouping of the selected cups is clear. It is also clear that the penetration rate of the cups is unrelated to the wear resistance of the polyethylene. The large variation in clinical penetration rates cannot be explained by variation in the wear resistance of the material of which they are made.

Two of the 24 tested specimens exhibited wear coefficients approximately one order of magnitude higher than those of the others, and no satisfactory explanation could be found for these results. These specimens did not come from cups which had exhibited high penetration rates in vivo, and although they both came from a worn region, the other specimens from the same cups did not show similarly high wear. On both counts, therefore, it seems unlikely that the bulk material was defective.

One possible explanation is that cement particles had become embedded in the cup surfaces in vivo and had not been removed by the soaking in dichloromethane prior to laboratory testing. Such particles might have produced an increased polyethylene wear rate by damaging the stainless steel counterfaces in the wear testing machine. The individual plots for the two specimens (one of which is shown in Figure 4) give some support for this view by showing increasing wear rates during the first 3 million cycles, but examination of the counterfaces after
Raw data for cup No. 5, one of the two acetabular cups (Nos 5 and 10) which exhibited accelerated wear in the specimen from the worn region.

Laboratory wear coefficient plotted against implantation time for specimens from 12 retrieved acetabular cups.

### Table I. Wear coefficients for specimens from 12 retrieved polyethylene cups, with period of implantation, patient weight and penetration rate

<table>
<thead>
<tr>
<th>Cup</th>
<th>Implantation time (yr)</th>
<th>Body mass (kg)</th>
<th>Penetration rate (mm/yr)</th>
<th>Specimen</th>
<th>Wear coefficient ($\times 10^{-7}$mm$^2$/Nm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>worn</td>
<td>0.049</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td>unworn</td>
<td>0.123</td>
</tr>
<tr>
<td>1</td>
<td>0.83</td>
<td>54.0</td>
<td>0.181</td>
<td></td>
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<tr>
<td>2</td>
<td>1.38</td>
<td>81.7</td>
<td>0.522</td>
<td></td>
<td></td>
</tr>
<tr>
<td>3</td>
<td>3.01</td>
<td>79.4</td>
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<tr>
<td>4</td>
<td>5.81</td>
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<td>5</td>
<td>6.26</td>
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<td>0.035</td>
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<td>6</td>
<td>6.52</td>
<td>60.8</td>
<td>0.575</td>
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<td>7</td>
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<td>8</td>
<td>10.20</td>
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<td>0.020</td>
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<td>9</td>
<td>12.19</td>
<td>70.0</td>
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<tr>
<td>10</td>
<td>14.58</td>
<td>74.8</td>
<td>0.123</td>
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<tr>
<td>11</td>
<td>16.35</td>
<td>52.2</td>
<td>0.193</td>
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<tr>
<td>12</td>
<td>17.5</td>
<td>82.6</td>
<td>0.029</td>
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</table>
the tests did not reveal undue damage. Furthermore, reference to the results of the examination of the two cups after their retrieval showed that while one of the articular surfaces contained cement particles and pits, the other did not.

Figure 5 shows quite clearly that the wear resistance of the polyethylene was not reduced by implantation in the human body for periods up to 18 years. Apart from the two specimens discussed above, the results are in general agreement with those from previous studies of un-implanted polyethylene on the same machine, which showed a mean wear coefficient of 0.107 × 10⁻⁷ mm²/Nm (Weightman and Light 1985). A more thorough analysis of the data, however, shows an interesting difference between material from the worn and from the unworn regions of the implanted cups. The mean wear coefficient of specimens from the worn regions (0.044 × 10⁻⁷ mm²/Nm) is significantly lower than that of un-implanted material, at the 99% confidence level, and significantly lower than that of the unworn region of implanted material (0.101 × 10⁻⁷ mm²/Nm) at the 95% confidence level. However, the wear coefficient from the unworn regions of implanted material is not significantly different from the un-implanted material, even at the 80% confidence level. These findings indicate that the difference between the worn and unworn regions is not due to a reduction in wear resistance in the unworn region, but rather to an increase in wear resistance in the worn region.

It seems likely that these differences in wear resistance are the result of different surface finishes of the polyethylene. That is, both the un-implanted and the unworn but implanted material had a machined surface finish, whereas the surface in the worn regions of the cups had been burnished by the wear process. Although this finding suggests that acetabular cups should be manufactured with as fine a surface finish as possible, it should be stressed that the improvement in wear resistance so gained would be very small compared with the variation in observed clinical penetration rates. It also appears that this improvement is achieved by natural processes soon after the implantation of cups with machined surface finishes.

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


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