The current state of bearing surfaces in total hip replacement

A. Rajpura, D. Kendoff, T. N. Board

From The Centre for Hip Surgery, Wrightington Hospital, Wigan, United Kingdom

We reviewed the literature on the currently available choices of bearing surface in total hip replacement (THR). We present a detailed description of the properties of articulating surfaces and review the understanding of the advantages and disadvantages of existing bearing couples. Recent technological developments in the field of polyethylene and ceramics have altered the risk of fracture and the rate of wear, although the use of metal-on-metal bearings has largely fallen out of favour, owing to concerns about reactions to metal debris. As expected, all bearing surface combinations have advantages and disadvantages. A patient-based approach is recommended, balancing the risks of different options against an individual's functional demands.

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Anatomy of the ‘perfect bearing’

The perfect THR bearing is an articulating surface that has virtually no wear, with a sufficiently large head to reduce the risk of dislocation. Any debris produced should not evoke a host immune response. The bearing materials should exhibit low friction to reduce forces on both the implant–bone interface and the modular head–neck interface and no noise should be generated. The material properties of the head and liner/acetabular components should be chemically stable in vivo, tough enough to reduce the risk of fracture, and hard and non-ductile so as to reduce any susceptibility to scratching and third-body wear.

Articulating surface wear

Adverse consequences of bearing wear can be either catastrophic implant failure or local host biological responses resulting in osteolysis or soft-tissue reactions. Catastrophic wear was originally demonstrated by Charnley's attempts at using metal-on-polytetrafluoroethylene bearings. This type of failure is now relatively uncommon, but can be associated with component malpositioning, impingement or polyethylene oxidation. Host biological reactions to wear debris and particle-driven aseptic loosening remain the most frequent causes of revision.

Not only should the volume of wear debris generated be considered, but also the biological activity of the individual particles. Fisher et al developed the concept of the ‘Functional Biological Activity’ (FBA) index for polyethylene wear debris, and defined this as the product of the volumetric wear and the specific biological activity (SBA) of the wear debris as determined by macrophage stimulation tests and particle concentration.

Factors that therefore need to be considered are the bearing couple itself, which would affect both the amount of volumetric wear and the SBA of the wear debris, and the effect of head size. Head size can affect the volumetric wear generated owing to its influence on lubrication conditions and sliding distance, and must be balanced against increasing stability obtained with larger sizes.

Hard-on-soft couples.

Hard-on-soft bearings consisting of metal on conventional non-cross-linked ultra-high molecular weight polyethylene (UHMWPE) have shown high amounts of wear in both laboratory and clinical studies. Simulator data have demonstrated volumetric wear rates of 23.2 mm³/million cycles and 32.8 mm³/million cycles for 22.25 mm and 28 mm heads, respectively. Patients with wear rates > 0.1 mm/year, equating to a volumetric wear rate of 38.8 mm³/year, have a high risk of revision.

Efforts to reduce the volumetric wear of UHMWPE have led to the use of cross-linked polyethylene (XLPE) acetabular components and ceramic femoral heads. A 30% reduction in volumetric wear was seen when moderately cross-linked polyethylene was compared with...
conventional UHMWPE. Ceramic femoral heads have been used as they have a lower surface roughness than metallic alloy heads. Early simulator data showed a 20-fold reduction in volumetric wear for ceramic versus metallic heads. This validity of this study has been questioned, however, owing to the use of water as the lubricant. Subsequent simulator studies using lubricant with physiological concentrations of protein have shown a 50% reduction in wear for ceramic versus metallic heads. Clinical studies of wear using linear penetration have rate of wear between 1.7 and 4 times higher for metal-on-polyethylene (MoP) than for ceramic-on-polyethylene (CoP).

Increasing head size increases the sliding distance of the bearing and therefore potentially increases the volumetric wear. Livermore, Illstrup and Morrey demonstrated this in vivo using radiological methods to compare volumetric wear in 22 mm, 28 mm and 32 mm MoP THR and a 74% increase in volumetric wear was noted when the 28 mm and 32 mm bearings were compared. Using hip simulators, Clarke et al found a linear increase in volumetric wear of 7.8% per millimetre increase in head diameter in MoP bearings. Hall et al analysed retrieved loose MoP acetabular components and found an increase in volumetric wear of 5.1 mm³/year for each millimetre increase in the radius of the head. However, hip simulator studies have shown no increase in wear related to increasing head size for highly XLPE, even when using 46 mm heads. In vivo studies however are somewhat conflicting. Two short term follow-up studies (three years) have shown no difference in linear wear rate of XLPE with increasing head size, volumetric wear was not reported. A more recent medium term study with a mean 5.7 year follow-up has demonstrated increased volumetric wear of highly XLPE with larger 36 mm and 40 mm head sizes compared with 26 mm, 28 mm and 32 mm head sizes. Therefore in vivo behaviour of XLPE may not be accurately represented by laboratory studies and longer-term studies regarding the outcome of large heads on XLPE are needed. This dichotomy is also seen in retrieval studies of XLPE, where unexpectedly high levels of surface damage were noted, which is not predicted by simulator data.

### Hard-on-hard couples
Volumetric wear is inversely proportional to the hardness of the softest bearing surface. This has led to the development of hard-on-hard bearings using combinations of metallic and ceramic bearing surfaces. Smaller radial clearances, smooth surfaces and larger radii may theoretically allow fluid-film lubrication in hard-on-hard bearings, thereby improving wear properties.

Metal-on-metal (MoM) bearings have shown significant in vitro reductions in wear rates compared with conventional MoP bearings. Hip simulator data have shown volumetric wear rates ranging from 0.2 mm³/million cycles to 2.3 mm³/million cycles for 28 mm MoM bearings. This compares favourably with volumetric wear rates of 32.8 mm³/million cycles and 9 mm³/million cycles for 28 mm metal-on-UHMWPE and highly XLPE bearings, respectively. Simulator data also support the use of smaller radial clearances, with slightly lower wear rates noted with sub-50 μm radial clearances.

Ceramic-on-ceramic (CoC) bearings have even lower volumetric wear rates, with steady-state wear rates as low as 0.004 mm³/million cycles for 28 mm alumina-on-alumina bearings. Other investigators have noted wear rates of between 0.05 mm³/million cycles and 0.1 mm³/million cycles for 28 mm alumina bearings, but this still represents a significant reduction compared with MoM bearings. A clinical study evaluating the performance of 28 mm CoC bearings has shown a mean annual linear wear rate of 6.7 μm/year.

A ceramic-on-metal (CoM) bearing was developed based upon the principle of differential hardness. Available data are limited, but early simulator data report volumetric wear rates of 0.01 mm³/million cycles for the CoM couple compared with 1.23 mm³/million cycles for a MoM couple. Wear behaviour in adverse conditions. Conditions in vivo can differ significantly from standard gait conditions studied in hip simulators. Therefore, the effect of conditions such as damage to femoral heads, third-body wear, microseparation and edge loading, which have all been noted in vivo, need to be considered.

Despite the superior wear performance of XLPE versus conventional UHMWPE under standard conditions, concerns have been raised regarding its behaviour against roughened surfaces or third-body wear. Sakoda et al examined the performance of chemically induced XLPE against scratched interfaces and found a 30-fold increase in the rate of wear compared with only a threefold increase for conventional UHMWPE under similar conditions. Others have reported a greater sensitivity of irradiated XLPE against roughened femoral heads, with a nine-fold increase in wear for XLPE and only a 5.1-fold increase for conventional UHMWPE, with the overall result being a similar absolute volumetric wear rate for both varieties of polyethylene.

In the presence of third-body polymethyl methacrylate debris, XLPE was found to have an 80-fold increase in wear in a hip simulator model compared with only a six-fold increase for conventional UHMWPE.

In contrast, McKellop et al found that XLPE had better wear resistance to roughened interfaces than conventional UHMWPE but this was observed in the presence of higher than physiological concentrations of protein in their lubricant, with the acetabular component positioned under the head in the simulator, which may have protected against the roughened head. It is difficult to draw conclusions about the true behaviour of XLPE in adverse conditions, but it may be sensible to use ceramic heads with XLPE because of their greater scratch resistance.

In vivo, joint laxity can allow the separation of the bearing surfaces during swing phase, with concentric relocation on heel strike. This microseparation is usually only a few...
millimetres, but can result in an abnormal contact of the head with the rim of the acetabular bearing surface at heel strike, resulting in edge loading. Retrieval analysis of second- and third-generation alumina CoC bearings has shown characteristic stripe wear patterns and median volumetric wear rates of 1.2 mm³/year, which is significantly greater than that predicted by conventional simulator studies. Microseparation has now been simulated in vitro, and more clinically relevant wear rates of between 1.2 mm³/million cycles and 1.84 mm³/million cycles have been seen for third-generation alumina CoC bearings. This still represents a significant wear advantage over conventional MoP bearings. With the introduction of Biolox Delta ceramic (CeramTec, Plochingen, Germany) bearings, very low wear rates of 0.11 mm³/million cycles and 0.16 mm³/million cycles have been observed, even under severe microseparation conditions and independent of socket abduction angle.

Edge loading and microseparation in MoM significantly accelerate wear. Retrieval studies of MoM bearings have also demonstrated 22- to 27-fold increased wear rate in edge-loaded components. Therefore MoM bearings seem to be more sensitive to edge loading than CoC bearings, and this is thought to be responsible for the accelerated failure of some hip resurfacing designs.

Microseparation is seen to a greater extent in hard-on-soft bearings, yet its effect on wear is opposite to that seen in hard-on-hard bearings. Williams et al noted an almost five-fold reduction in the volume of wear when simulating microseparation in a hip simulator with CoP bearings compared with non-microseparation. A review by Harris concluded that edge loading is of little consequence in hard-on-soft bearings.

Biological effects of wear debris

Polyethylene debris. Polyethylene wear debris extracted from explanted tissue can range from sub micrometre to several millimetres in size, with most in the sub micrometre range. Particles measuring < 0.5 μm in diameter induce the strongest macrophage response and subsequent cytokine cascade leading to aseptic loosening. The morphology of wear debris differs between UHMWPE and XLPE, where the latter has been shown to have a higher proportion of debris in the 0.1 μm to 0.5 μm range (88% vs 68% for non-cross-linked UHMWPE). As a result, the SBA for XLPE was approximately 50% higher. It is reported that 0.1 μm³ of XLPE debris per cell is needed to increase the level of tumour necrosis factor (TNF)-α significantly at 24 hours, whereas 10 μm³ of the UHMWPE debris is needed to stimulate a response.

Using a murine model, 10 Mrad XLPE and UHMWPE particles of identical size were implanted under the peristeum of calvaria and the percentage area of osteolysis subsequently induced was measured. A significantly greater osteolytic response was noted in the XLPE group (35% vs 9% area of osteolysis). Cross-linking in itself therefore may increase the inflammatory potential of the wear debris, as an independent effect from the altered size distribution of the size of the wear particles.

These studies suggest that the superior wear properties of XLPE may be somewhat offset by the greater osteolytic potential of its wear debris.

Ceramic wear debris. Analysis of explanted tissues from CoC THR has a bimodal distribution of particle sizes, with particles ranging from 5 nm to 90 nm in diameter, and larger particles similar to UHMWPE debris ranging in size from 0.05 μm to 3.2 μm. Ceramic wear particles with this size distribution of size have been tested in vitro using human macrophages, and have been shown to be significantly less active than UHMWPE debris, with a minimum volume of 100 μm³ needed to induce TNF-α production.

Considering the improved wear characteristics of CoC bearings, this volume is unlikely to be reached in vivo.

Metallic wear debris. Doorn et al found particles ranging in size from 6 nm to 744 nm, with a mean of 42 nm in tissues from around MoM prostheses. They estimated that between 6.7 x 10¹² and 2.5 x 10¹⁴ particles per year would be produced, which is between 13 and 500 times that of an average MoP bearing. Therefore, despite a reduction in wear volume, the overall surface area may be greater, and therefore concerns have been raised regarding a potentially greater response per unit volume.

Explanted tissues from failed MoM THRs have a lymphocyte-dominated cellular response in contrast to MoP implants, which have a greater macrophage/giant cell-dominant response. Giant cells are rarely seen in conjunction with MoM THRs. Willert et al termed this phenomenon aseptic lymphocyte-dominated vasculitis-associated lesions (ALVAL). Macroscopically severe soft-tissue reactions have been noted in conjunction with MoM THRs, causing premature failure. The aetiology of such reactions is still unknown, but metal ions acting as haptons inducing an immune response have been cited as a potential mechanism.

Distant dissemination of metal debris and chromosomal aberrations in peripheral blood lymphocytes have been noted. Epidemiological studies have not established an increased cancer risk associated with the use of MoM implants. Exposure to CoCrMo particles has also shown a reduction in fibroblast and macrophage viability.

Clinical outcomes

A meta-analysis by Kuzyk et al comparing XLPE with conventional UHMWPE demonstrates a risk ratio of 0.4 for radiological evidence of osteolysis favouring XLPE, with a respective overall incidence of 9.7% versus 23.8%. Kurtz, Gawel and Patel’s systematic review confirmed a favourable risk ratio of 0.13 when only studies with more than five years’ follow-up were included; however, ten-year follow-up studies are lacking, and a reduction in the risk of osteolysis is not established for heads > 32 mm against XLPE.

Medium to long-term data for CoC bearings demonstrate an even lower incidence of osteolysis. Data for
184 hips with third-generation CoC bearings at ten years revealed only 1.6% of acetabular and 2.7% of femoral components with radiolucentcies.\textsuperscript{76} Minimum 20-year follow-up data for 85 hips with second-generation CoC bearings showed a 1.2% and 7.1% incidence of radiolucentcies measuring > 2 mm around the femoral and acetabular components, respectively.\textsuperscript{77} A comparative study of CoC versus MoM bearings with a mean eight-year follow-up demonstrated osteolysis in 1.4% and 30.5% of hips with CoC and MoP bearings, respectively.\textsuperscript{78}

In terms of osteolysis, MoM bearings have displayed low rates comparable with those of CoC bearings, with reported rates ranging between 0% and 3%.\textsuperscript{79-84} However, the incidence of soft-tissue reactions related to MoM bearings has raised concern. Reported rates of adverse reactions to metal debris (ARMD) for the 28 mm Metasul bearing range from 0% to 5%.\textsuperscript{85} Higher rates of soft-tissue reactions have been noted for large-diameter heads, especially when used in stemmed THRs.\textsuperscript{86} Designs with a sub-hemispherical design and low radial clearances have a higher rate of failure.\textsuperscript{87} This may be due to the increased risk of edge loading and increased wear from equatorial bearing secondary to acetabular component deformation.\textsuperscript{54,88}

With regard to CoM bearings, only short-term follow-up data exist.\textsuperscript{89,90} The largest series has demonstrated no difference at 12 months in serum levels of cobalt and chromium compared with MoM bearings of a similar size, despite their superior performance \textit{in vitro}.\textsuperscript{89,90}

**Friction of the bearing surface**

Cadaveric studies have shown torque values between 6.8 Nm and 46 Nm are capable of causing sudden loosening of uncemented acetabular components, and 170 Nm for cemented implants with pegs.\textsuperscript{91} Minimising frictional torque was the basis of the Charnley LFA, and this was primarily achieved by maximising the difference between the diameter of the head and the external diameter of the acetabulum.\textsuperscript{92}

Lubrication is essential in reducing friction. If fluid-film lubrication can be achieved, then potentially friction could be significantly reduced, as this would depend only upon shearing of the lubricant film separating the two bearing surfaces.\textsuperscript{93} Lubrication theory states that fluid-film lubrication can be achieved if hard bearing surfaces with smooth surfaces and tight radial clearances are used.\textsuperscript{94} Experimental studies have unfortunately shown this not always to be the case.\textsuperscript{26,94}

The friction factors of the available bearing couples have been studied \textit{in vitro} using hip simulators. Using 28 mm bearings and 25% calf serum as the lubricant, Brockett et al\textsuperscript{95} found CoC to exhibit the lowest friction factor (0.04), followed by CoM (0.05), then CoP (0.055), then MoP (0.06), and finally MoM with the highest friction factor of 0.12.\textsuperscript{95} Other investigators have explored friction factors in 28 mm bearings.\textsuperscript{96,97} Only the CoC and CoM bearings were shown to be capable of full fluid-film lubrication, with the MoM, CoP and MoP bearings all working in a mixture of ways.\textsuperscript{95-97} Interestingly, XLPE has been shown to generate higher frictional torque than UHMWPE over a range of bearing sizes;\textsuperscript{98} the reason for this is unknown.

Large-diameter MoM bearings (50 mm) have demonstrated higher frictional moments of 7.9 nm \textit{versus} 4.6 nm for 28 mm MoP bearings under the same test conditions.\textsuperscript{99} This is now within the range of the torque values required to cause loosening of uncemented implants, albeit at the lower end of that range. Such frictional torques may, however, play a role in fatigue failure, leading to loosening, or cause failure in poorly osteo-integrated implants.\textsuperscript{97-99}

The influence of the interruption of movement has also been studied using hip simulators.\textsuperscript{91,100} The static friction of MoM bearings after being stationary for 60 seconds increases by 250%, compared with only a 50% increase in MoP bearings. This is thought to be due to interactions between decomposed proteins that have adhered to the bearing surfaces.\textsuperscript{91} Therefore, if a 50 mm MoM bearing is used, then peak frictional torques of 15 nm may be generated upon initiating movement after rest, predisposing to either implant fixation failure or taper junction failure.

The effects of adverse conditions such as edge loading and third-body inclusions that may be encountered \textit{in vivo} also need to be considered. Under edge-loading conditions simulated by a 75° acetabular component inclination angle, Sari-ali et al\textsuperscript{101} demonstrated a 4.3-fold ($\mu = 0.085$) increase in friction factor for 32 mm CoC bearings. The addition of third-body particles caused a 16-fold ($\mu = 0.32$) increase with the introduction of alumina powder and a 26-fold ($\mu = 0.52$) increase when a 2 mm $\times$ 0.1 mm chip was introduced. Bishop et al\textsuperscript{99} investigated the effect of poor lubrication in large-diameter CoC and MoM bearings and found that 48 mm CoC bearings run in serum showed a two-fold increase in friction factor, with a 60° acetabular component inclination. In contrast, 32 mm CoC and 50 mm MoM bearings showed no increase in friction factor with increasing cup inclination. When run in dry conditions to simulate a potential worse case \textit{in vivo} scenario, 48 mm CoC bearings showed a 10-fold increase in friction factor ($\mu = 0.58$), with a 5 and 1.7 times increase ($\mu = 0.49$, $\mu = 0.28$) noted for 32 mm CoC and 50 mm MoM bearings respectively. CoC bearings, especially the newer larger diameter CoC bearings, seem therefore to display a greater sensitivity to adverse conditions.

**Bearing-generated noise**

Squeaking is a phenomenon unique to hard-on-hard bearings, with a greater incidence in CoC than in MoM.\textsuperscript{102} Squeaking in MoM bearings appears to be a self-limiting condition, with most cases noted within the first six months of surgery and none persisting past two years.\textsuperscript{103,104} This may be related to the bedding-in period of MoM bearings, after which lubrication improves the performance of the bearing and the squeaking resolves.\textsuperscript{103} In contrast, squeaking in CoC bearings presents later and can be persistent, and is a potential indication for revision.\textsuperscript{105,106} A meta-analysis by Stanat and Capozzi\textsuperscript{107} revealed a mean incidence of 2.4% (0.7% to 20%) for CoC bearings.
Increased friction due to adverse tribological conditions leading to suboptimal lubrication is suggested as the root cause of squeaking.\textsuperscript{108} This can be due to a variety of causes including edge-loading, third body ingress and subluxation. Clinically, raised body mass index (BMI) and the use of a Stryker Accolade stem (Stryker, Mahwah, New Jersey) are risk factors for squeaking.\textsuperscript{107} Walter et al\textsuperscript{109} found a correlation with acetabular positioning outside a safe zone of $+45^\circ/-10^\circ$ inclination and $+25^\circ/-10^\circ$ anteverision safe zone. This may be due to the increased risk of edge loading outside this orientation.

Stripe wear secondary to edge-loading is a consistent finding in retrieved squeaking implants.\textsuperscript{110} Small medium-term follow-up studies suggest no adverse consequence of squeaking in terms of osteolysis or implant failure.\textsuperscript{111,112} However, due to its relationship with increased friction, longer-term studies are needed to rule out its potential to induce frictional torque related problems such as taper wear or prosthetic loosening.\textsuperscript{113,114}

**Material properties of the head and liner**

**Degradation in vivo.** Oxidation, corrosion and phase transformation are processes that *in vivo* can lead to degradation of bearing components. Oxidation is primarily a problem related to UHMWPE. Free radicals generated from either sterilisation or cross-linking processes using gamma or electron beam irradiation drive the oxidation process. This results in chain disruption, leading to a reduction in mechanical properties such as strength, ductility and wear resistance.\textsuperscript{115} Modern techniques of sterilisation in inert environments have reduced the ‘on-shelf’ oxidation of implants, but residual free radicals can still cause in vivo oxidation once implanted.\textsuperscript{116}

Cross-linking processes that use higher doses of radiation in the range 5 to 10 Mrad, generate greater amounts of free radicals. Post-irradiation processing, involving either annealing or re-melting, is therefore used to try and eliminate any residual free radicals.\textsuperscript{117} Re-melting is more effective but can also lead to a reduction in crystallinity, increasing the risk of fracture.\textsuperscript{118} Despite these efforts, both annealed and re-melted retrieved XPLE liners have shown in vivo oxidation.\textsuperscript{119,120} Annealed XPLE undergoes greater oxidation, particularly at the rim, with half the specimens showing severe oxidation to the level that may significantly impair the liners mechanical properties.\textsuperscript{119} Remelted XPLE liners show lower levels of oxidation, but in contrast to annealed XPLE, oxidation was noted at the bearing surface.\textsuperscript{119} The long-term consequences of these oxidative changes are unknown.

Second-generation XLPE using vitamin E as an antioxidant has been developed as an alternative to post-irradiation annealing or re-melting.\textsuperscript{120} The rationale is two-fold: first, to improve the oxidative stability compared with annealing or re-melting; and secondly, to avoid the reduction in fatigue strength caused by first-generation techniques.\textsuperscript{121} In vivo studies have demonstrated significant oxidative resistance afforded by either vitamin E diffused or blended UHMWPE, with diffused polyethylene demonstrating no detectable oxidation at three years of ageing.\textsuperscript{122-124} In vivo data is, however, awaited.

Ceramic components, specifically zirconium, can undergo in vivo phase transformation.\textsuperscript{113} Zirconium exists in one of three crystalline phases and changes from one phase to another, which can be caused by temperature and the presence of water, resulting in changes in volume and surface roughness. Such phase transformations can lead to significant increases in surface roughness of zirconium heads resulting in increased polyethylene wear.\textsuperscript{113,125} Zirconium implants have therefore been withdrawn from the market. Alumina implants, in contrast, do not undergo phase transformation. Fourth-generation ceramics consisting of alumina composites do contain 17% zirconium, but phase transformation is inhibited by the stabilising effect of the surrounding alumina particles, and therefore no increases in surface roughness are seen in vivo.\textsuperscript{126-128}

**Risk of fracture.** Fractures have been described in XLPE liners\textsuperscript{129-131} and in both ceramic heads and liners. XLPE has a higher fracture risk than conventional UHMWPE owing to radiation-induced reduction in ductility, which occurs in a dose-dependent manner.\textsuperscript{115,132-135} Post-irradiation processing may also affect the risk of fracture, as annealing preserves fracture resistance better than re-melting.\textsuperscript{136} Retrieval studies have also shown a higher incidence of rim cracks, thought to be precursors of complete fractures, in XLPE (15%) versus conventional UHMWPE (3%).\textsuperscript{24}

Fracture rates of up to 13.4% were reported for ceramic heads manufactured prior to 1990.\textsuperscript{132} However, with manufacturing by hot isostatic pressing, improvements in materials leading to a reduction in grain size and increased density, as well as improvements in Morse taper fixation, the rate of fracture has fallen to 0.004% for third-generation alumina.\textsuperscript{137-139} Fourth-generation alumina composite ceramics have further reduced risk of fracture to 0.002% by introducing zirconium particles and strontium oxide platelets, which help prevent the initiation and propagation of cracks.\textsuperscript{138,140}

**Surface hardness and scratch profile.** Ceramics have the greatest surface hardness, and both in vitro tests and explanted ceramic heads exhibit the least amount of surface damage from third-body particles compared with metal alloy heads.\textsuperscript{44,114} The scratch profile is also more desirable, as little pile-up is seen in comparison with metal heads.\textsuperscript{117} MoM bearings, however, can self-correct, as scratches can be polished out with further movement owing to the ductile properties of the metal.\textsuperscript{141}

The scratching vulnerability of XLPE versus conventional UHMWPE liners has been compared in vitro.\textsuperscript{142} Using visual inspection, there was a weak association with increased severity of scratching of XLPE liners. Scratch height and width were also slightly higher for XLPE liners.\textsuperscript{142} These findings are consistent with retrieval studies.
of XPLE liners that showed similar surface damage when compared with conventional UHMWPE.\textsuperscript{23,24}

In an effort to combine the surface hardness and desirable scratch profile of ceramics with the resistance to fracture of metal, oxidized zirconium (OxZr) (Oxinium, Smith & Nephew, Memphis, Tennessee) has been developed.\textsuperscript{143} Originally developed for the femoral components of the knee arthroplasties, a thermally driven oxidation process is used to transform the surface of zirconium–niobium alloy femoral heads into zirconium. In vitro tests of surface hardness have shown that the OxZr surface is more than twice as hard as a CoCr head.\textsuperscript{143,144} Wear data from hip simulator studies demonstrate similar performance to monolithic ceramic heads.\textsuperscript{145} However, an early in-vivo radiostereometric study failed to demonstrate any advantage of using an OxZr versus a CoCr head when articulating against a polyethylene acetabular component.\textsuperscript{146} Concerns have also been raised regarding the in vivo vulnerability of the OxZr surface to damage.\textsuperscript{147,148}

Early retrieval studies following open reductions for recurrent dislocation have reported significant surface damage with exposure of the underlying metal substrate. Deeper scratches were noted compared with retrieved CoCr heads. This may be due to the reduced hardness of the underlying OxZr substrate. The retrieved heads demonstrated a 50-fold increase in wear compared with a new head. McCalden et al\textsuperscript{147} retrieved a head after only 48 hours following a revision for leg length discrepancy and even after such a short time noted significant surface damage.

**Novel bearings**

Conventional bearing surfaces operate predominantly within a mixed lubrication environment, with hard-on-hard bearings able under certain conditions to exhibit fluid-film lubrication.\textsuperscript{95} The natural hip joint, in contrast, can operate within a full fluid-film lubrication environment for most of the time because it has a compliant articular cartilage bearing surface. This allows a combination of elastohydrodynamic, microelastohydrodynamic and squeeze-film lubrication to occur, resulting in an exceptionally low coefficient of friction of 0.01.\textsuperscript{149} In order to emulate the natural joint, compliant soft layer bearings have been developed.

Hydrophilic polyurethane (PU) has a similar tensile modulus to articular cartilage and has been used to produce a compliant soft acetabular liner.\textsuperscript{150} Hip simulator data have shown friction factors of < 0.01 for PU bearings articulating against a metal head; significantly better than conventional bearing couples; and are suggestive of full fluid-film lubrication.\textsuperscript{150} Under dry conditions, however, friction factors of 1.0 were measured, which would cause structural failure of the component.\textsuperscript{150} Start-up friction factors of 0.55 following 20 minutes of inactivity were also seen, but this rapidly falls within half a walking cycle and fluid-film lubrication is reinstituted.\textsuperscript{150}

Wear data from a hip simulator study up to 20 million cycles have shown a volumetric wear rate of 7.7 mm\textsuperscript{3}/million cycles, comparable with XLPE but inferior to hard-on-hard couples.\textsuperscript{151} Particle generation has been measured at $3 \times 10^6$ particles/million cycles, approximately five times lower than UHMWPE.\textsuperscript{151} Most particles also have a diameter of > 10 μm and are therefore unlikely to generate a macrophage response.\textsuperscript{151} Even particles with a diameter of < 10 μm exhibit less osteolytic potential than XLPE.\textsuperscript{152,153}

In vivo PU can degrade by hydrolysis, metal ion oxidation, stress cracking or calcification.\textsuperscript{154} Bionate PU (DSM Biomedical, Berkeley, California), previously known as Corethane, is the substrate used to produce the currently available liners. Bionate has demonstrated good resistance to biodegradation in both in vitro and in vivo ovine studies.\textsuperscript{154,155} Case reports of single explanted liners have also shown minimal degradation in vivo, comparable with experimental studies.\textsuperscript{156,157} Despite the encouraging experimental data, clinical data are sparse and conflicting. Two short-term clinical studies have shown revision rates of between 10% and 30%.\textsuperscript{158,159}

Another novel bearing material is carbon-fibre-reinforced polyether etherketone (CR-PEEK). The incorporation of carbon fibres into PEEK allows the production of material with a Young’s modulus that closely matches that of cortical bone, while also being sufficiently strong to allow the production of thin acetabular components.\textsuperscript{160} These properties were used to produce a novel thin horse-shoe-shaped acetabular component that would be bone conserving and also allow sufficient flexion to avoid stress shielding, seen with contemporary acetabular components.\textsuperscript{161,162} The initial design, the Cambridge Cup (Howmedica International Inc, Limerick, Ireland), was a composite of a UHMWPE bearing surface with a CR-PEEK backing with a hydroxyapatite coating.\textsuperscript{160,162} This design evolved into an acetabular component made of CR-PEEK, with a titanium–hydroxyapatite coating on the non-articular surface to encourage osteo-integration, and was articulating against a ceramic head.\textsuperscript{163} The ability to incorporate the fixation surface directly onto the bearing surface is another unique property of CR-PEEK.

Hip simulator wear studies for the CR-PEEK–ceramic couple have shown mean rates as low as 0.3 mm\textsuperscript{3}/million cycles over a 10 million cycle period.\textsuperscript{164} This is comparable with MoM bearings, and is significantly better than conventional hard-on-soft couples. Friction factors are, however, significantly higher than those of conventional bearings, with values ranging from 0.23 to 0.35 even in lubricated conditions.\textsuperscript{164,165} Particles generated from a CR-PEEK bearing have a mean size of 0.1 μm\textsuperscript{166} and show no cellular cytotoxicity in cell culture experiments;\textsuperscript{167} CR-PEEK particles demonstrate a similar inflammatory potential to those from UHMWPE.\textsuperscript{168} The clinical results of the original Cambridge Cup design were encouraging.\textsuperscript{169} This design, however, had a UHMWPE bearing surface. Short-term results of the evolution of this design with a CR-PEEK bearing surface are, however, inferior. At three years, five of 50 patients showed signs of acetabular migration and calcar resorption, with three subsequently undergoing
revised. The authors are investigating the causes of these early failures. Potential issues include high frictional torque causing failure or taper wear also as a result of the high friction factors.

Conclusions
All combinations of bearing surface have advantages and disadvantages. Some have recently fallen out of favour, particularly MoM and CoM. In assessing the best bearing surface for an individual patient we feel it is necessary to analyse the specific implications for each patient. For example, the unknown, long-term performance of XLPE may not be of relevance to a patient with a life expectancy of <15 years, particularly if used with ‘forgiving’ head size such as 28 mm or 32 mm. However, the same issue is of paramount importance to a 35-year-old otherwise normal patient in whom long-term wear of the bearing surface is arguably the biggest issue. In this patient, the trade-off between long-term low wear from a CoC bearing may outweigh the small risk of fracture and squeaking. The perfect bearing surface remains an enigma.

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