Polyethylene wear in unicondylar knee prostheses

106 retrieved Marmor. PCA. and St Georg tibial components compared

Gordon W. Blunn[1], Atul B. Joshi[2], Peter A. Lilley[1], Eckart Engelbrecht[3], Leif Ryd[4], Lars Lidgren[4], Kevin Hardinge[2], Elmer Nieder[3] and Peter S. Walker[1]

106 unicondylar knee replacement tibial components were retrieved and analyzed for the amount and type of polyethylene wear. Three different designs were retrieved which had essentially the same femorotibial conformity. Each design showed a characteristic failure pattern. The polyethylene of PCA tibial components showed serious delamination after only short durations, as a result of heat pressing. St Georg sledge prostheses showed some delamination after 4 years' duration due to sub-surface cracks which were initiated by fusion defects in the polyethylene; metal backing of the components did not affect delamination of this prosthesis. The Marmor designs showed the least wear, with shiny depressions and surface pitting; no delamination was observed in the Marmor prosthesis. Molecular weight determination by gel permeation chromatography and analysis of crystallinity using Fourier transformation infra-red spectroscopy demonstrated that St Georg polyethylene had higher molecular weight and crystallinity than Marmor polyethylene. In some of the components investigated, crystallinity and molecular weight of the polyethylene were reduced under the wear track when compared with the unworn polyethylene.

Since fusion defects may cause delamination of polyethylene we urge manufacturers to reduce the number of such defects.

[1]Institute of Orthopedics, University College London, Stanmore, Middlesex, U.K.; [2]The Centre for Hip Surgery, Wrightington Hospital, Wigan, U.K.; [3]EndoKlinik, Hamburg, Germany; [4]Department of Orthopedics, University Hospital at Lund, Lund, Sweden

Correspondence: Dr. G. W. Blunn, The Institute of Orthopedics (University College London), Royal National Orthopedic Hospital, Stanmore, Middlesex, HA7 4LP, United Kingdom. Tel +44–81 954 0268. Fax +44–81 420 6497

Submitted 92-03-24. Accepted 92-04-05

One of the factors that dictate the long-term success of total joint prostheses is the wear of articulating plastic, usually ultra-high molecular-weight polyethylene (UHMWPE) bearing surfaces. Severe polyethylene wear is associated more with tibial components of the knee than with acetabular components of the hip (Landy and Walker 1988). This is due to the differences in contact area and congruity between tibial and femoral components which determine the contact stresses. Previous studies have indicated that contact stresses in the knee often exceed the yield strength of UHMWPE (Walker and Hsieh 1977, Bartel et al. 1986, Walker 1988).

The main types of wear seen on retrieved knee prostheses include delamination caused by subsurface failure of the polyethylene, pitting of the surface, three-body abrasion due to cement particle entrapment, and adhesive wear (Hood et al. 1983, Landy and Walker 1988, Collier et al. 1991). In a relatively conforming design, surface damage was more often associated with entrapped acrylic particles than with other mechanisms (Hood et al. 1983), whereas low conformity designs were associated with delamination (Engh et al. 1992). Even for one particular design there was often a wide range of surface damage from cracking caused by delamination to shiny depressions caused by deformation and adhesive wear of the polyethylene (Landy and Walker 1988). This range of damage may be attributed to the different kinematic conditions occurring at the bearing surface, with excessive sliding leading to delamination wear, whereas rolling or cyclic loading at the same contact point resulted in minimal wear (Blunn et al. 1991).

There are great variations in the properties of polyethylene between different manufacturers and also different production runs (Landy and Walker 1988, Li and Howard 1990). Crystallinity and molecular weight are cited as the chief material properties which control in vitro wear rates (Rose and Radin 1990). However, these properties have not been correlated with wear on retrieved prostheses.

We have analyzed the wear on three different unicondylar knee prostheses, which are essentially similar with respect to the conformity between the femoral and tibial components. The first design, the St Georg Sledge, was first used in September 1969 (Engelbrecht 1971), the second design, the Marmor, was used from November 1972 (Marmor 1973), while the third design, the PCA Uni, was not introduced until 1983 (Lindstrand et al. 1988).

Material and methods

106 tibial components of the three designs were retrieved (Table 1). The reasons for their removal were variable but over 40 of the St Georg components were removed due to instability. Loosening of one or both of the components was the main cause of failure in the Marmor and PCA groups.

The retrieved components were photographed and assessed for wear by visual examination and low power light microscopy using a scoring system, modified from Hood et al. (1983). In our system the tibial component was divided into three areas with respect to anterior-posterior position. Each area was rated 0 (no wear) to 5 (most severe wear) for deformation, delamination, abrasion, scratching, cement entrapment and pitting. A separate cumulative score for wear type was calculated for each component.

Sub-surface failure and deformation of polyethylene were examined by cutting sections through the middle of the wear track on a microtome and examining the sections in a light microscope using transmitted or plain polarized light. In addition, the bearing surface and the sectioned face of the bulk component were examined in a JEOL 35C scanning electron microscope (SEM) after being sputter-coated with a layer of gold 20 nm thick.

The length of the wear track was measured using vernier callipers and the position was noted with respect to the middle of the tibial component. Wear tracks were measured on Marmor and St Georg prostheses.

Infra-red analysis

Polyethylene sections 170 μm thick were taken from the mid-wear track of six specimens (3 Marmor and 3 St Georg). These specimens had been in patients for over 8 years and were chosen because delamination of the surface was minimal. Sections were placed in a Fourier transform infra-red (FTIR) microspectrometer with an attached light microscope. This allowed alignment of the section and space-localized analysis using a beam diameter of 200 μm. Each spectrum was obtained with 500 scans. Crystallinity was measured according to Gueugnaut et al. (1988) by measuring the absorbance bands at wavelengths 908, 1303 and 1894 cm^{-1}. For each section, two profiles were taken. One vertical profile with 8 sampling points, 0.5 mm apart was taken under a nonworn region, and one under the worn region of the polyethylene.

Molecular weight determination

Four specimens were taken (2 Marmor and 2 St Georg). These had been in patients for 8 to 10 years. Sections (300 μm thick) were taken through the polyethylene as described above. For each specimen the upper 2 mm of polyethylene from under the wear track was taken, and a similar amount taken from below the nonworn region. These samples were dissolved in trichlorobenzene with N-Phenyl-2-Naphthylamine at 150 °C. The molecular weight of the dissolved polyethylene was determined by gel permeation chromatography, after the column had been calibrated with polystyrene standards. Dissolution of polyethylene in trichlorobenzene was incomplete and results for molecular weight are for the soluble fraction.

Results

The predominant wear types were deformation, delamination, and pitting of polyethylene with a different wear pattern for each design (Table 1).

St Georg prostheses

Deformation and delamination were the predominant wear types on the St Georg prostheses. Although highly variable, there was a trend towards higher deformation with increasing time of implantation. Delamination of this design occurred in retrieved prostheses approximately 4 years after insertion. Metal backing did not affect wear or delamination (Figure 1). Sub-surface failure was first evident as white areas in the polyethylene (Figure 2).

Examination of these areas in the SEM showed subsurface cracking 1–2 mm below the surface (Figure 3). These cracks occurred in the most highly stressed region as evidenced by deformation of the polyethylene in polarized light (Figure 4). Numerous fusion defects were seen in sections through the polyethylene in both light microscopy and in the SEM (Figure 5). On average these defects measured 146 μm in diameter. The defects were present under both the worn and nonworn regions. Under the nonworn regions, the defects were attached by fine threads to the surrounding polyethylene (Figure 5). Under the wear track in the highly stressed areas, the threads were not apparent, and the defect was debonded from the surrounding

Table 1. Retrieved unicompartment knee prostheses

	St George	Marmor	PCA Uni
Tibial components (n)	69	26	11
Patients (n)	57	15	10
Average duration (yrs)	9.8	6.4	3.0
Range of duration (yrs)	1–19	1–13	1–5
Average score			
Delamination	3.4	0	7.8
Pitting	< 1	2.8	< 1
Cement entrapment	0	< 1	< 1
Cement abrasion	0	< 1	< 1
Scratching	0	< 1	0
Deformation	4.3	5.8	1.3
Molecular weight $\times 10^3$ and (*SE*)			
under nonworn region	2151 (*1097*)	1655 (*782*)	
under worn region	842 (*39*)	837 (*777*)	

polyethylene (Figure 6). In these regions, high stresses around the defects were apparent (Figure 7). In addition, defects in this region formed initiation sites for cracks (Figure 8).

Marmor prostheses

Wear of the polyethylene component of the Marmor prostheses was very different from wear seen on the St Georg prostheses. A shiny wear track formed on all components with no evidence of delamination (Figure 9). Pitting of the polyethylene also occurred with pits which measured from 250–1000 μm. The pits were shallow, measuring less than 0.5 mm deep (Figure 10), and were formed in regions where there appeared to be high local stresses directly at the surface (Figure 11). Deformation of the polyethylene around the pits appeared to be higher than in nonpitted polyethylene (Figure 11).

PCA prostheses

Severe delamination wear occurred on the PCA prostheses even though they were retrieved after short durations. SEM of sections through areas of delaminating polyethylene showed cracking parallel to the surface and also development of cracks between fused polyethylene granules (Figure 12).

Wear tracks

Both in the St Georg and Marmor prostheses the average length of track was 24 (15–35) mm and was positioned posteriorly (Figure 13).

Molecular weight

The average molecular weight of the soluble fraction of the St Georg polyethylene was over 2 million. The polyethylene under the worn area showed a marginally reduced molecular weight. For the Marmor prostheses the average molecular weight was below 1 million for polyethylene under both the nonworn and worn regions (Table 1).

Polyethylene crystallinity

In sections taken through the unworn regions of both St Georg and Marmor prostheses there was a tendency towards decreased crystallinity at both the bearing and cement surfaces, with an increase in crystallinity in the center of the polyethylene.

The average crystallinity at the surface of unworn polyethylene was 52 percent for the St Georg design and 46 percent for Marmor designs. There was a decrease in crystallinity of the polyethylene under the wear track in all the Marmor components investigated, with crystallinity as low as 34 percent at the surface of the polyethylene for one component (Figure 16).

Changes in crystallinity under worn and nonworn areas of St Georg prostheses were not as large, and although there was a trend to lower crystallinity under worn areas, it was not so apparent as that observed for Marmor designs (Figure 16).

Discussion

The three designs investigated in this study were similar: the tibial polyethylene surfaces were essentially flat in St Georg and PCA designs, with a slight radius

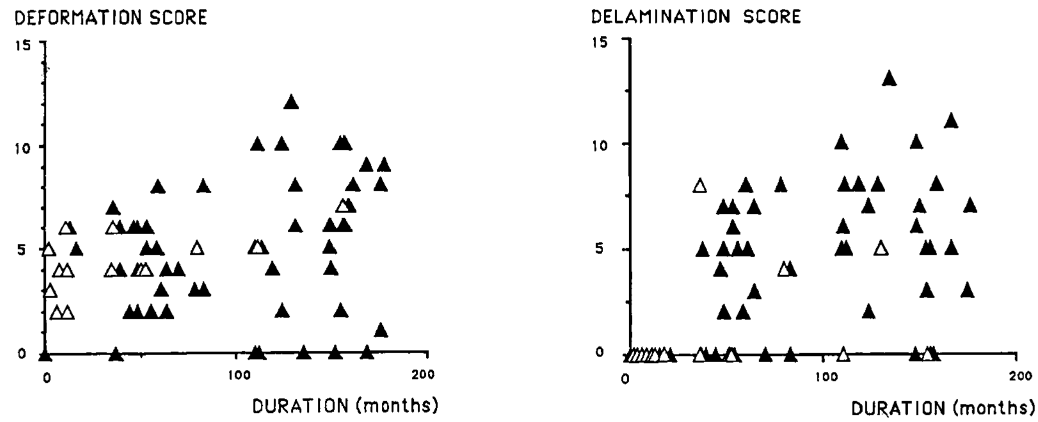

Figure 1. Deformation and delamination scores against duration of implantation of St Georg design; (▲) nonbacked and (△) metal backed.

Figure 2. Photograph of retrieved St Georg tibial component (duration 7 years) showing break-up of the polyethylene surface due to delamination.

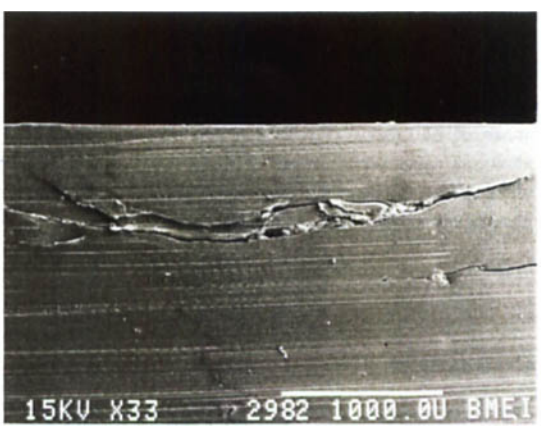

Figure 3. Scanning electron micrograph of section through St Georg tibial component (duration 4 years) showing sub-surface cracking, ×35.

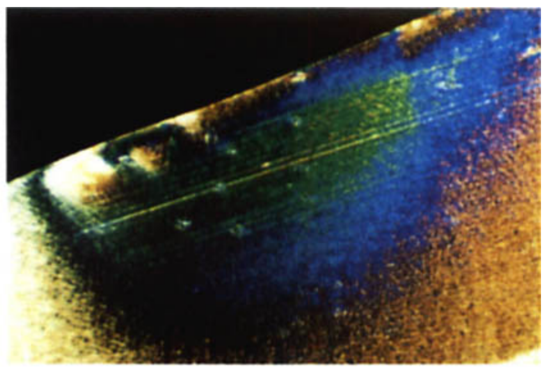

Figure 4. Polarized light photomicrograph of a thin section through retrieved St Georg Sledge polyethylene showing deformation of polyethylene which occurs under the wear track. The highest deformation (yellow color) is seen approximately 0.5–1 mm below the polyethylene surface, ×14.

Figure 5. Scanning electron micrograph of fusion defect under non-worn region of polyethylene. The defect is partially bonded to the surrounding polyethylene, ×400.

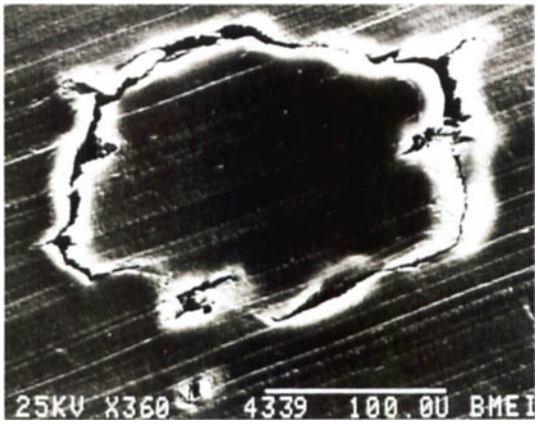

Figure 6. Scanning electron micrograph of a fusion defect under worn region in the highly deformed region of the polyethylene, ×400.

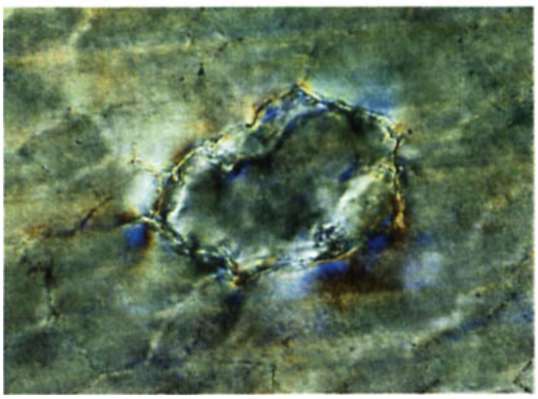

Figure 7. Polarized light photomicrograph of a thin section through retrieved St Georg polyethylene under the wear track showing deformation (stress) of the polyethylene around the edges of the defect, ×260. Duration 6 years.

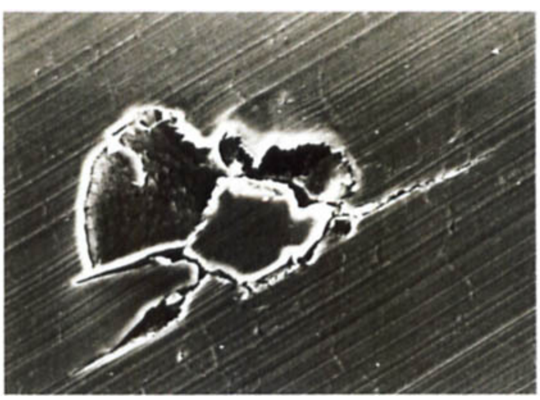

Figure 8. Scanning electron micrograph of fusion defect under the wear track showing initiation of cracks from the edges of the defect, ×140. Duration 7 years.

Figure 9. Photograph of retrieved Marmor tibial component. The component has been sputter-coated with a layer of gold to show clearly deformation of the wear track and the numerous pits, ×5. Duration 7 years.

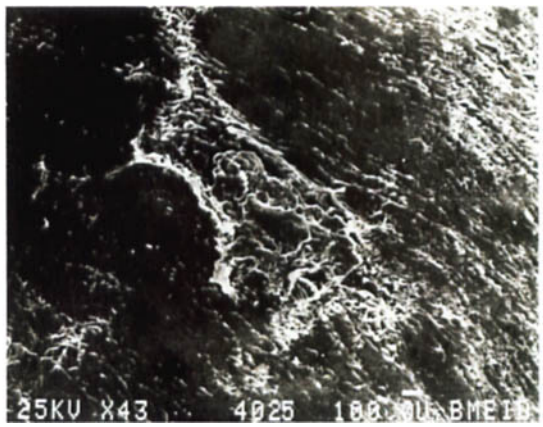

Figure 10. Scanning electron micrograph of the surface of Marmor component showing shallow pit with partially detached particle of polyethylene, ×50.

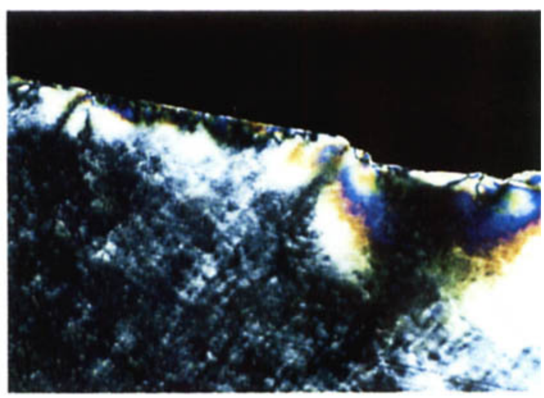

Figure 11. Polarized light photomicrograph of a section through Marmor polyethylene showing pit with deformation of polyethylene around the pit and deformation at the tibial surface, ×160. Duration 7 years.

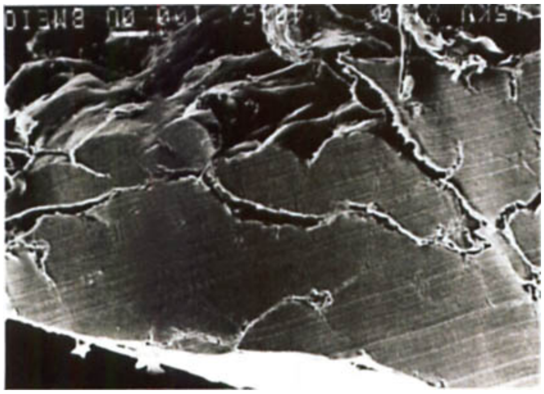

Figure 12. Scanning electron micrograph of section through PCA tibial component showing delamination and intergranular failure of polyethylene, ×120. Duration 1 year.

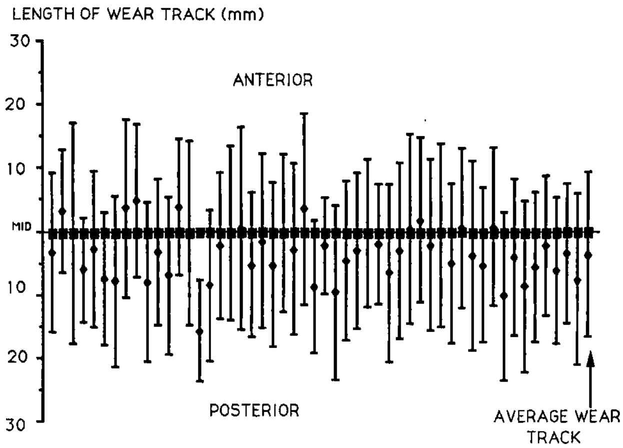

Figure 13. Position and length of wear tracks on St Georg Sledge prostheses.

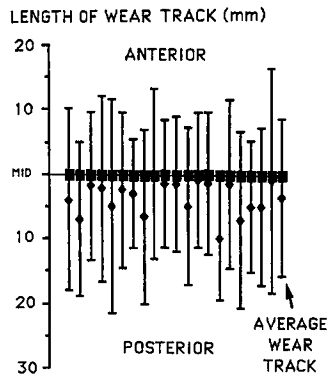

Figure 14. Position and length of wear tracks on Marmor prostheses.

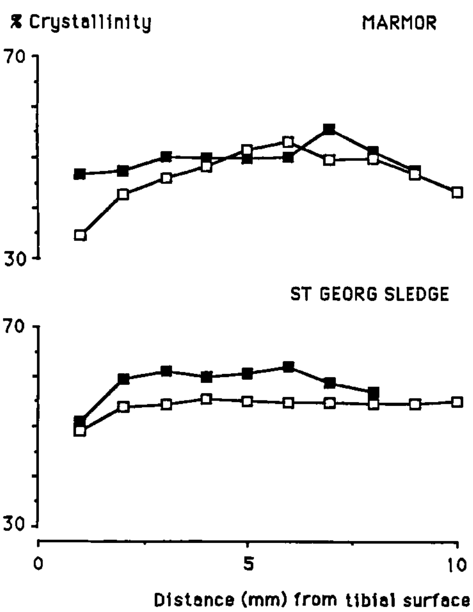

Figure 15. Profiles for percentage of crystallinity under the wear track and non-worn regions for St Georg Sledge and Marmor tibial components; (■) unworn and (□) worn.

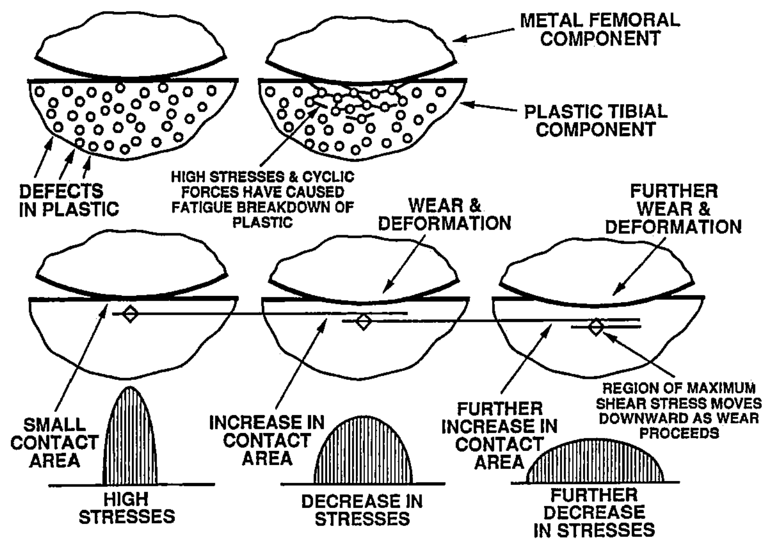

Figure 16. Mechanisms of wear on Marmor and St Georg tibial components. For the St Georg, high stresses and cyclic forces cause fatigue breakdown of the polyethylene due to initiation of cracks from defects. For the Marmor the small contact area caused by flat plastic surface leads to high stresses in the polyethylene. With time, deformation and wear results in an increase in contact area and reduced stresses. Cracking does not occur due to the small number of defects in the plastic and the reduced stresses. In addition with time, the region of maximum stress moves downwards due to further wear and deformation so that fatigue failure never occurs.

in the polyethylene of the Marmor designs. The revision rates of the St Georg and Marmor prostheses appear to be similar but a much higher rate has been recorded for the PCA prosthesis (Knutson et al. 1992).

Analysis of retrieved prostheses indicates specific wear mechanisms associated with each design. The St Georg and PCA both delaminated, however, in the PCA delamination occurred earlier and was more severe. Delamination of PCA total knees has been attributed to hot pressing of the polyethylene surface coupled with the low conforming surface (Bloebaum et al. 1991, Collier et al. 1991, Engh et al. 1992, Jones et al. 1992). Hot pressing leads to reduced crystallinity of the polyethylene in a well defined zone, extending to approximately 1 mm below the surface (Bloebaum et al. 1991). The interface between this zone and the underlying polyethylene occurs in a position where subsurface stresses are greatest and it results in early delamination of the polyethylene surface.

St Georg polyethylene tibial components start to delaminate after about 4 years of use; in our series prostheses retrieved before 4 years' duration showed no delamination. This is characteristic of fatigue-type failure. Pitting wear, however, was not identified in this design, but in the Marmor prosthesis it was due to locally high stresses at or near the surface of the polyethylene. Pitting wear is not associated with three-body abrasion and has been identified in other retrieval studies (Wright and Bartel 1986, Collier et al. 1991).

The wear tracks on retrieved St Georg and Marmor prostheses were similar in length; compared to anterior-posterior displacements observed in kinematic studies of patients with total knees they were extremely long. For example, Thatcher et al. (1987) found 16 mm anterior-posterior displacement whilst El Nahass et al. (1991) found 14 mm. This indicates that the unicondylars may have been unstable prior to retrieval, and that the femoral component may slide rather than roll on the tibial surface. An in vitro study (Blunn et al. 1991) on the effects of the kinematics at the bearing surface on the wear of polyethylene suggested that sliding rather than rolling of the femoral component predisposed to delamination. Nevertheless this does not account for the different types of wear seen on the Marmor and St Georg prostheses.

The reason for the greater delamination wear on the St Georg as compared to the Marmor is not the differences in crystallinity or molecular weight of the polyethylene; our study demonstrated that the St Georg polyethylene has a higher crystallinity and molecular weight than that of the Marmor. Both high crystallinity and high molecular weight are associated with an increase in the wear resistance of polyethylene (Bartel et al. 1986, Li and Howard 1990).

Crystallinity increased towards the center of the polyethylene in sections under nonworn areas of all components examined; this has been attributed to the effect of radiation sterilizing polyethylene which has been shown to generate free radicals and oxidative degeneration resulting in reduced crystallinity at or near polyethylene surfaces (Nusbaum et al. 1979, Streicher 1991). In both St Georg and Marmor prostheses crystallinity of the polyethylene was reduced under the wear track. This implies that stresses alter the chemical nature of the polyethylene in these designs. A recent study has demonstrated higher levels of free radicals under the wear track of retrieved polyethylene knee components than in nonworn areas (Jahan et al. 1991), which may account for the reduction in molecular weight seen in the St Georg design in this study.

Our results suggest that sub-surface cracks are initiated by the fusion defects which were seen in the St Georg polyethylene. They are due to the manufacture of polyethylene which is processed by extrusion or compression-moulding into bar form. Components are then machined into their final shape. Total consolidation is not possible because of the high melt viscosity. The number of fusion defects remaining in processed polyethylene will depend upon the degree of consolidation. Rose and Radin (1990) speculated about the role that fusion defects played in "severe wear" regimes, suggesting they played a crucial role in propagating cracks. Our study demonstrates that Marmor polyethylene which did not contain fusion defects never delaminated. It also shows that the defects in highly stressed regions of the polyethylene produce locally increased stressed concentrations. This leads to debonding of the defect and initiation of cracks which gradually extend and coalesce to liberate flakes and particles of polyethylene from the surface. Once the surface layer becomes discontinuous, the cracking will steadily progress. Even in the plastic without defects, the cyclic sub-surface shear stress will eventually lead to potential crack development but the process will require a higher number of cycles than material with defects. Meanwhile, as deformation and wear at the surface proceed due to adhesive and abrasive mechanisms, the maximum shear stresses will occur at a lower level. Furthermore, the magnitude of the stresses will decrease with time due to an increased contact area. Hence a given area of plastic will receive an insufficient number of cycles at high enough stresses to produce fatigue cracking (Figure 16).

Our study demonstrates that delamination is due to numerous fusion defects which result from poor consolidation of the polyethylene. The critical number of defects in polyethylene which are necessary for delamination to occur has not been identified, but even single defects present in highly stressed regions of the polyethylene may well cause cracks and lead to delamination. Manufacturers should therefore aim to supply polyethylene to the orthopedic industry with as few defects as possible or even plastic with zero defects.

Regarding the clinical significance, it was evident that delamination which rapidly progresses, as in the PCA, can lead to early failure. The fine particulate debris resulting from deformation wear and pitting of Marmor prostheses could result in a more adverse capsular and interface response than larger particles released by delamination of the polyethylene. Finally, it is evident that low conformity per se does not necessarily lead to an excessive amount of wear and deformation, even at more than 10 years follow-up.

Acknowledgements

The authors wish to acknowledge the contribution made by the orthopedic surgeons who supplied retrieved specimens, the Department of Health in the U.K. for supporting the study and the Swedish Medical Research Council, Project 9509. Thanks are due to Arnold Keller of Waldemar Link and to James Davidson of Smith & Nephew Richards for helpful cooperation in terms of components and material data.

References

Bartel D L, Bicknell V L, Wright T M. The effect of conformity, thickness, and material on stresses in ultra high molecular weight components for total joint replacement. J Bone Joint Surg (Am) 1986; 68 (7): 1041–51.

Bloebaum R D, Nelson K, Dorr L D, Hofmann A A, Lyman D J. Investigation of early surface delamination observed in retrieved heat pressed tibial inserts. Clin Orthop 1991; 269: 120–7.

Blunn G W, Walker P S, Joshi A, Hardinge K. The dominance of cyclic sliding in producing wear in total knee replacements. Clin Orthop 1991; 273: 253–60.

Collier J P, Mayor M B, McNamara J L, Surprenant V A, Jensen R E. Analysis of the failure of 122 polyethylene inserts from uncemented tibial knee components. Clin Orthop 1991; 273: 232–42.

el Nahass B, Madson M M, Walker P S. Motion of the knee after condylar resurfacing an in vivo study. J Biomech 1991; 24 (12): 1107–17.

Engh G A, Dwyer K A, Hanes C K. Polyethylene wear of metal backed tibial components in total and unicompartmental knee prostheses. J Bone Joint Surg (Br) 1992; 74 (1): 9–17.

Gueugnaut D, Wackernie P, Forgeit J P. Infrared microspectrometric study of the thermal history of fusion welding of polyethylene. J Appl Polym Sci 1988; 35: 1683–94.

Hood R W, Wright T W, Burstein A H. Retrieval analysis of total knee prostheses a method and its application to 48 total condylar prostheses. J Biomed Mater Res 1983; 17 (5): 829–842.

Jahan M S, Wang C, Schwartz G, Davidson J A. Combined chemical and mechanical effects on free radicals in UHMWPE joints during implantation. J Biomed Mater Res 1991; 25: 1005–17.

Jones S M, Pinder I M, Moran C G, Malcolm A J. Polyethylene wear in uncemented knee replacements. J Bone Joint Surg (Br) 1992; 74 (1): 18–22.

Kilgus D J, Moreland J R, Finerman G A, Funahashi T T, Tipton J S. Catastrophic wear of tibial polyethylene inserts. Clin Orthop 1991; 273: 223-31.

Knutson K, Lewold S, Lidgren L. Outcome of revision for failed unicompartmental knee arthroplasty for arthrosis. Poster Exhibit to 59th Annual Meeting of the American Academy of Orthopaedic Surgeons, Washington, . USA 1992.

Landy M M, Walker P S. Wear of ultra high molecular weight polyethylene components of 90 retrieved knee prostheses. J Arthroplasty (Suppl) 1988; 3: S73–85.

Laurencin C T, Zelicof S B, Scott R D, Ewald F C. Unicompartmental versus total knee arthroplasty in the same patient. A comparative study. Clin Orthop 1991; 273: 151–6.

Li S, Howard E G. Characterisation and description of an enhanced ultra high molecular weight polyethylene for orthopaedic bearing surfaces. Soc of Biomaterials 16th Annual Meeting, Charleston, South Carolina May 20–23 1990.

Lindstrand A, Stenström A, Egund N. The PCA unicompartmental knee. A 1–4 year comparison of fixation with or without cement. Acta Orthop Scand 1988; 59 (6): 695–700.

Nusbaum H J, Rose R M, Paul I L, Crugnola A N, Radin E L. Wear mechanisms for ultra high molecular weight polyethylene in total hip prosthesis. J Appl Polym Sci 1979; 23: 777–89.

Rose R M, Radin E L. A prognosis for ultrahigh molecular weight polyethylene. Biomaterials 11, 1990: 63–68.

Steicher R M. The behavior of UHMWPE when subjected to sterilization by ionizing radiation. In: Ultra high Molecular Weight Polyethylene as Biomaterial in Orthopaedic Surgery (Eds. Willert, H. G, Buckhorn, E. P) Hogrefe & Huber, Toronto 1991: 66–73.

Thatcher J C, Zhou X M, Walker P S. Inherent laxity in total knee prostheses. J Arthroplasty 1987; 2 (3): 199–207.

Walker P S. Bearing surface design in total knee replacement. Eng Med 1988; 17 (4): 149–56.

Walker P S, Hsieh H H. Conformity in condylar replacement knee prosthesis. J Bone Joint Surg (Br) 1977; 59 (2): 222–8.

Wright T M, Bartel D L. The problem of surface damage in polyethylene total knee components. Clin Orthop 1986; 205: 67–74.