

# Wear Mechanisms for Ultrahigh Molecular Weight Polyethylene in the Total Hip Prosthesis

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## Synopsis

Ultrahigh molecular weight polyethylene is now extensively used in total joint replacements, and there is ample clinical evidence that this material wears at a significant rate in such applications. Six commercially available total hip replacements were subjected to an accurate simulation of clinical mechanical and chemical conditions, and the wear behavior of the polyethylene components was analyzed by friction coefficient measurements and scanning electron micrography of the wear surfaces. The results indicate that all of the proposed wear mechanisms excepting adhesive and abrasive wear may be eliminated. It is highly likely that abrasion fatigue and related phenomena are central to the wear process in many prostheses, but the wear process may be ultimately determined in many cases by the type of defects present in the manufactured material.

## INTRODUCTION

Total joint prostheses are commonly manufactured with a metal component articulating against a component made of ultrahigh molecular weight polyethylene (to be referred to as UHMWPE). There is general agreement that the debris generated by wearing of the UHMWPE component is physiologically inflammatory and that sufficiently large rates of generation of debris will have serious clinical consequences.<sup>1</sup> In addition, wear and plastic deformation of the UHMWPE component may lead to loss of mechanical function or outright failure of the prosthesis. Generally, the mechanisms of wear of such prostheses are inferred from the surface morphology of devices that have been removed after some period of service. Figures 1 and 2 are typical of the observed surface morphology.

The numerous mechanisms for clinical wear of UHMWPE suggested in the literature include adhesive wear, abrasive wear, wear by roll formation, penetration wear, fatigue wear, brittle fracture wear, and wear by delamination.<sup>2-6</sup> However, none of these mechanisms has been evaluated with regard to the physical or mechanical properties of the UHMWPE and/or counterface that these mechanisms require. It is our intention to do so here.

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Fig. 1. Typical UHMWPE articular surface as seen in the scanning electron microscope. This prosthesis has been implanted and used ca. eight months.

## EXPERIMENTAL

### Materials

The materials used here were six commercially available total hip prostheses of the Charnley-Mueller design, from different manufacturers. At the same time, six companion prostheses were purchased for molecular weight characterization.

UHMWPE as manufactured for the acetabular (cup) components of these hip prostheses has a nominal molecular weight in the  $(1-10) \times 10^6$  range. However, this can be radically altered by processing. Because it has been radiation sterilized, there is also a substantial crosslinked fraction which is insoluble under all conditions. Thus, complete molecular weight characterizations have not been available. The characterizations were performed in a separate study,<sup>7</sup> and the results will be briefly cited here. The low molecular weight fraction, with  $\bar{M}_v$  ca. 9000, usually is between 10% and 25%, depending on the manufacturer, but can run as high as 75% at the surface. Total soluble fractions can run from 10% to 100% (again, depending on the manufacturer) with average ( $M_w$ ) molecular weights from 50,000 to  $10^6$ . Thus, there is considerable variation in molecular weight distribution. As will be seen below, this did not influence the wear mechanism, although it may have a role in determining the wear rate.

Another concurrent study<sup>8</sup> of importance here is that of the overall effects of the radiation sterilization procedure; these may be summarized as extensive molecular crosslinking (see above) and oxidative degradation. The latter is



Fig. 2. Another typical clinical wear surface, also after eight months of service.

essentially the formation of hydrophilic carbonyl side groups which lead to considerable weight gain due to water absorption. Consequently, it is not possible to quantitatively characterize wear, under simulated (or real) clinical conditions, by measurement of weight loss. Alternatively, recovery of wear debris from synovial fluid or bovine serum is difficult, and only recently have such quantitative recovery techniques become available.

Several other measurements were made in order to characterize the UHMWPE. Density was measured in an isopropanol–water density gradient column (ASTM-D 1505). The density is 0.943 as received, but rises to  $0.946 \pm 2$  due to fluid absorption. Hardness, or the Shore D (ASTM 2240) scale, was  $69.3 \pm 0.2$ . Tensile tests could not be performed directly on the cups but were done on companion materials (Dixon Pennlon Lot #8045-R) purchased for that purpose. The tensile properties can be summarized as follows: The yield stress and strain are 3250–3350 psi and 0.15–0.19, respectively; the lower yield stress and strain are 2600–2900 and 0.67–0.87, respectively. Reductions in area were ca. 75%. There is some indication that immersion in serum or water increases the yield stress 100–200 psi and decreases yield strain a few percent.

Spectrometric analyses revealed traces of some metallic elements in the UHMWPE. A few hundred ppm of Al, Ca, Si, and Ti were found, presumably as residues from the catalyst and other incidental additives.

The crystallinity and crystal morphology of this material have been studied<sup>8</sup> by small-angle and wide-angle x-ray scattering and differential scanning calorimetry. The general picture which emerges is that of a highly inhibited crystallization; the structure is ca. 60% crystalline, with no uniformity of crystallite size or form, and also a few percent of voids.

TABLE I  
Friction Coefficients in Simulated Hips

Prosthesis no.	Friction Coefficient		
	Initial	500 Hr	1000 Hr
1	0.0803	0.0803	0.0724
2	0.107	0.0892	0.0982
3	0.107	0.0892	0.134
4	0.0803	0.0892	0.134
5	0.0447	0.0447	0.0268
6	0.0982	0.0447	0.0357
	Average	0.0863	0.0728

### Methods

The mechanical and chemical environment of the human hip joint was reproduced by the total hip simulators of the M.I.T.-Harvard Joint Physiology Group. These simulators have been used for comparative evaluations of total hip prostheses and have been previously described in detail in the literature.<sup>9,10</sup> These were used because the friction and wear of UHMWPE run dry or in water or other simple media are quite different from the corresponding behavior in physiologic media, e.g., synovial fluid or serum. In addition, the *in vivo* stress distribution should be reproduced reasonably well by these simulators. The purpose of the simulations was to obtain accurate friction coefficients and to generate wear topographies which could be conveniently analyzed in the scanning electron microscope. The methods used for the friction measurements have been previously described.<sup>9,10</sup> They were made prior to simulation and after 500 and 1000 hr; 1000 hr in the simulator is equivalent to one year of normal gait. Scanning electron micrography was performed directly on the plastic (acetabular) cups afterwards.

## RESULTS AND DISCUSSION

The friction coefficient measurements are reported in Table I. The wear topography will be reported where appropriate in the discussion below; it is our intention to examine each of the mechanisms of wear separately with regard to their applicability for UHMWPE as encountered in total joint prostheses. Also, another mechanism of wear will be suggested that does have broad applicability in prosthetic devices.

### Adhesive Wear

Evidence of a transfer film should be sufficient to include adhesive wear as a possible mechanism. Several authors have noted a transfer film in the presence of bovine serum or bovine synovial fluid.<sup>11,12</sup>

The classical theory of friction and wear<sup>13</sup> relates the friction force to adhesion and the shear strength at or near the interface. For polymers an additional term is used to account for energy dissipation by deformation so that Bowden and Tabor's equation for the friction force  $\mu_D$  becomes

$$\mu_D = \frac{9\pi(1-\nu^2)P}{32E}(\alpha) \quad (1)$$



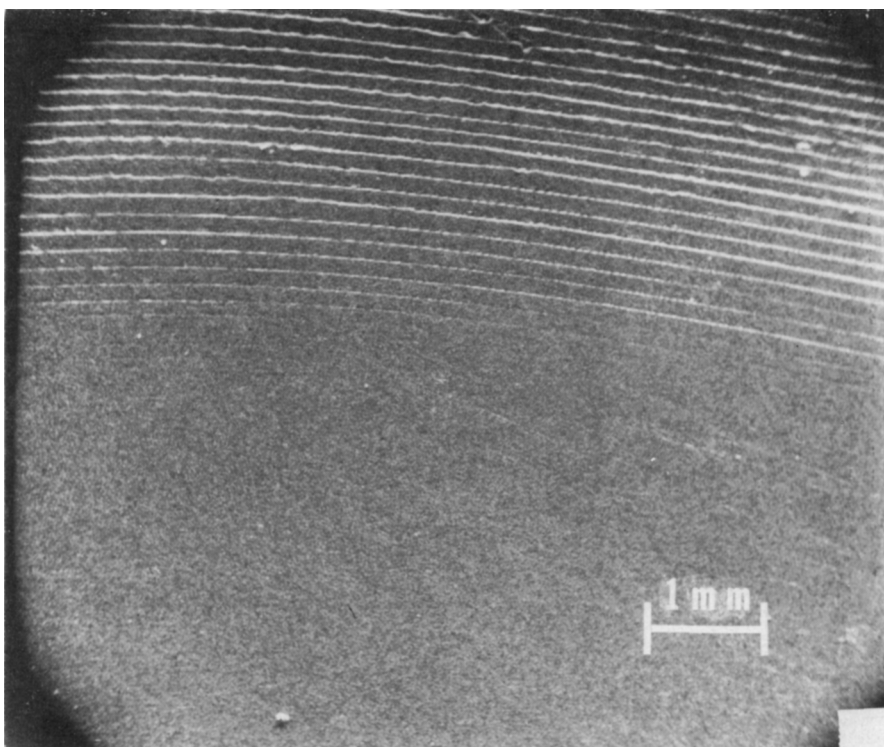


Fig. 3. Smoothing out of machine markings on prosthesis #2 (see Table I) by 1000 hr (one year simulated) of wear.

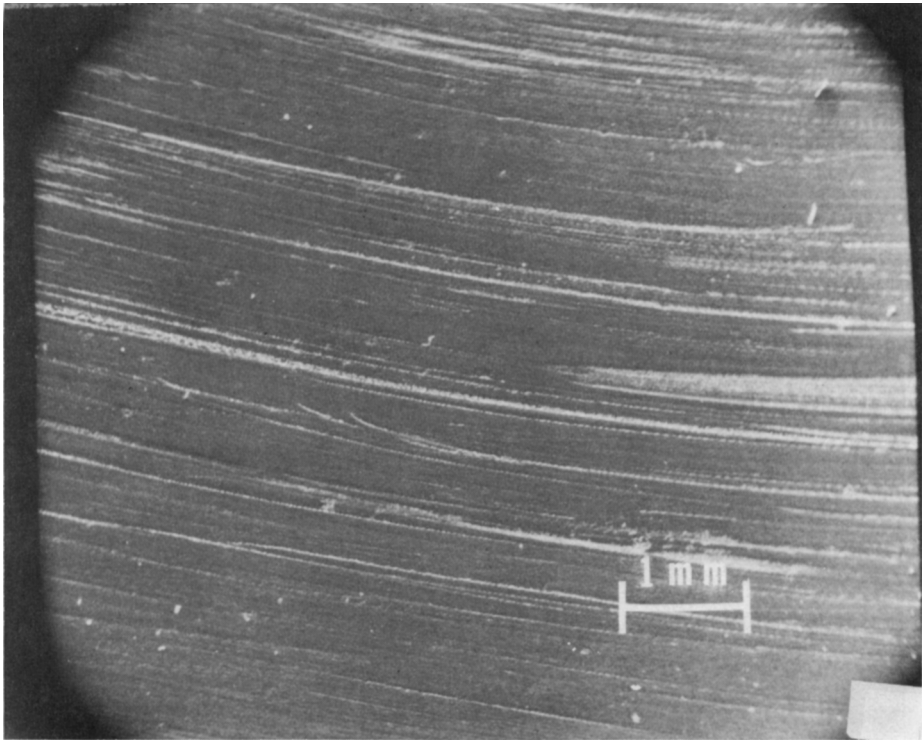
where  $\nu$  is Poisson's ratio,  $E$  is Young's modulus,  $p$  is the mean pressure, and  $\alpha$  is the energy loss factor. We can estimate the latter as 0.5. The result is a calculated value of the fraction coefficient of 0.06. Since the experimentally observed values (for total joint prostheses) range from 0.03 to 0.13, as shown in Table I, with 0.05 being typical of physiologic (synovial fluid) lubrication and 0.1 typical of dry running, adhesive wear is consistent with the observed friction coefficients.

The wear particle size  $d$  in this case for adherent particles has been predicted by Rabinowicz<sup>14</sup> to be

$$d = 24\gamma_a E / \sigma_y^2 \quad (2)$$

where  $\sigma_y$  is the yield stress,  $E$  is Young's modulus, and  $\gamma_a$  is the effective surface energy. Studies of UHMWPE fracture<sup>15</sup> have determined  $\gamma_a$  as  $10^7$  ergs/cm<sup>2</sup> for the randomly oriented polymer and  $10^5$  ergs/cm<sup>2</sup> for the oriented polymer. Contact angle investigations<sup>16</sup> indicated that  $\gamma_a$  is 40 ergs/cm<sup>2</sup>. As eq. (2) is correct for processes in which elastic energy is converted to surface energy, the particle diameter for the oriented polymer is approximately 1.5 mm.

The polymer is most probably oriented at the articular surface; as Pooley has shown, the polymer surface becomes oriented with the initial pass along the direction of movement.<sup>17</sup> The type of wear particle formed was dependent upon the surface orientation and the direction of sliding. Also, two different types of particles are seen. The first is lumpy and is only seen with the orienting pass. The second are extremely thin discontinuous sheets of the order of 10 Å size which are maintained and do not become larger, i.e., no further transfer takes place. Also, microstructure appears to not be important.<sup>18</sup>



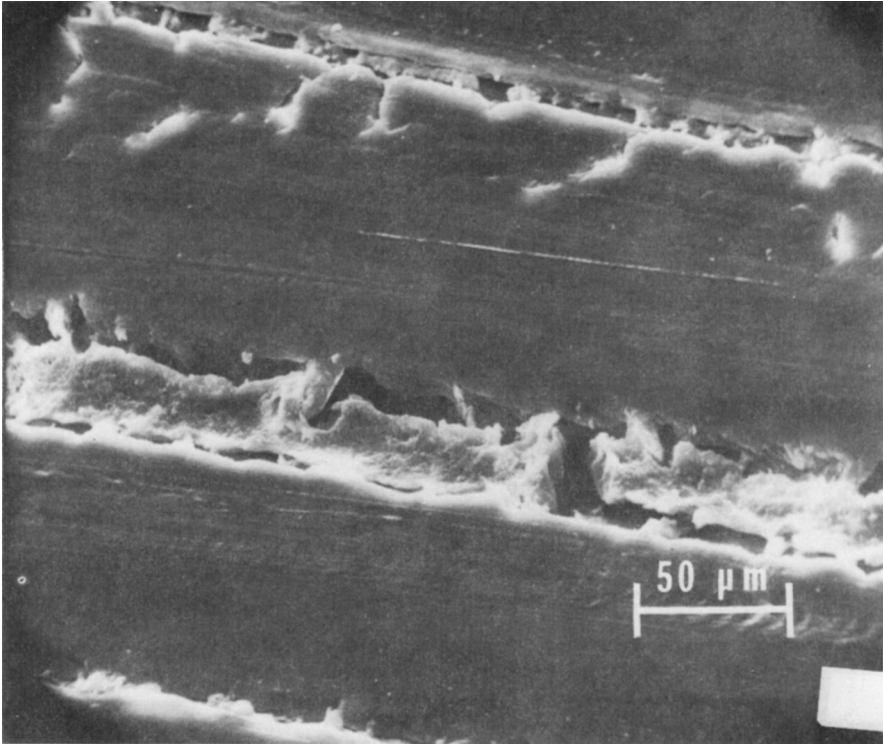
(a)

Fig. 4. Longitudinal scratches and tearing on prosthesis #6, indicating abrasive or three-body wear.

The wear particles found by Parkinson<sup>11</sup> and Shen and Dumbleton<sup>12</sup> are in fact of two types, and are in general agreement with Pooley's work. For a loose wear particle, the equation

$$d = 6EW_{ab}/v^2\sigma^2y \quad (3)$$

may be used. As the surface interaction energy between the metal and the UHMWPE particle is the relevant quantity here, the adhesion energy  $W_{ab} = \sigma_a + \sigma_b$  is determined by  $\sigma_a = 40$  ergs/cm<sup>2</sup>. According to eq. (3), the critical diameter for a loose wear particle is about 30 $\mu$ . Thus, if an adherent wear particle forms, it should be immediately released. In fact, no such large wear particles (1 mm) were observed, either in the serum or in the SEM micrographs of the articular surface. In one case the surface was actually smoothed, as shown in Figure 3. Other micrographs (see above and below) indicate much smaller particles. Thus, while it is probable that some adhesive wear occurs (perhaps in the initial orienting passes of the femoral head), it should not be one of the dominant mechanisms of wear in this system, especially under the usual lubricated condition.



(b)

Fig. 4 (Continued from previous page.)

### **Abrasive Wear**

Longitudinal (to the direction of motion) grooves are generally considered to be an indication of two-body or three-body abrasive wear. The "third body" may be acrylic debris or even metallic debris. In any case, as Figure 4 shows, there are longitudinal scratches which involve tearing or removal of UHMWPE.

### **Abrasive Fatigue**

This mechanism, which we have so named in order to distinguish it from either abrasion or fatigue, is seen in the wearing of rubber; it is shown schematically in Figure 5, as first suggested by Thomas<sup>19</sup> after the work of Rivlin and Thomas. The machine markings (or other irregularities) on the articular surface could serve as the necessary "trouser legs." Large adhesive forces are not necessary to this mechanism; the crack grows incrementally with each cycle. Figures 6 and 7 show cracks transverse to the sliding direction, and Figure 7 shows fibrils within the crack. This surface morphology is consistent with the abrasion fatigue mechanism. Figure 8 shows the surface of prosthesis #3, in which machine marks transverse to the sliding direction have apparently generated a series of cracks. This phenomenon was noted in clinical specimens as well.

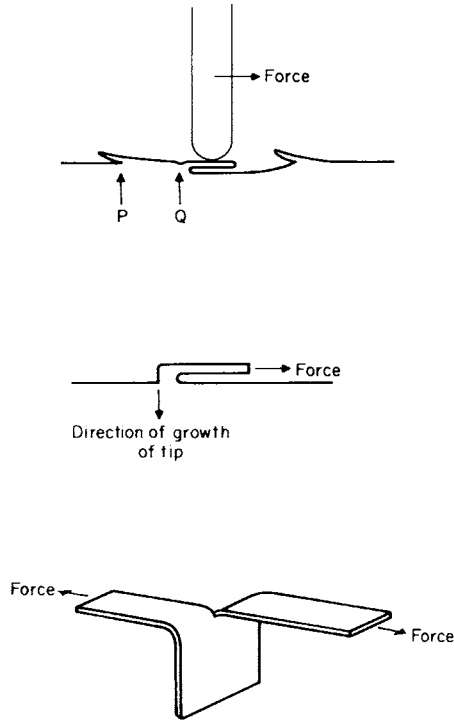


Fig. 5. Schematic description of the abrasion fatigue wear mechanism as proposed by Thomas.<sup>19</sup>

### Fatigue Wear

The response of ductile semicrystalline polymers to cyclic loading is to soften; the rate of softening increases with strain amplitude, and no softening occurs at sufficiently low strains.<sup>20</sup> This should be true for UHMWPE as well. The failure mechanism undoubtedly involves cumulative damage by inelastic processes. The characteristic features of surface fatigue wear are similar to those observed in bulk specimen fatigue tests. In general, it is expected that (1) fatigue wear particles tend to be larger than adhesive wear particles; (2) surface-active lubricants should increase wear; and (3) service life is markedly load dependent, generally being proportional to the inverse cube of the load.

Since the maximum elastic stress  $\sigma_{max}$  for spherical contact varies with the cube root of the load, the time to failure is given by

$$t \propto \sigma_{max}^{-2}$$

Thus, although cumulative damage must be involved, we must conclude that the "classical" fatigue wear mechanism<sup>12</sup> is not operative in this case.

### Brittle Fracture Wear

The mechanism of brittle fracture wear is now well known<sup>14</sup>: the maximum tensile stress behind the junction is about one third the maximum compressive stress under the junction. If the material is sufficiently brittle that the tensile strength is less than one third the compressive yield strength, then tensile cracks

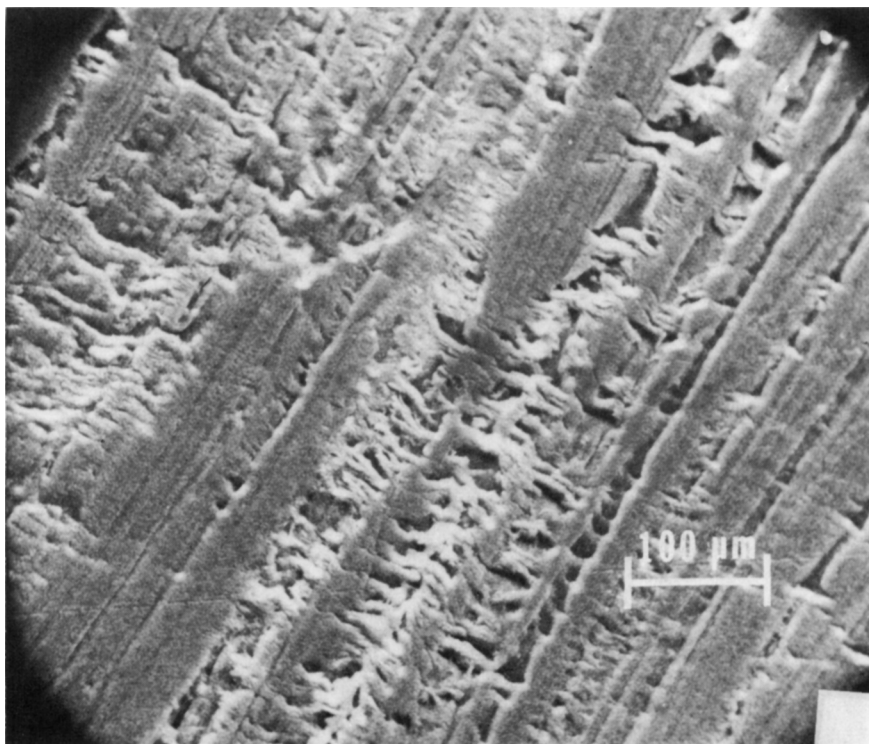


Fig. 6. Prosthesis #1 after 1000 hr, showing deformation and cracking.

will occur behind the contact, transverse to the direction of motion. Figures 6–8 do suggest the presence of such cracks. However, the tensile strength and compressive yield strength of UHMWPE are approximately equal.<sup>21,22</sup> Also, brittle fracture wear is expected to assume catastrophic rates very early in the wear process. Finally, UHMWPE exhibits very ductile behavior, except possibly at cryogenic temperatures. Although this mechanism has been proposed for the wear of total hip prostheses,<sup>10</sup> we conclude that the observed surface morphology only resembles brittle fracture wear or that the surface has been locally embrittled by, for example, a chemical or mechanicochemical effect. One example of the latter is environmental stress cracking (ESC). However, UHMWPE is not expected to be sensitive to ESC.

### Roll Formation

According to Aharoni<sup>23</sup> there are two general prerequisites for wear by roll formation: (1) at least one of the two sliding surfaces must be an elastic or ductile material, possessing the ability to undergo bulk deformation, and (2) “give” under the normal pressure must be not too low or too high. Under too low a pressure (3–5 kg/cm<sup>2</sup>), the wear, especially that of ductile polymers, seems to be produced by one surface ploughing into the other, with the debris in the shape of ill-formed chips and slivers. Under too high a pressure ( $\geq 20$  kg/cm<sup>2</sup>), shearing of one of the two surfaces occurs; and if a few rolls do form, they shear and smear. Shear of the surface yields a layer of material oriented in the direction of motion. (This was found by Pooley with regard to adhesive wear.<sup>17,18</sup>)

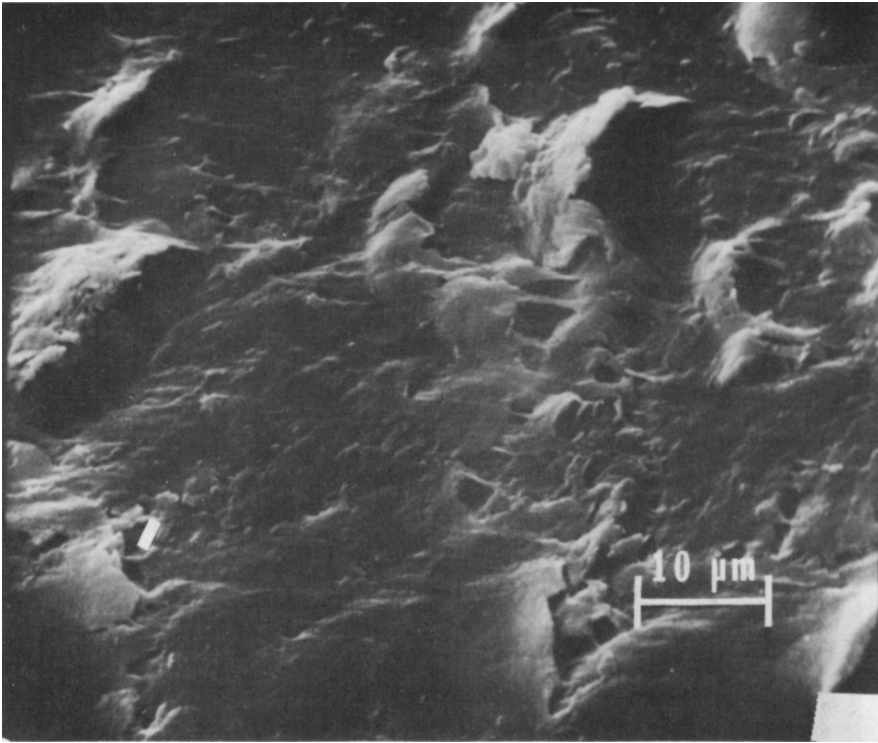


Fig. 7. Prosthesis #6 also showing cracks transverse to the sliding direction. Note the fibrils in the cracks, at higher magnification.

The area of contact in the Mueller-type hip prostheses is less than the total surface area of the cup. Even if the peak load were distributed evenly over that overestimated area, the peak contact pressure would be approximately  $20 \text{ kg/cm}^2$ , which is too high for this mechanism of wear to be significant.

### Penetrative Wear

The term penetrative wear is meant to classify all those situations where a locally harder surface penetrates and wears a locally softer surface. This would include classic abrasion of a soft material by abrasive paper, but only in those cases where the load on the individual grit is sufficient to cause penetration. This definition would also include wear resulting from local variations in hardness due to work hardening, impurities, or phase differences which cause penetration on contact.

The mechanism formulated accounts for the formation of plate-like wear debris found in penetrative wear. These particles form in a three-part process. First, as sliding proceeds, a bulge forms ahead of the slider. Second, a shear crack initiates at the surface and propagates into the substrate parallel to the slider face. Third, the platelet is pushed up the face of the slider and attaches to the next particle.<sup>24</sup>

The mechanisms of abrasive wear have been discussed previously. The second stage of this particular model for penetrative wear produces plate-like particles. This requires a shear crack to initiate at the surface and propagate into the substrate parallel to the slider face. Neither the cracks described nor the wear

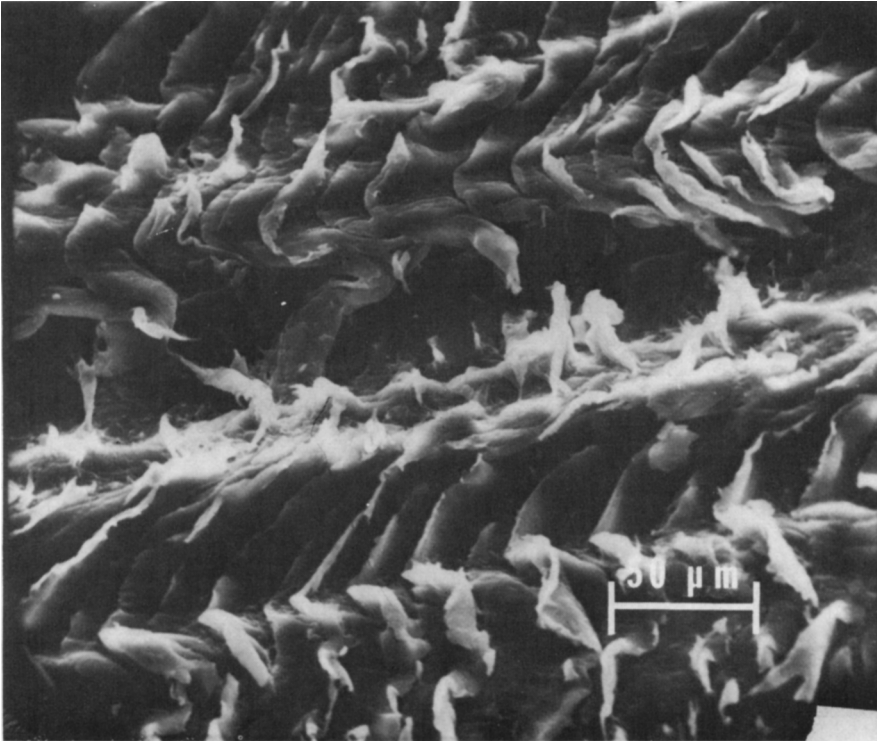


Fig. 8. Transverse cracks which have originated in machine marks; prosthesis #3.

debris found for this mechanism have been observed for PE. Parkinson saw no cracks in the deformed surface layer.<sup>11</sup> Further, high surface traction and low elongation to fracture are required for the formation of these cracks. PE does not meet these requirements; therefore, this mechanism also does not operate.

### Delamination Theory of Wear

There are six steps to the production of long thin sheet debris for this mechanism. They are as follows<sup>25</sup>:

1. Asperity deformation and removal with concurrent subsurface deformation.
2. This generates a surface (in metals) free of a coherent oxide layer which allows the image force to drive out dislocations near the surface to the surface. This dislocation instability leads to a flow stress gradient from the surface to the interior where there is again a dislocation concentration, i.e., a soft surface layer is produced.
3. Penetration and ploughing in addition to adhesion cause high surface traction forces.
4. Coalescence of dislocations forming voids occurs below the surface.
5. Coalescence of voids produces subsurface cracks.
6. Finally, long thin sheet debris is produced.

The existence of dislocations, while seen in single crystals of PE at very low

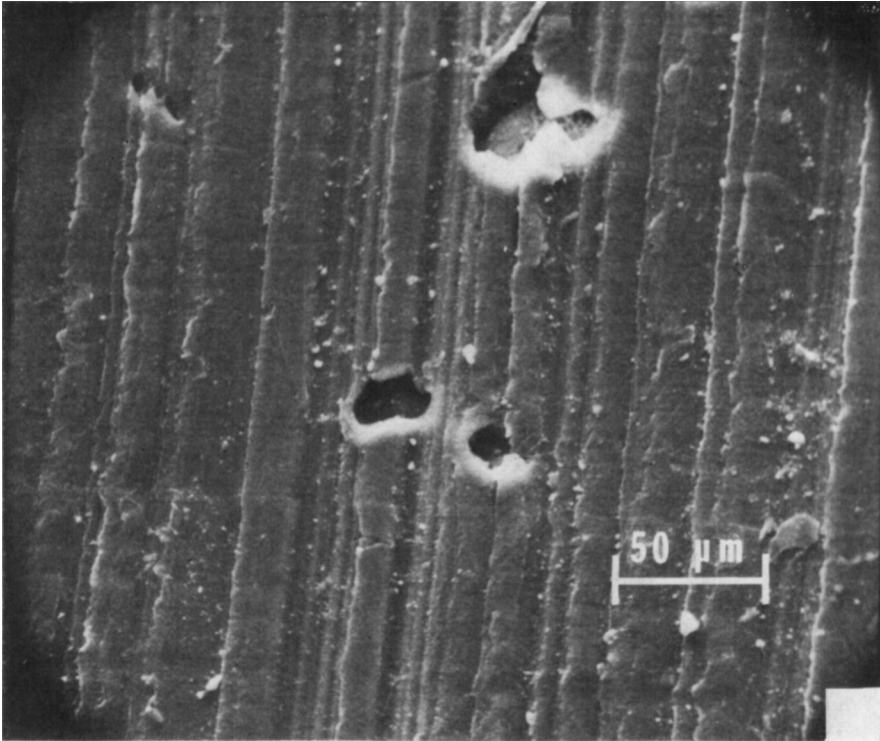


Fig. 9. Typical surface defects in prosthesis #3, as received: machine markings and porosity.

deformations, plays a questionable role in UHMWPE, especially in material that is only 60% crystalline. However, even if voids are present as a result of the processing difficulties encountered with UHMWPE, surface traction forces are too small to cause subsurface cracks to form.

Further, Parkinson<sup>11</sup> noted the presence of voids as a result of processing in solid-phase-formed UHMWPE and concluded that they were not effective in increasing wear. Thus, we conclude that this mechanism is also not operable in UHMWPE.

### Surface Artifacts

In the literature of the wear of artificial joints, the methods used to determine the mechanism of wear has been to visually examine the surface morphology with the SEM and assign the wear mechanism by comparison with surfaces produced by particular mechanisms.

One of the mechanisms cited is brittle fracture wear because of the cracks running transverse to the wear track. As is shown in the micrographs and accompanying figures, for example, in Figure 8, at least some of these markings are due to surface artifacts as a result of machining during processing.

The machining can also be seen as a source of crack initiation and generation of some wear debris. (This is discussed in the earlier section on abrasion fatigue).



### Processing Defects

A final mechanism of wear is also artifactual; this is wear due to processing defects. In Figure 9, a micrograph of one as-received acetabular component, such defects are obvious.

As in the case of machine marks providing the "trouser legs" for the abrasion fatigue mechanism, processing defects may also cause one of the proposed mechanisms of wear to become dominant and operate.

### CONCLUSIONS

The friction data and wear topography eliminate many of the proposed wear mechanisms for UHMWPE in vivo. The remaining possibilities are adhesive wear, abrasive wear, and abrasion fatigue wear.

The initial condition of the surface influences the wear mechanism and the ultimate condition of the worn surface.

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### References

1. H. G. Willert and M. Semlitsch, *J. Biomed. Mater.*, **11**, 157 (1977).
2. H. Bentler and M. Lehmann, *Wear*, **33**, 337 (1975).
3. P. S. Trent and P. S. Walker, *Wear*, **36**, 175 (1976).
4. H. C. Amstutz, *J. Biomed. Mater. Res.*, **8**, 547 (1968).
5. B. B. Seedhom, D. Dowson, and V. Wright, *Wear*, **24**, 35 (1973).
6. J. H. Dumbleton and C. Shen, *Wear*, **37**, 279 (1976).
7. A. M. Crugnola, E. L. Radin, R. M. Rose, I. L. Paul, S. R. Siman, and M. B. Berry, *J. Appl. Polym. Sci.*, **20**, 809 (1976).
8. H. J. Nusbaum and R. M. Rose, *J. Biomed. Mater. Res.*, submitted.
9. B. Weightman, I. Paul, R. Rose, S. Simon, and E. Radin, *J. Biomech.*, **6**, 51 (1973).
10. B. Weightman, I. Paul, R. Rose, S. Simon, and E. Radin, *J. Biomech.*, **6**, 299 (1973).
11. R. W. Parkinson, Ph.D. Thesis, University of Leeds, 1973.
12. C. Shen and J. H. Dumbleton, *Wear*, **30** 369 (1974).
13. F. P. Bowden and D. Tabor, *The Friction and Lubrication of Solids*, Oxford at Clarendon Press, London 1964.
14. E. A. Rabinowicz, *Friction and Wear of Materials*, Wiley, New York, 1965.
15. I. Anderton and J. Treloar, *J. Mater. Sci.*, **6**, 562 (1971).
16. F. D. Petke and B. R. Ray, *J. Colloid Interface. Sci.*, **31**, 216 (1969).
17. C. M. Pooley and D. Tabor, *Nature Phys. Sci.*, **237**, 88 (1972).
18. C. M. Pooley and D. Tabor, *Proc. Royal Soc. London*, 251 (1972).
19. A. G. Thomas, *J. Polym. Sci. Symp.*, **No. 48**, 145 (1974).
20. Rabinowitz and Beardmore, *J. Mater. Sci.*, **9**, 81 (1974).
21. H. W. Starkweather, Jr., T. F. Jordan, and G. B. Dunnington, *Polym. Eng. Sci.*, **14**, 678 (1974).
22. Hoechst Co., *Hostalen*, June 1976.
23. S. M. Aharoni, *Wear*, **25**, 309 (1974).
24. T. R. Bates, K. C. Ludema, and W. A. Brainard, *Wear*, **30**, 365 (1974).
25. N. P. Suh, *The Delamination Theory of Wear*, Second Progress Report to Adv. Res. Agency DOD, 1975.

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