Wear of ultra-high molecular weight polyethylene (UHMWPE) in total knee prostheses: A review of key influences

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Wear of ultra-high molecular weight polyethylene (UHMWPE) in total knee prostheses: a review of key influences

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Abstract: The formation and development of wear is now widely accepted as one of the major concerns in the long-term survivorship of contemporary knee prostheses *in vivo*. This review examines the role of surface topography, third-body debris, load, contact mechanics and material quality in the wear process. Some of the kinematic and physiological issues that need to be modelled in the development of wear testing regimes for evaluation of material combinations and geometrical combinations in total knee implant designs are considered. Wear testing procedures and some of the results from wear tests are discussed and the need to consider the impact of rolling and sliding in the study of wear in total knee components is highlighted. The dominant wear mechanisms that occur *in vivo* are identified and the role of these mechanisms is currently being examined experimentally at the University of Limerick wear testing machine.

Keywords: polymer wear debris (PWD), total joint replacement (TJR), total knee replacement (TKR), third body debris (TBD)

1 INTRODUCTION

Over the last 30 years, ultra-high molecular weight polyethylene (UHMWPE) has been the choice of material used as bearing surfaces in total joint replacements (TJRs). It is renowned for its toughness, low friction and biocompatibility [**1**–**3**]. Designs of total knee implants have a variety of contact geometries. Many now have relatively high conformity whereby the radii of the curvature of both femoral and tibial components are similar. A considerable number of designs, however, use geometries that have relatively low conformity with a convex metallic surface articulating on a softer non-conforming concave UHMWPE surface. While design and surgical fixation techniques have improved, the concern is not only that the polyethylene component will wear out but, that the development of polymer wear debris (PWD) will evoke undesirable effects and limit the long term survival of total knee replacements (TKRs) [**4**–13]. The influence of surgical techniques on the long-term performance of TKRs has been well documented and will be discussed later. *In vivo* the generation of submicron-sized PWD can cause adverse cellular reactions in the surrounding tissues, causing macrophage activity and necrosis of the bone–prosthesis interface [**14**–**19**]. The biological response to loose PWD, is believed to be a contributing factor to prosthetic loosening in TKRs.

In laboratory wear studies, it is important to know what conditions can cause an increase in the wear rate of polyethylene and hence an increase in the number of polymer particles. Polyethylene wear has been related directly to its molecular properties, and factors have been identified that affect the molecular characteristics of polyethylene and thereby its wear characteristics. These factors include the quality of the initial resin [**20**], fabrication [**21**], sterilization techniques [**22**] and atmospheric exposure that can lead to oxidative degradation [**23**].

Other design factors affecting the wear characteristics of UHMWPE include, the geometries of prostheses [**24**, **25**], topography [**26**], the presence of third-body debris (TBD) [**27**], loading [**28**], motion [**29**] and lubrication conditions [**30**]. The following review will discuss these factors in more detail.

2 *IN VITRO* STUDIES OF WEAR

2.1 Surface topography

Surface topography of the metallic femoral component in a wide range of total knee implants remains one of

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the most dominant factors determining both the volume and the morphology of UHMWPE wear debris [**22**, **31**–**38**]. It is affected by the machining and polishing processes during manufacture of the devices, but the presence of TBD, such as particles of loose bone and bone cement, can also damage the bearing surfaces and promote increased wear.

Dowson *et al.* [**35**] predicted an optimum counterface R_a value in the range 0.05–0.1 μm under dry wear-testing conditions. Later investigation showed that the wear rate of polyethylene increased steadily with effective counterface roughness under lubricated conditions. Generally the wear rates are lower for lubricated conditions when the counterface roughness is somewhat below the optimum R_a value predicted by Dowson *et al.* [**35**] (Fig. 1). Both wet (non-protein) and dry conditions produce a polymer transfer film on to the metallic counterface [**36**, **39**]. Under water-lubricated conditions, a high counterface R_a value causes an initially high wear rate and, after a period of time, steady state wear is observed. Subsequently, sliding on a smooth counterface produces a continuous steady state wear of UHMWPE but may lead to a sudden increase in wear rate, as the polymer surface approaches a strain limit due to cyclic loading, resulting in imminent surface cracking [**33**, **40**].

Dry or water-lubricated conditions are currently considered an inaccurate representation of relative wear rates [**40**, **41**], as a polymer transfer film has not been observed on surgically retrieved knee implants [**33**, **41**]. However, Pappas *et al.* [**42**] recommended water as the preferred medium for joint simulation, since they found no polymer transfer film during long term testing (48×10^6 cycles). During wear testing, the use of a protein lubricant environment, shows a steady increase in polyethylene wear and with supposedly no noticeable polymer transfer film [**40**]. The greater the percentage of synovial fluid added to distilled water, the lower is the wear rate of UHMWPE [**30**]. The present authors' research at the University of Limerick has confirmed that significantly increased wear rates occur for a water-lubricated condition (3.2×10^{-4} g per 10^6 cycles) compared with a protein lubricant condition (0.5×10^{-4} g per 10^6 cycles), under stiff gait knee simulation [**43**].

Hailey *et al.* [**26**] found that various lubricants had little influence on the size and shape of wear debris generated, provided that there was no polymer film transfer. They discovered that a rough counterface produced a larger volume and smaller size of wear debris than did a smooth counterface which produced larger smoother plate-like debris. McKellop *et al.* [**44**] noticed that, under protein lubrication, a transfer film of polyethylene on to the metallic counterface occurred after every fresh batch

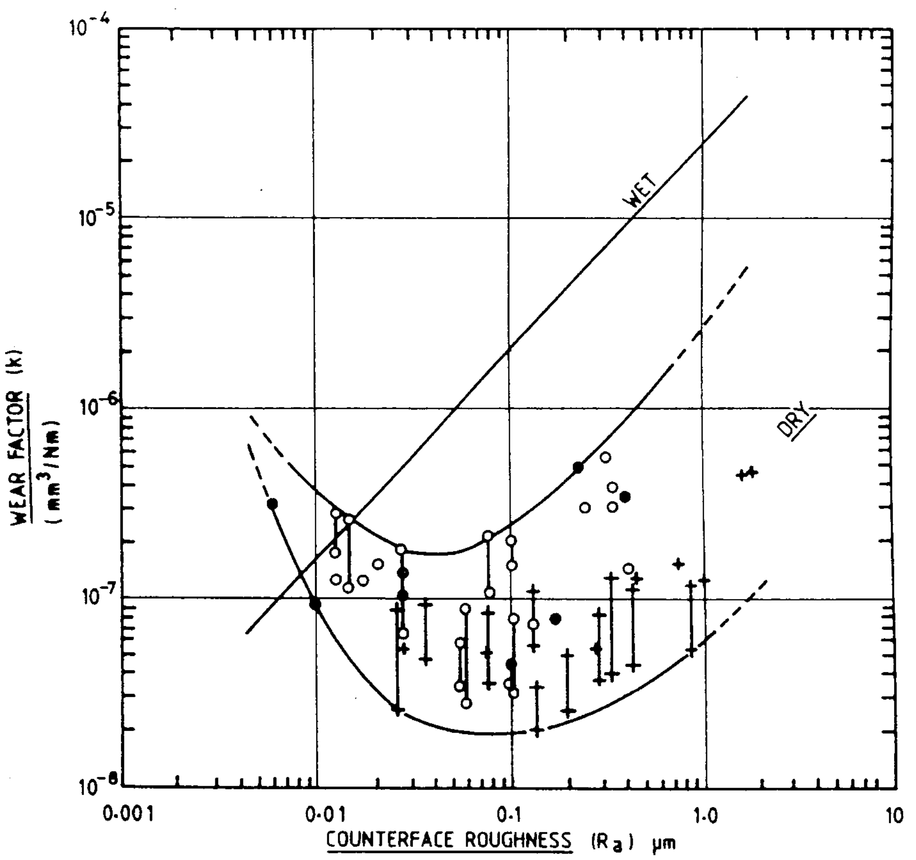

Fig. 1 Wear factor k as a function of the counterface roughness R_a, under wet and dry lubricated conditions. (From Dowson *et al.* [**35**])

of lubricant during testing. This resulted in an increase in friction between the bearing surfaces, and a break-up of the film, causing a significant amount of polyethylene wear. This transfer of polymer film during protein lubrication was not found at our laboratory, although an initial increase in friction for several hundred cycles was observed when each old batch of lubricant was replaced [**43**]. The observation by McKellop *et al.* of polymer transfer to the metallic surface appears to contradict other research in the same field. The precise biological mechanism for testing in protein containing lubricants is still poorly understood.

Lankford *et al.* [**32**] carried out an investigation, varying the initial surface roughness of UHMWPE sliding against surgical stainless steel in bovine serum. Similar to Cooper *et al.* [**33**, **40**], they found that the volume loss of polymer increased steadily for rough surfaces ($R_a = 2.0$ μm), while there was an initial low wear rate for the smooth surfaces ($R_a = 0.06$ μm), followed by an acute increase in wear rate. According to Fisher *et al.* [**45**], varying the sliding velocity during *in-vitro* testing has negligible effect on the wear factor with relatively smooth surfaces (0.05 μm or less). An increase in counterface roughness of between 0.07 and 0.08 μm results in a statistically significantly higher wear factor for lower velocities (0.2 m/s). Fisher *et al.* [**45**] thus showed that with relatively smooth counterfaces it was not unreasonable to use higher sliding test velocities (0.4 m/s).

2.2 Third-body debris (TBD)

The roughening of the counterfaces, during articulation is of concern to the survival rate of TKRs. Particles such as bone and bone cement have been found embedded beneath and on the surface of UHMWPE components after clinical retrieval and these particles can clearly cause TBD [**46**]. Some researchers have concluded that TBD is the dominant cause of increased polyethylene wear and premature failure *in vivo* [**5**] and suggested that the use of bone cement in joint replacement can increase the probability of increased counterface roughness. *In vivo*, particles of bone cement may separate from the joint and become entrapped between the articulating surfaces. These hard particles may promote increased counterface roughness (Fig. 2) during articulation and hence increased wear rates and damage to the polyethylene surfaces.

A single scratch transverse to the sliding direction can produce a remarkable seventyfold increase in wear rate [**31**]. Caravia *et al.* [**27**] showed that fragments of polymerized bone cement with zirconia additive produced a considerable increase in counterface roughness and wear rate during pin-on-flat testing. However, optical studies showed that the severity of damage and number of particles entering the contact zone was dependent on the initial surface roughness of the polymer component, the contact stress, the conformity and the size and geometry of particles. Poggie *et al.* [**47**] carried out work on thin film coatings of counterfaces and showed that counterface abrasion decreased with increasing coating hardness. Their work agreed with that of Mishra and Finnie [**48**], who concluded that mild to virtually no abrasion occurs when the hardnesss of the substrate is increased from 0.8 to 1.2 times that of the third bodies. The abrasion of metallic surfaces *in vitro* is of concern. After 2×10^5 cycles, in recent wear tests, scratches were observed on stainless steel counterfaces even though testing was carried out in a clean environment and similar findings have been found by McKellop *et al.* [**44**]. The reasons for such scratching are not fully understood but may be related to the hardness of the counterface.

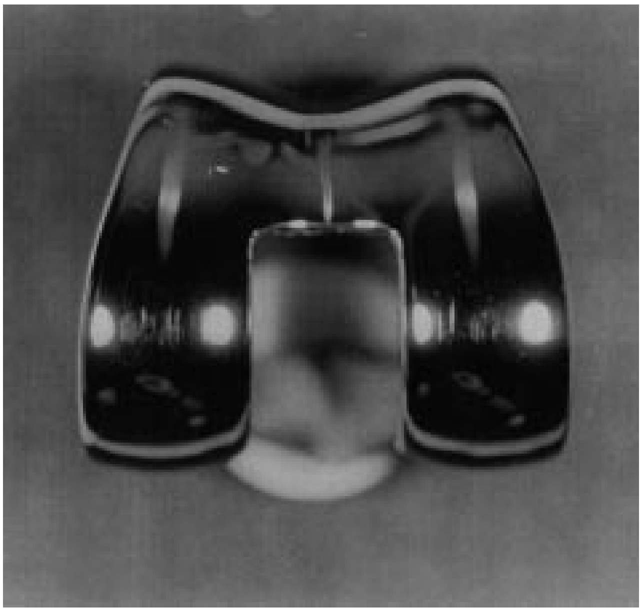

Fig. 2 Damaged femoral surfaces due to the presence of microscopic hard particles during articulation (University of Limerick)

2.3 Load, contact stress and geometry of the contact

The loads on the polyethylene tibial inserts can reach peak values of three times the body weight during normal walking conditions [**49**] and can peak between four to five times the body weight during more stressful activities, such as walking upstairs or during a running activity. This is of concern as contact stresses can exceed the compressive yield stress of UHMWPE (approximately 23 MPa) (**3**, **24**, **25**, **50**–**53**). Such excessive stresses may promote delamination and pitting of the polymer surface [**50**–**54**]. The geometry of the prosthesis and the type of loading determine the contact area and size of the contact stresses in the polyethylene component. Work in this area has been carried out, both experimentally and using finite element analysis

(FEA)[**3**, **24**, **25**, **50**, **54**]. Walker *et al.* [**55**] agreed with the hypothesis that the presence of thin tibial inserts, increases contact stress and hence increases the wear rate of UHMWPE. Bartel [**3**], Pappas [**53**] and Argenson and O'Connor [**11**] share the view that high conformity may reduce wear rates. In a non-conforming contact with sliding, McGloughlin and Monaghan [**56**] found that the peak shear stress moved closer to the surface, clearly a more severe condition of damage to the polymer. This also supports the view that higher conformity would reduce wear.

In a later section discussing the influence of physiological and kinematic conditions on the wear rate of polyethylene, the effect of increasing the load and contact stress is discussed. Generally FEA and *in-vitro* contact studies indicate advantages in using high conformity articulating surfaces, as contact stresses are decreased. However, these were static tests. McGloughlin [**50**] showed that deformation of the UHMWPE insert could cause the polymer to cycle from tension to compression during the walking cycle. They believed that this condition together with counterface abrasion contributes to high wear rates. Observed from a dynamic point of view, there have been many test regimes used for assessing the wear of prosthetic materials. However, the variety of kinematic conditions has led to contradictory results, in relation to the influence of load and contact stresses.

2.4 Quality of UHMWPE

The quality of UHMWPE may be one of the dominant factors affecting the development of wear in the knee prosthesis [**10**, **12**, **20**, **21**, **55**]. The polyethylene implants used today are machined from extruded bar stock and compression moulded sheet or compression moulded directly from polyethylene powder. Furthermore, it is agreed that low molecular weight, the inclusion of Al, Si and Ti in the catalyst and very high crystallinity can subsequently affect the morphology of the fabricated UHMWPE component [**57**]. However, such catalysts have a small influence on the wear rate of UHMWPE compared with the manufacture and irradiation processes. Prior to implantation the tibial components are sterilized, most frequently using gamma irradiation of 2.5 Mrad, which has a further effect on the properties and wear performance of UHMWPE [**22**, **23**, **58**–**64**].

Bankston *et al.* [**21**] carried out a radiographic evaluation of 236 hip prostheses manufactured from machined extruded bar and direct compression moulds. Their results showed a significant increase in the wear of polyethylene that had been machined from extruded bar (0.11 mm/year), compared with polyethylene that had been compression moulded (0.05 mm/year). Tanner *et al.* [**20**] noticed that, of the 29 tibial components retrieved at revision surgery, custom-made products, made using the direct compression moulding process,

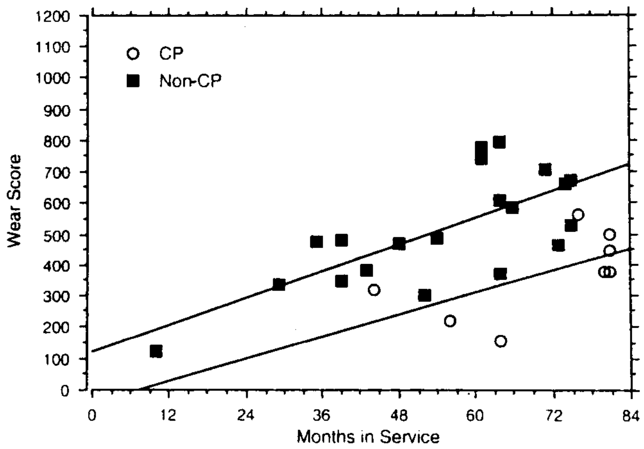

Fig. 3 Graph of wear scores versus time in service for custom product (CP) and non-custom product components *in vivo*. (From Tanner *et al.* [**20**])

had minimal wear compared with non-custom ram extruded components (Fig. 3). Laboratory tests carried out by Blunn *et al.* [**65**] observed large subsurface cracks, 1–2 mm below the surface, which may have initiated at material fusion defects, or other subsurface defects. Both Bankston *et al.* [**21**] and Tanner *et al.* [**20**] concluded that the severity of wear would decrease with improved quality control of polyethylene. Compression moulding appears to offer better wear rates but there is clearly an understandably higher manufacturing cost associated with this process.

Sterilization techniques, sterilization dosage, packaging atmosphere and subsequent ageing conditions can have a profound effect on the *in vitro* physical and mechanical properties of UHMWPE, and overall wear characteristics [**22**, **57**, **58**, **61**, **66**]. The importance of molecular architecture is largely determined by the polymerization process but is subsequently altered by irradiation and oxidation of UHMWPE. Gamma irradiation is the most frequent form of sterilization causing some cross-linking, chain scission and the generation of free radicals which may lead to surface and subsurface oxidative degradation over time. Fisher *et al.* [**22**] showed that the wear of UHMWPE increased with increasing irradiation dosage, under a standard pin-on-flat wear testing device. They further showed that the wear of UHMWPE increased with ageing of the polyethylene. Roe *et al.* [**64**] carried out an extensive study of the effect of radiation sterilization and ageing, showing that radiation levels tend to increase the elastic modulus and yield point stress and to reduce the ultimate elongation. Ageing produced similar effects to increased radiation doses. Blunn *et al.* [**23**], Rimnac *et al.* [**60**] and Kurth and Eyerer [**61**] studied the extent of oxidative degradation from post-gamma-irradiated components. They found that oxidation diminishes the polyethylene wear properties. Blunn *et al.* [**23**] suggested stabilizing the polymer by reducing free radicals generated during irradiation, thereby improving the wear properties of the polyethylene. Fisher *et al.* [**58**], Hinsch [**62**] and Kurth and Eyerer [**61**] concluded that other techniques for irradiation were more favourable, such as irradiation in a protective gas or electron beam irradiation. In contrast, Wang *et al.* [**29**] and Tanner *et al.* [**20**] concluded that the kinematic conditions and the presence of fusion defects had a greater influence on the wear rates of UHMWPE than the effects of either irradiation dosages or ageing.

2.5 Physiological and kinematic conditions

The understanding of the physiological and kinematic condition of prosthetic joint implants *in vivo* is important when attempting to represent and reproduce similar wear rates and mechanisms *in vitro*. It has been suggested that the kinematic condition at the bearing surfaces is the most dominant factor affecting the wear of polyethylene and subsequently determines the type of wear process present [**65**]. It is difficult to quantify the kinematic motions of TKRs. There have been numerous techniques carried out, for both the natural knee and TKRs, using both skin and bone markers as reference points between the upper femur and lower tibia [**67**–**70**].

A range of surgical considerations influences the long-term performance of TKRs. Malpositioning of the femoral and tibial components can cause poor motion and local stresses in the polyethylene [**8**, **9**, **12**, **13**, **37**, **41**, **46**, **67**]. Recent developments in instrumentation have helped to reduce this problem. Malalignment has also been found to occur surgically where there has been a poor soft tissue structure. This too can lead to higher wear rates. There are many test rigs for screening of new biomaterials in today's orthopaedic industry. Knee joint simulators are used to assess device performance in the form of *in-vivo* kinematic conditions and wear rates prior to surgical performance. They are complex and expensive to construct. However, it may not be economic to fabricate prototype knee prostheses for a large number of material combinations, or for investigating and observing factors affecting wear rates. Nevertheless joint simulation testing of knee prostheses is essential prior to clinical trial. Consequently, less expensive and less time consuming wear tests are usually carried out on simple polymer pin-on-metallic-flat machines. However, these pin-on-flat devices poorly represent the kinematic and physiological conditions found in the knee [**29**, **65**, **71**].

The results from numerous wear-testing rigs have been highly variable. Wang *et al.* [**29**] found contradictory wear rates both from a polymer pin on a metallic flat and from a knee simulation rig, when the influence of irradiation dosages was being investigated. Wang *et al.* proposed that component representation and kinematic test conditions between the sliding surfaces were more dominant than the influence of irradiation dosages.

Table 1 Wear factors of UHMWPE under various kinematic conditions, e.g. sliding, rolling and gliding. (From Cornwall *et al.* [**72**])

Test	Condition	Load (N)	Contact stress (MPa)	Wear factor ($\times 10^6$)
1	Sliding	30	18	0.9
2	Sliding	190	45	0.8
3	Rolling	190	24	2.5
4	Gliding	190	18	2.7
5	Gliding (ASTM)	190	3	24.0

Cornwall *et al.* [**72**] contradicted the contact stress hypothesis proposed by Bartel *et al.* [**54**] that greater contact stresses increase the wear rate of polyethylene. The results of Cornwall *et al.* obtained for a metallic sphere in sliding, rolling and gliding contact against a flat UHMWPE surface are shown in Table 1. Sliding is a condition when the contact point on the polymer remaines stationary as the contact point on the metallic surface oscillates. Rolling is a condition when the relative contact point velocity of both the metal and the polymer are equivalent. Finally gliding is a condition when the contact point on the metal is stationary and the contact point on the polymer reciprocates. Graphical representations of the kinematic conditions are shown in Fig. 4.

Earlier experimental work from Rose and Cimino [**73**] and Rose and Goldfarb [**74**] agreed with the finite element studies of Bartel *et al.* that an increase in load may result in an increase in wear. The apparatus used by Rose and co-workers closely modelled the *in vivo* physiological and kinematic conditions occurring in TKRs, as the apparatus induced unidirectional spatially varying stresses on the polymer, typical of a fatigue failure found in retrieved TKRs. This could be considered a gliding condition. However, one of the test conditions (Table 1, sliding, tests 1 and 2) employed by Cornwall *et al.* [**72**] contradicted the computational work of Bartel *et al.* [**3**, **54**]. Cornwall *et al.* tested a kinematic condition found at the near end of flexion during articulation of a low-conforming TKR and showed that an increase in stress resulted in a decrease in the formation of wear.

Barbour *et al.* [**28**] computationally monitored the effect of contact stress on the wear of polyethylene. They discovered that their results (Fig. 5) contradicted the work of Rose and Goldfarb [**74**] and Rostoker and Galante [**75**] and demonstrated that an increase in contact stress resulted in a decrease in wear. Barbour *et al.* [**28**] used a polymer pin on a metallic flat. The present

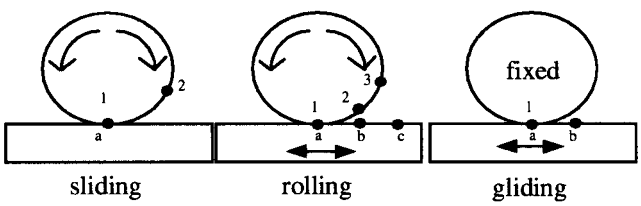

Fig. 4 Sliding conditions carried out by Cornwall *et al.* [**72**]

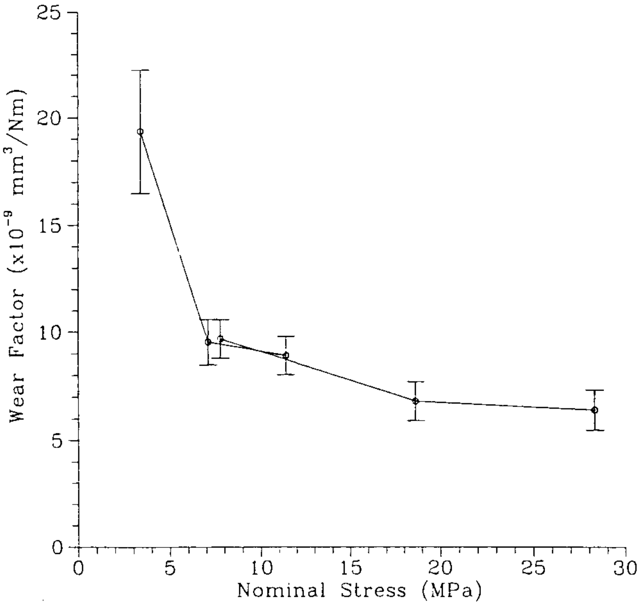

Fig. 5 Mean wear factor versus the nominal contact stress, with error bars showing standard error. (From Barbour *et al.* [**28**])

authors believe that such a test regime does not model the physiological behaviour of a TKR.

During the past 20 years, laboratory studies on the wear of polyethylene have been conducted using unidirectional or reciprocating linear wear-type rigs consisting of a polymer pin on a metallic flat. These rigs bear little resemblance to the motion and loading configurations experienced in the knee joint and further produce little correlation with clinical findings [**29**, **31**, **72**]. Laboratory wear tests carried out by Walker *et al.* [**56**], Cornwall *et al.* [**72**], Blunn *et al.* [**65**] and and Wang [**76**] more closely represented physiological and kinematic conditions and yielded results which resembled wear behaviour observed in clinical retrievals. Bragdon *et al.* [**71**] showed that the physiological motion was a key factor. The latter workers concluded that it is necessary to model accurately the physiological and kinematic conditions of the joint both for screening of new biomaterials and for investigating factors that influence the wear of these materials.

3 WEAR MECHANISMS AND PROCESSES

Wear mechanisms of UHMWPE have been well described qualitatively, by observation of the polyethylene surface and debris formation under a high magnification. To date, there are three classical mechanisms of wear described for sliding of a hard counterface on UHMWPE. These are (a) polymer-to-metallic adhesive wear, (b) counterface abrasive wear and (c) polymer fatigue wear. A well-known mechanism for TKRs is the so called counterface abrasion–fatigue wear mechanism.

Under a reciprocating pin-on-flat motion, McKellop *et al.* [**44**] showed that polyethylene contact areas, free from third-body damage, had local flow or remoulding and a ripple texture, with randomly oriented surface cracks. This was a result of the alternate stressing of the polymer surface and subsequent fatigue cracking. However, McKellop *et al.* [**44**] found no increase in wear rate associated with this effect. Kernick and Allen [**30**] carried out similar tests and extended the hypothesis of McKellop *et al.* Under applied cyclic loading, a deformed and hardened polyethylene surface layer formed, resulting in enhanced mechanical properties and crystallinity of the surface. Similar to McKellop *et al.*, Kernick and Allen found that no measurable wear took place. However, after further loading and as the surface layer reached a state of maximum strain, wear initiated through debonding at the interface of the bulk polymer and hardened surface layer. Cooper *et al.* [**33**] had previously observed similar findings.

Blanchet and Kennedy [**34**] discovered that under dry conditions a similar polymer surface of enhanced crystallinity formed. However, Blanchet and Kennedy described this surface as a thin polymer transfer film on the metallic counterface and suggested that it occurred by a fibular pull-out mechanism. Marcus and Allen [**77**] and Lankford *et al.* [**32**] carried out similar testing to Blanchet and Kennedy [**34**] and observed a similar fibular pull-out mechanism. Generally, under close observation, so called ‘anchors’, from loose micrometre-sized polymer particles, attached to the free metallic surface, resulting in a ‘fibril’ pull-out mechanism from the loose particle (Fig. 6). Cumulative fibril pull-outs or drawing resulted from repeated reciprocation and final fibril rupture. Finally the strongly adhered fibril eventually plastically smeared over the counterface due to further

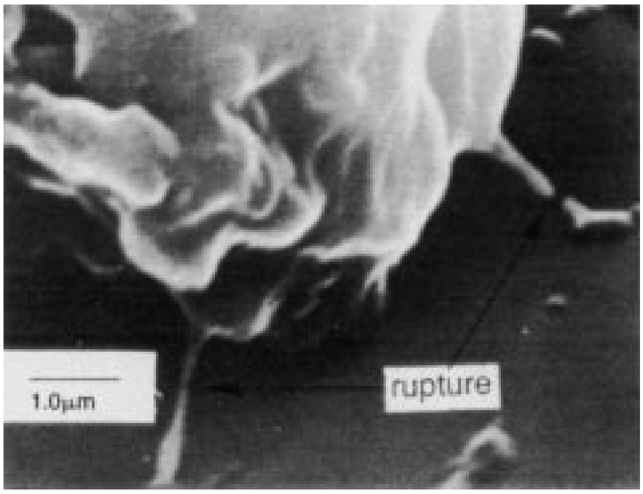

Fig. 6 Fibril rupture leaving residual UHMWPE anchors upon the counterface. (From Blanchet and Kennedy [**34**]). Reproduced by permission of the Society of Tribologists and Lubrication Engineers, published by *Lubrication Engineering and Tribology Transactions*)

cumulative shear formation between the polymer and metallic surfaces [**34**]. The transfer film became highly oriented with an increased anisotropical crystalline structure. Polymer film transfer is a complicated process, with many descriptive theories [**30**, **32**, **33**, **77**]. From the literature evidence thus far, polymer adherence is not characteristic on femoral component surfaces in prosthetic knee retrievals or from *in vitro* testing using protein-containing lubricants [**33**]. However, Cooper *et al.* [**33**] believed that this process cannot be omitted as there is no clear evidence clinically that a polymer transfer film on the metallic surfaces followed by polymer break-up exists during articulation of the components.

Blunn *et al.* [**65**] modelled the geometry and kinematic conditions of the knee and described a fatigue wear mechanism with subsurface cracking, resulting in surface delamination. They suggested that the amount of damage to the polymer surface was kinematically sensitive. Using low-power microscopy, it was found that cyclic loading and sliding resulted in the greatest damage with delamination and cracking at a depth of 100 μm and evidence of large subsurface cracks 2 mm below the surface (Fig. 7).

To date, the work carried out by Cooper *et al.* [**40**] is the most practical description of mechanistic wear and harmonizes with the classical mechanisms of wear mentioned earlier. They described two separate types of wear process: a microscopic counterface wear process and a macroscopic polymer wear process. The microscopic counterface wear process was more associated with abrasion of polymer material, while the macroscopic polymer wear process occurred by fatigue failure of polymer due to high subsurface strains. Fisher [**41**] reviewed the wear processes described by Cooper *et al.* [**40**], while Wang *et al.* [**29**, **76**] later extended these observations and showed that the transition between processes depended critically on mechanical properties, such as the ultimate tensile strength (UTS) and ductility of UHMWPE. Wang *et al.* [**29**] and Bragdon *et al.* [**71**] concluded that the wear mechanism present was strongly sensitive to the kinematic motion between the polymer–metal components. They stated that polyethylene strain hardened under uniaxial stretching through molecular orientation which induced an anisotropy structure and reduced wear rates. However, during multi-axial motion, molecular orientation did not occur and failure was likely to be initiated by intermolecular or transverse shear rather than by long-chain rupture.

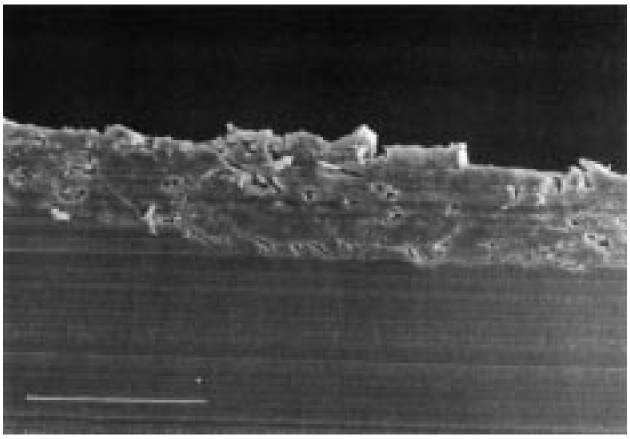

Fig. 7 Scanning electron micrograph of a section through a specimen showing delamination of the surface, which is severely cracked. The bar represents 100 μm. (From Blunn *et al.* [**65**])

4 DISCUSSION

The bearing surface topography of both the femoral and the tibial components have a large influence on the wear of UHMWPE, and from laboratory tests it is apparent that rougher surfaces can create a larger volume of wear debris [**26**, **31**, **33**, **35**]. Over the years there have been several means of lubrication for wear testing, and it has been discovered that the lubrication conditions strongly affect mechanism and rate of polyethylene wear [**30**, **36**]. Many descriptive mechanisms have been suggested for polymer transfer to counterfaces; however, the exact biological mechanism for protein containing lubricants is not yet fully understood. From a prosthetic design point of view it is recommended that meaningful wear testing data can only be obtained using physiological lubrication in laboratory tests [**33**]. However, Pappas *et al.* [**42**] suggests that, due to the standard procedures of synovectemy and capsulectemy, little synovial fluid is generated after arthroplasty. Pappas *et al.* suggest the introduction of a standardized lubricant for wear testing prosthetic joint materials. It has been suggested that the generation of smaller particles and larger volumes of polymer debris was linked to osteolysis and implant loosening, causing adverse cellular reactions [**14**]. Therefore, it is recommended that smooth bearing surfaces should be maintained at all times in the design of TKRs. Ceramics and thin-film surface coatings have been shown to exhibit excellent low surface roughness values, producing exceptionally low wear rates against UHMWPE [**46**, **78**].

UHMWPE tibial components are susceptible to abrasion from harder counterfaces, which in turn may be abraded by harder third bodies such as bone and bone cement [**27**]. Damage to the femoral component counterface is very sensitive to hard TBD and the rate of UHMWPE wear can increase dramatically [**31**]. The possibility of minimizing damage to the counterfaces and hence abrasion of UHMWPE was shown by Poggie *et al.* [**46**]. These workers recommended minimizing the development of hard TBD, possibly by the application of improved cementless fixation techniques or improved cementing methods, and harder and smoother counterfaces. However, whether the problems of prosthetic loosening can be solved in this manner is still unknown.

Bartel *et al.* [**24**], McGloughlin [**50**], Pappas *et al.* [**53**] and Sathasivum and Walker [**79**] carried out computational prosthesis metal–polymer contact analyses. They predicted optimum design features of the bearing surfaces and that contact stresses in tibial inserts could be minimized by making articulating surfaces more conforming. From these studies they assumed that the wear of UHMWPE may be reduced by introducing these surface features. The effect of contact stress on the wear of UHMWPE has been evaluated at different centres and findings have been ambiguous. Cornwall *et al.* [**72**] found that an increase in contact stress resulted in a decrease in wear. The results of Barbour *et al.* [**28**] agreed with some test conditions of Cornwall *et al.* but contradicted the earlier work of Rose and Cimino [**73**] and Rose and Goldfarb [**74**], demonstrating that an increase in contact stress results in an increase in the wear of UHMWPE. Wang *et al.* [**29**] stated that contact stress alone is not the dominant factor for polyethylene wear. These workers agreed with Barbour *et al.* [**28**] that the physiological and kinematic representation during wear testing are very important; otherwise different wear mechanisms and inaccurate or contradictory wear rates may arise. The influence of contact stress on the wear of UHMWPE has been somewhat misleading, due to different test conditions between centres.

It is agreed that the quality of UHMWPE is an important factor influencing the wear rate. UHMWPE undergoes various physical and mechanical changes from fabrication to clinical implantation [**22**, **23**, **57**–**60**]. From a fabrication point of view, fusion defects must be kept to a minimum, as cyclic loading may initiate these fabrication defects into microcracks, which may further propagate towards the surface, resulting in delamination. The current method of sterilization is gamma irradiation in inert gas. The recent introduction of this technique is expected to reduce ageing effects and oxidation and may lead to reduced wear rates. Apart from total knee simulation, many researchers have represented wear and factors affecting wear of polyethylene using simple test rigs. The kinematic conditions of the knee are among the most complex in the body and changes in physiological representation can lead to contradictory wear rates [**28**, **72**]. The dominant motion of the knee is flexion, resulting in combined cyclic loading, rolling and sliding between the bearing surfaces under a low conformity. Furthermore various degrees of rolling and sliding exist during a gait cycle and can be termed the slip ratio of the femoral component to the tibial component. The present authors agree with Barbour *et al.* [**28**] and Blunn *et al.* [**65**] that the kinematic conditions between TKRs is a dominant factor influencing the rate of wear and the type of wear mechanisms present. The influence of such kinematic conditions including various slip ratios is currently being investigated by the present authors at the University of Limerick, and a three-station test rig was built to evaluate these conditions and the effect of various slip ratios [**43**]. Workers such as Walker *et al.* [**55**], Cornwall *et al.* [**72**], Blunn *et al.* [**65**], Rose and Cimino [**73**], Rose and Goldfarb [**74**] and Rostoker and Galante [**75**] carried out laboratory tests under physiological conditions of the knee. Blunn *et al.* [**65**] concluded qualitatively that, under cyclic loading, gliding of a non-conformal metallic indenter on a flat polyethylene surface produced the greatest severity of wear. The results of Cornwall *et al.* [**72**] were in agreement with the work of Blunn *et al.* and showed quantitatively ascending wear factors for metallic sliding at a point on polyethylene, tractive rolling and gliding between metallic and polyethylene surfaces respectively. However, Andriacchi *et al.* [**69**] and Wimmer and Andriacchi [**80**] suggested that tractive forces generated during pure rolling can produce higher tangential forces than those occurring during sliding since tractive forces are dependent upon the static coefficient of friction rather than the lower dynamic coefficient of friction. Some mechanical and tribological factors necessitating extended research are the true effect of various contact geometries and conformity on the wear of polyethylene under physiological knee conditions and the proper knowledge of lubrication conditions. Tests on the wear of polyethylene in the knee must be distinguished clearly from those on the wear of the hip and other joints. A complete understanding of the physiological and kinematic conditions of contemporary prosthetic knee implants *in vivo* is very necessary when attempting to reproduce wear rates and determining factors affecting detrimental wear in laboratory tests. Otherwise investigations from different research groups may become ambiguous and contradictory.

To date, there are few theoretical mechanisms relating to the wear of UHMWPE in TKRs [**81**–**83**]. Descriptive wear mechanisms and processes for wear in TJRs in general have been provided by a number of researchers [**29**, **32**, **34**, **40**, **41**, **44**, **65**, **71**, **76**, **77**, **84**]. The complexity of the wear process in TKRs is demonstrated by the lack of a coherent theoretical model and consistent data from these researchers. There are many factors affecting wear rates, and hence many wear mechanisms observed from different factors. Under laboratory tests, it has been shown by Wang *et al.* [**29**], Blunn *et al.* [**65**] and Bragdon *et al.* [**71**] that the wear rate and wear mechanisms for prosthetic implants depend strongly on material quality and properties of UHMWPE and the physiological representation during the tests. Other workers have described counterface abrasion or particle formation from shearing of the polymer surface by microscopic counterface asperities, and polymer adhesion to the counterface by the fibular pull-out mechanism from loose polymer particles. However, the process of polymer adherence is still unclear. Blunn *et al.* [**65**] and Saiko [**78**] showed that high subsurface strains of the polymer asperities may lead to cracking, delamination and pitting of the polymer. From this review there are three dominant wear mechanisms or processes that produce wear debris in contemporary knee prostheses. These are (a) the microscopic counterface asperity wear process, (b) the macroscopic polymer asperity wear process and (c) structural subsurface failure [**41**].

5 CONCLUSIONS

The formation and development of wear are now widely accepted as a major concern in the long term survivorship of contemporary knee prostheses *in vivo*. The generation of polyethylene wear debris has been linked to osteolysis, which can result in bone resorption and eventual implant loosening.

The multi-factorial nature of knee wear has resulted in many complex factors that influence the development of polyethylene wear *in vivo*. *In vitro* tests have been carried out to investigate these factors. Some of these factors are the topography conditions of the articulating surface components, the lubricating medium during *in vitro* testing, the presence of TBD, the loading conditions, the design of components, the quality of UHMWPE from starting powder to clinical evaluation and the kinematic conditions between the components. Considering this range of factors influencing the wear rate of UHMWPE, further theoretical and experimental modelling needs to be carried out to allow prediction of *in-vivo* behaviour of UHMWPE in TKRs.

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