Metal-on-metal bearings surfaces: materials, manufacture, design, optimization, and alternatives

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Abstract: When first introduced, total hip replacements offered pain relief and improved mobility in elderly patients. The success of this procedure in terms of long-term durability and restoration of function has led to its use in younger, more active patients. This has resulted in a commensurate increase in patient expectation regarding longevity and the degree to which function and lifestyle is restored.

The bearing surface is a key feature of the performance of replacement joints. It is generally accepted that excessive amounts of wear debris preclude their long-term survivorship and hence there is an ongoing requirement for bearing surfaces which minimize debris generation. The purpose of this paper is to review the factors which affect the performance of so-called metal-on-metal bearings, to compare their performance with that of the other commonly used contemporary alternatives, metal and ceramic articulating against highly cross-linked polyethylene, and ceramic-on-ceramic, and finally to consider the potential solutions offered by new developments such as ceramic-on-metal and coatings applied to metal-on-metal bearings.

Keywords: total hip replacement, wear, hard-on-hard bearings, metal-on-metal bearings, ceramics, polyethylene

1 INTRODUCTION

Total hip replacement is a very successful procedure that provides pain relief and restoration of mobility. There are no hard data; however, it is estimated that approximately one million such procedures are carried out per year worldwide. A critical component of a replacement hip joint is the bearing surface, and good wear performance is now seen to be necessary for the long-term survivorship and overall performance of the joint.

Currently, bearing surfaces for hip joints fall into two broad categories: firstly, the so-called 'hard-on-soft' combinations in which hard materials such as metal (sometimes coated) or ceramics articulate against polyethylene; secondly, the so-called hard-on-hard bearings in which the same hard components articulate against themselves. The inherent risk of introducing modified materials and new designs of bearing surface into clinical use has been counterbalanced by patient demand and changes in patient demographics. Replacement-hip patients are becoming younger, have higher expectations, are generally more active, may specifically wish to participate in impact sports, and finally have a greater longevity. All these factors are driving the search for new hip joint designs. Specifically, new bearing surfaces are being sought which give significant reductions in both the volume of wear debris generated and its effect on the body.

Renewed interest in metal-on-metal bearings has been fuelled by encouraging medium-term results when used with contemporary conventional hip replacement and the re-introduction of surface-replacement-type components for which a metal-on-metal system is currently the only practical solution. That both of these arthroplasty solutions are targeted at younger, more active patients makes it even more imperative to optimize performance because of the increased likelihood that these components are in...
situ for extremely long periods of time, potentially 30–40 years.

The aim of this paper is to review the development of metal-on-metal bearings, to examine the ways by which their performance may be optimized, to consider the advantages and disadvantages when compared with other currently available solutions, and finally to highlight the potential future solutions for wear reduction.

2 BEARING SURFACE HISTORY

Replacement hip joints were first used in significant quantities in the late 1940s. However, the first designs which have the appearance of modern devices are typified by the components developed by McKee in the UK in the mid-1950s. These consisted of a femoral stem with a large-diameter (35–40 mm) head, made from a cobalt–chromium alloy, articulating against a similar-sized acetabular cup made from the same material (Fig. 1). Similar devices were subsequently developed by Ring (in about 1962) and Muller (in about 1965). Such components were introduced with little or no testing and certainly nothing that simulated their performance as replacement joints. These devices tended to suffer from a high initial failure rate and their use declined. However, reviews of long-term results indicated a very low wear rate of surviving components (less than 8 μm/year) after 10–20 years [1]. This prompted renewed interest from the late 1980s onwards.

At the same time as the early work on hard-on-hard bearing surfaces, John Charnley was developing replacement hip joints based on the low-frictional torque principle, fixed in place with acrylic bone cement. A combination of science, determination, and serendipity led to the clinical testing by Charnley of high-density polyethylene articulating against stainless steel in 1962. This combination, initially using 22 mm stainless steel heads, became the benchmark for bearing surfaces producing very low volumes of wear debris. From the late 1960s this bearing surface entered general widespread use with low wear rates being reported, for example by Griffith et al. [2]. Refinements on this basic principle included heads of slightly larger diameter (28–32 mm) made from cobalt–chromium alloys, the introduction of ceramic heads, and a more advanced polyethylene called ultra-high molecular weight polyethylene. However, this design remained essentially unaltered until the early 1990s when longer-term problems became apparent with this bearing surface combination. These include an osteolytic reaction caused by polyethylene wear debris [3], susceptibility of gamma-irradiated polyethylene to oxidation [4], and the effect of roughened femoral heads [5, 6]. This led to the development of harder (ceramic) femoral heads and new highly cross-linked polyethylenes.

By 2000 there was an increasing belief that these advanced polyethylenes would minimize the adverse effects associated with wear debris but would not solve the problem entirely and alternative solutions were sought.

As discussed above, after a high early failure rate with first generation metal-on-metal devices, the surviving components had a relatively high level of long-term success. This led to renewed interest from the 1980s onwards and the introduction of re-engineered metal-on-metal bearing surfaces manufactured from cobalt–chromium alloys in conjunction with more conventional acetabular and femoral components. These have produced some encouraging results. For example, Wagner and Wagner [7] reviewed a series of 78 patients with a 5 year follow-up. There were three revisions all unrelated to wear. There was no evidence of metallosis, and no osteolysis apparent radiographically. Similar short- to medium-term results were recently reported by Jacobs et al. [8] and Lombardi et al. [9]. Sieber et al. [10] reported on the analysis of 118 retrieved second-generation metal-on-metal implants. Linear annual wear was found to be 25 μm in the first year, dropping to 5 μm after the third year, and was equally divided between head and cup. The self-polishing capacity of metal-on-metal joints was noted whereby scratches are worn away by further joint movement.

Ceramic-on-ceramic components had a similar history. They were first introduced by Boutin et al. [11] in the 1970s who in 1988 reported disappointing clinical results. However, a more encouraging outcome was reported by Mittelmeier and Heisel [12],

Fig. 1 McKee–Farrar prosthesis. (Image supplied by D. Dowson, Leeds University, UK)
which helped to re-establish the concept. More recently Bizot et al. [13] reported on a group of 234 patients with ceramic-on-ceramic components at an average follow-up of 7.8 years. There were 11 revisions none of which was related to wear. Radiographic analysis showed no signs of osteolysis, a conclusion also reached by Pignatti et al. [14] in their review of 123 ceramic-on-ceramic components.

These results have led to the more widespread use of these so-called ‘alternative’ or ‘hard-on-hard’ bearings.

Semlitsch and Willert [15] reviewed available clinical data. They noted the reduction in wear when polyethylene articulated against a ceramic compared with a metal. They also identified that a much greater reduction in clinical wear rate could be achieved by using both metal-on-metal and ceramic-on-ceramic bearings, as shown in Fig. 2.

3 CONTEMPORARY ISSUES WITH METAL-ON-METAL BEARINGS

3.1 Biological issues

Despite encouraging results there remain concerns with metal-on-metal bearings. These are related to the long-term effect of wear debris. The osteolytic effect of polyethylene debris is well known but the long-term effect of the debris from hard-on-hard bearings is less clear. The wear particles are much smaller (10–60 nm) than those from polyethylene (usually in the range of less than 0.1–10 μm) and it has been estimated that as a consequence, whilst there is significant reduction in wear volume, there may be an increase by a factor of 1000 in the number of particles [16].

The potential harmful effects of this debris, or more specifically the effect of the breakdown of the debris into metal ions, were first highlighted by Black [17] who indicated that ‘isolated clinical observations [of adverse carcinogenic, metabolic, immunological, and bacteriological effects] support the presence of systemic effects, especially associated with immune responses and metal overload and accumulation conditions’. Despite extensive searches for evidence of this relationship, most notably by Paavolainen et al. [18], no link has been found after 40 years of use. Willert et al. [19] recently reported on a group of patients who had early post-operative pain. The histological findings indicated few metal particles but were consistent with a possible lymphocyte-dominated immunological response. Because of these concerns it would still be prudent to try to minimize the volume of wear debris generated in metal-on-metal bearings.

3.2 Theory

The key to the performance of metal-on-metal bearings, is lubrication theory which identifies three lubrication regimes.

1. Boundary lubrication. This occurs by ‘slippery’ molecules chemically adhering to moving surfaces, in which there is direct asperity contact. Examples of boundary lubricants in the human body are phospholipids and glycoproteins.

2. Fluid-film lubrication. Conditions are such that an interposed fluid separates the moving surfaces.

![Fig. 2](https://example.com/fig2.png)  
*Fig. 2* Review of clinical wear rates with various bearing surfaces. (After Semlitsch and Willert [15])
An example of a fluid lubricant in the body is synovial fluid.

3. Mixed lubrication. The load is partially supported by a combination of contact with boundary lubricants (at the asperity tips) and by pressure developed in a fluid film that separates some, but not all, of the asperities of the interacting surfaces.

Ideally, for hard-on-hard bearings to work, full fluid-film lubrication is required but the service cycle of hip joints, the engineering materials available for use, and the variable properties of lubricants render this very difficult. However, evidence of satisfactory clinical performance of a hip joint bearing surface was reported by Clarke et al. [20] which, from the lubrication theory adapted for hip joint replacement proposed by Jin et al. [21], was calculated by Dowson et al. [22] to be operating in the mixed regime rather than in a regime of continuous fluid-film lubrication. The full benefits of fluid film can nevertheless be encouraged by the optimization of key manufacturing parameters.

The adapted lubrication theory proposed by Jin et al. [21] for hip joints identified the factors affecting their wear. This work indicated that wear would be reduced by using large-diameter components, which have a small diametral clearance between the femoral head and acetabular cups. The components should be as smooth as possible, which suggests that components should be as hard as possible since harder surfaces are more likely to achieve and maintain a very smooth surface in the human body. At first sight the inverse relationship between femoral head diameter and wear is at odds with the low-friction arthroplasty principle proposed by Charnley [23] and observed clinically by Livermore et al. [24] and Kabo et al. [25], which indicated that volumetric wear is directly proportional to femoral head diameter. The explanation for this apparent dichotomy is that all polyethylene bearings operate in the boundary lubrication regime in which wear is related to sliding distance, which increases with increasing femoral head diameter. Conversely, hard-on-hard bearings, if correctly engineered, have the potential to operate in the mixed and fluid-film lubrication regime where the increased head diameter leads to increased sliding speeds, which aids lubrication and hence reduces wear. Interestingly this state of affairs was correctly predicted by Charnley et al. [26] in 1969.

3.3 Materials
The effect of materials is unclear and, as discussed above, there is some dispute regarding their effect.

All metal-on-metal joints are cobalt–chromium–molybdenum alloys. It was reported as early as 1996 by Streicher et al. [27] that those alloys with a higher carbon content outperform those classified as low-carbon alloys. It is now generally accepted that this is the case. It has also been reported by Cawley et al. [28] that a superior wear performance was achieved with high-carbon alloys that have a microstructure consisting of coarse blocky carbides compared with other conditions. This was attributed to better abrasion resistance. It was further proposed by McMinn [29] that the use of heat-treated high-carbon alloys was responsible for early failures in some surface replacement components. Conversely Chan et al. [30] found no difference between cast and wrought high-carbon material. In addition, Bowsher et al. [31] found no difference between as-cast and cast and heat-treated material. Dowson et al. [32] reported a hip simulator study which compared the wear performance of several high-carbon cobalt–chromium–molybdenum alloys using 36 mm components with various manufacturing routes which produced extremes of microstructure. The microstructures varied from the as-cast condition with the aforementioned blocky carbides to the wrought condition with a fine dispersion of small carbides. No significant difference was found between any of these conditions, as illustrated by Fig. 3. This conclusion from simulator studies has important implications for the manufacture of metal-on-metal bearings, specifically that material choice is not dictated by wear performance but by other factors such as ease of manufacture and non-wear-related benefits. Therefore, for example, it makes sense for metal femoral heads of intermediate diameter (28–36 mm) to be made from wrought bar, this material condition lending itself to high-volume precision manufacture using modern computer numerically controlled technology. Conversely the intricate internal geometry required for the femoral portion of surface-replacement-type components lends itself to a cast route. The use of sintering to attach beads or a mesh for fixation naturally leads to a cast and heat-treated material for either femoral or acetabular single-piece components. On the other hand for modular acetabular components the fixation portion is separate from the bearing surface. In this configuration the former is usually forged (and possibly heat-treated depending on fixation methods). The latter would favour wrought bar for ease of manufacture.

3.4 Design optimization
The theories associated with metal-on-metal bearings have been tested in hip joint simulators and
were reported by Chan et al. [30], Scholes et al. [33], and Dowson et al. [22]; they indicated how the performance of metal-on-metal designs may be optimized. It was found that reducing the clearance between the femoral and acetabular components reduced the amount of wear debris generated, as shown by Fig. 4. A significant reduction \( p = 0.05 \) was noted for large diameters (54 mm) when clearances were reduced from 287 to 107 \( \mu \)m. A modest reduction in clearances (from 143 to 105 \( \mu \)m) with medium-diameter (36 \( \mu \)m) components also produced a significant reduction \( p = 0.05 \) in wear rates. A previous study by Smith et al. [34] determined the percentage of the loading cycle when the components were separated by a fluid film by measuring the voltage drop between the femoral and acetabular components. This showed that relatively small differences in clearance with 36 \( \mu \)m components resulted in no separation (170 \( \mu \)m) or separation for the entire loading cycle (130 \( \mu \)m). Fears have been expressed that reducing clearance may limit fluid entrainment and ultimately lead to lubricant starvation. Whilst component seizure was reported in early designs [1], these were short-term failures attributed to negative clearance which caused equatorial surface contact associated with high frictional torques and high wear. Late failures attributed to long-term wear processes have not been reported. There is no evidence to support these concerns in modern, appropriately engineered components. On the contrary, Clarke et al. [20], when reporting on the performance of conventional hip replacements used in conjunction with 28 \( \mu \)m bearings which had a diametral clearance of 60–80 \( \mu \)m, noted low ion levels when compared with larger-diameter (48 \( \mu \)m) components which were known to be in the range of 250–300 \( \mu \)m. This result is entirely consistent with lubrication theory. However, Farrar and Schmidt [35] reported an increase in wear when clearances dropped to 30 \( \mu \)m. This was thought to be caused by geometrical errors in manufacturing rather than by a breakdown in lubrication theory. They also reported seizure of components which had negative clearances. A factor that does need to be taken into consideration when designing clearances is that of cup size. A factor that does need to be taken into consideration when designing clearances is that of cup size. A factor that does need to be taken into consideration when designing clearances is that of cup size.

The effect of diameter was examined by Dowson et al. [22] and their results are summarized in Fig. 5. At small diameters (16 and 22 mm), wear increased with increasing bearing surface diameter. This indicated that these components were operating in the boundary lubrication regime in which wear increases with increasing sliding distance. At larger diameters (28, 36, and 54.5 mm) the wear rate decreased with increasing diameter, clearly indicating that these components were operating in the mixed and fluid-film lubrication regimes. It also shows that this...
Fig. 4  Bar charts showing the effect of clearance on running-in wear \((2 \times 10^6\) cycles) with (a) 36 mm and (b) 54 mm diameter heads. (After Dowson et al. [22])

Fig. 5  Graph showing the effect of diameter on wear: curve a, 16 mm, \(n = 5\), and \(C_d \approx 53–70\ \mu m\); curve b, 22.225 mm, \(n = 5\), and \(C_d \approx 46–66\ \mu m\); curve c, 28 mm, \(n = 4\), and \(C_d \approx 55–70\ \mu m\); curve d, 36 mm, \(n = 3\), and \(C_d \approx 76–78\ \mu m\); curve e, 54.5 mm, \(n = 4\), and \(C_d \approx 83–129\ \mu m\). (After Dowson et al. [22])
improvement in performance continued up to diameters which are significantly greater than those which are generally used for total hip replacement. However, such diameters are commonplace with surface replacement implants, a design philosophy which is being increasingly used for younger patients. It is encouraging that such components not only have the clinical advantages of conserving bone, a large range of motion, and reduced likelihood of dislocation [37] but also, if correctly engineered, have the potential for the lowest wear rates. An example of a surface replacement device is shown in Fig. 6.

Wear in metal-on-metal components is known to consist of two distinct phases. A relatively high-wear ‘running-in’ phase which lasts for \((0.5–2) \times 10^6\) cycles, followed by a ‘steady state’ phase when the wear rate is relatively constant and much lower [10, 28, 38]. Figure 7, derived from data reported earlier [19], demonstrates that the effects of optimizing wear characteristics in metal-on-metal components are most marked in the bedding-in phase. Figure 7(a) shows that wear rates are reduced when bearing surface diameters are increased both in the running-in phase \([(0–2) \times 10^6\] cycles\) and in the steady state phase \([(2–5) \times 10^6\] cycles\). Figure 7(b) shows that reduced clearances have the most marked effect in the running-in phase. This was attributed by Hu et al. [39] to greater penetration and hence to greater volumetric wear necessary to develop a given contact area with components which have a larger clearance. A comprehensive review of steady state wear reported in various hip simulators was undertaken by Dowson [40] which showed that theoretical film thickness alone, independent of small variations in surface roughness, was a good predictor of the long-term performance of metal-on-metal bearings. A film thickness of less than 12 nm produced relatively high wear rates. A dramatic drop in wear rates occurred in the range 12–20 nm, with increasing film thickness having little further effect on wear rates.

The effects of surface roughness are well known and have been reported by Chan et al. [30] who concluded that, in line with lubrication theory, increased surface roughness was associated with higher wear rates. However, recently researchers have tended to ignore the effects of surface roughness because of the self-polishing capacity of metal-on-metal components [10]. There is evidence to suggest that, below certain limits, roughness does not affect wear rates [40], and the limitations placed on surface finish are due to the nature of the metal itself.

Metal-on-metal components have given good clinical results but there are still concerns over long-term exposure to elevated levels of metal ions. Hip simulator studies have demonstrated that significant reductions in wear and thus metal ion levels can be
made if designs are optimized by utilizing larger-diameter components with reduced clearances between the femoral head and acetabular cup but clinical studies measuring metal ion levels are still required to validate these laboratory studies.

It can be seen from the above discussion that the metal-on-metal combination provides an attractive option as a bearing surface. However, continued concerns regarding their use warrants a review of other solutions for low-wear bearing couples.

4 CURRENT ALTERNATIVE SOLUTIONS

4.1 Polyethylene-on-metal and ceramic femoral heads

Polyethylene-on-metal or ceramic femoral heads have been the most widely used bearing combination over the last 40 years [2, 23–25]. Higher demand and more active patients with longer life expectancies have led to increased levels of failure due to polyethylene wear debris-induced osteolysis [41]. Acceleration of wear and wear debris-induced osteolysis in conventional polyethylene bearings gamma irradiated in air have been associated with oxidative ageing of the polyethylene and roughening and damage to the femoral heads. A greater than twofold increase in wear rate was found with oxidative ageing following a shelf life of 5 years [42], while a longer shelf life was found to cause further increase in wear [43]. Oxidation was also found to produce smaller and more reactive wear particles [43, 44]. Damage to femoral heads [5] was found to work synergistically with oxidation to increase wear further [43]. Tipper et al. [45] quantified a series of retrieved Charnley hip prostheses with a lifetime of 10–20 years and found a doubling of the polyethylene wear rate from 40 to 80 mm³ per year cycles for prostheses with damaged femoral heads. This was supported by simulator tests on polyethylene gamma sterilized in air which showed a threefold increase in wear with scratched metal heads [46]. In contrast, alumina ceramic femoral heads have been found to be resistant to scratching and damage in vitro and in vivo [47, 48].

In the last 10 years, there has been a focus on improving both the polyethylene material, by making it oxidation resistant and through intentional cross-linking, and also by more widespread use of ceramic femoral heads. Polyethylene sterilized in an inert atmosphere and stored in an oxygen-free environment has been shown to be more resistant to oxidation and wear. One example of such a material gamma irradiated with 4 MRad in a vacuum and foil packed, GVF GUR1020 polyethylene, has been shown to have a wear rate of 35 mm³ per 10⁶ cycles against metallic femoral heads and 25 mm³ per 10⁶ cycles against alumina ceramic femoral heads in a hip joint simulator [49, 50] compared with 50 mm³ per 10⁶ cycles for material gamma irradiated and stored in air. Intentionally cross-linked polyethylene (based on a higher-molecular-weight resin GUR1050) was reported by McKellop et al. [51] to have produced an 80 per cent reduction in wear with a medium level of cross-linking 5MRad. Over 95 per cent reduction in wear, essentially zero wear, in highly cross-linked polyethylene (10 MRad) was reported by Muratoglu et al. [52]. The virtually zero wear and the retention of surface machining marks found in these early simulator studies with highly cross-linked polyethylene [52] has not been found to represent clinical performance where finite wear rates and loss of surface machining marks have been found in retrievals by Bradford et al. [53]. A more recent hip simulator study of highly cross-linked polyethylene, with lower, physiologically relevant, serum protein levels, and higher loads consistent with the ISO standards reported on the effects of various levels of radiation [54]. Wear decreased with increased exposure to radiation but even at the 10MRad level exhibited finite wear rates up to 9 mm³ per 10⁶ cycles, loss of machining marks and linear penetrations consistent with clinical retrievals in bearings of size 28 mm, as shown in Fig. 8. These clinical and realistic laboratory wear rates should be compared with the value of 1.6 mm³ per 10⁶ cycles reported for the same-diameter components in metal-on-metal bearings [22].

However, the biological response to wear debris is quite different for polyethylene and metal. Polyethylene debris produces an inflammatory response [41], and indeed it has been shown recently that higher-molecular-weight and cross-linked polyeth-ylene wear debris is smaller and more reactive than the conventional lower-molecular-weight GUR1020 polyethylene debris [55]. In contrast, metal wear particles produce a very low level of inflammatory response [56]. However, concerns remain about cyto-toxicity of metal particles if they accumulate in high concentrations [57] and the potential for a hypersensitivity reaction in a small number of patients [19]. It is important to note that the small nanometre-sized metallic particles are readily transported away from the prosthesis and disseminated throughout the body.

As there is a move towards larger femoral heads, 36 mm or greater, there is concern that this will further increase wear in polyethylene bearings, in
contrast with an improved lubrication and reduction in wear in metal-on-metal bearings. A further concern is that the reduced polyethylene thickness associated with larger-diameter components may lead to a weakened overall construct.

### 4.2 Ceramic-on-ceramic bearings

Alumina ceramic-on-ceramic bearings have an extensive clinical history. Both the Mittelmeier Autophor prosthesis specifically (Ceramtec AG, Plochingen, Germany) and other ceramic-on-ceramic bearings produced by the same manufacturer have over 20 years' clinical experience. Although during that period there have been substantial changes in design and materials, and issues with fixation \([11]\), the overall clinical experience with the bearing surface has been good \([58, 59]\).

Laboratory hip simulator studies which have been carried out by Nevelos and co-workers \([60–64]\) under standard conditions have shown extremely low wear rates for alumina ceramic-on-ceramic bearings, as low as 0.01–0.1 mm\(^3\) per 10\(^6\) cycles. These wear rates can be tenfold lower than metal-on-metal hips, as shown by Fig. 9, and over a hundredfold lower than polyethylene. However, these standard simulator tests did not replicate the wear rate, wear mechanisms, and patterns of wear debris found clinically or on retrievals. Explant studies on two different types
of ceramic-on-ceramic hip have shown higher wear rates, of the order of 0.5–1 mm$^3$ per year [62], characteristic stripe wear scars on the head [62, 63], and a bimodal debris distribution with both nanometre- and micrometre-size wear particles [64].

A new simulator methodology has been introduced which included microseparation of the head and cup, and this resulted in contact of the head on the rim of the cup at heel strike and stripe wear formation on the head [62]. Microseparation simulator testing produced similar wear rates, wear mechanisms, and wear debris to that found on retrieved prostheses [62, 65, 66]. The alumina ceramic wear debris has been found to be highly biocompatible [67], producing a lower level of osteolytic cytokines than cross-linked polyethylene, and being less cytotoxic than metallic debris [57]. This makes alumina-on-alumina ceramic bearing couples an attractive option with sizes in the range 28–36 mm. Larger-diameter components, which in theory should further reduce wear, are not used clinically because of limitations dictated by the commensurate reduction in cup thickness.

However, concerns remain about brittle fracture of ceramic components and chipping of acetabular inserts during implantation. This, together with general design constraints, limits their current use in many countries. More recently a new alumina matrix composite, namely zirconia-toughened alumina, has been developed. The increased toughness gives lower fracture risk, more design flexibility, and most importantly reduction in wear under microseparation conditions [68].

5 FUTURE ALTERNATIVE SOLUTIONS

Despite the significant improvements offered by the technologies discussed above, there nevertheless remain concerns related to polyethylene wear (cross-linked polyethylenes), metallic debris, and subsequent potential metal ion release (metal-on-metal), and ceramic liner fracture (ceramic-on-ceramic). Therefore, in addition to these current alternative solutions, it would seem appropriate to review some potential future alternatives to metal-on-metal bearings.

5.1 Ceramic-on-metal bearings

The brittle nature of ceramics is such that liners can be subject to occasional fractures, despite the best efforts of the component designers to minimize the risks. Typically, ceramic liners may fracture either in surgery, through incorrect insertion into the metal acetabular shell, or alternatively post-surgery through a high-impact trauma which overloads the component. A recent review of this issue in ceramic bearings by Hannouche et al. [69] reported on a large series of total hip arthroplasties with a liner fracture rate of five out of 3300 implanted. They concluded that 'Ceramic fracture, although of great concern, has to be considered with the perspective of the gain obtained in using a highly performing material regarding wear'.

Currently in orthopaedics, softer polyethylenes are always used with a dissimilar harder, metal, or ceramic femoral head. This is consistent with common engineering practice which tends to avoid wherever possible the use of like materials as bearing surface combinations and favours the use of dissimilar materials with a differential hardness. However, hard-on-hard bearings such as metal-on-metal or ceramic-on-ceramic are used as like-on-like bearings. Ceramic-on-metal articulation offers a combination with an order-of-magnitude hardness differential, while the use of materials that can be finished to very tight manufacturing tolerances also maintains fluid-film lubrication during articulation.

Firkins et al. [70] investigated the wear properties and debris morphology of a novel differential hardness prosthesis combining an alumina ceramic femoral head and high-carbon cobalt–chromium–molybdenum alloy acetabular liner, in comparison with metal-on-metal articulations in a physiological anatomical hip joint simulator. The results revealed that the ceramic-on-metal pairings were found to have wear rates approximately a hundredfold lower than the metal-on-metal pairings. The wear rate of the ceramic-on-metal pairings, as shown in Fig. 9, was found to be approximately 0.01 mm$^3$ per $10^6$ cycles for the duration of the test with very little wear detected on the surface of the components. The initial bedding-in period which is characteristic of conventional metal-on-metal components was absent in the ceramic-on-metal tests. The higher wear rate of metal-on-metal articulation continued into the steady state wear rate of $1.23 \pm 0.5$ mm$^3$ per $10^6$ cycles.

The size and shape of the particles from both articulations were similar, metallic, round to oval in shape, and in the nanometre size range, with a diameter of $30 \pm 2.25$ nm for metal-on-metal bearings and $17.57 \pm 1.37$ nm for ceramic-on-metal bearings after $10^6$ cycles.

The same factors that optimize hard-on-hard bearings in general apply to a ceramic-on-metal coupling. Therefore fluid-film thickness should be maximized.
and design issues such as diametrical clearance, sphericity, and surface roughness remain important to achieving successful articulation. It should further be emphasized that the testing and analysis reported here employed the combination of a ceramic femoral head and an acetabular metal liner. The use of the combination in reverse (metallic head and ceramic liner) has not been tested. The known historical failures of polyethylene heads articulating against metal liners, as well as the potential for scratching of the head if rim contact occurred during articulation make this unlikely to be a viable bearing combination.

In summary, ceramic-on-metal bearings demonstrated the advantages of exceptionally low wear, reducing the osteolytic potential and ion release associated with polymeric and metallic wear debris respectively. Furthermore, it offers reduced potential for acetabular component liner fracture, relative to ceramic-on-ceramic articulation, and reduced damage to femoral heads under the microseparation and potential rim contact conditions reported by Lombardi et al. [71].

5.2 Surface engineered coatings on metal-on-metal hip replacements

Surface engineered coatings have been investigated by Fisher et al. [72, 73] to examine the potential in reducing the volume of wear, the concentration of metal debris, and the levels of cobalt, chromium, and molybdenum ions released.

Thick (8–12 μm) surface engineered coatings [chromium nitride (CrN) and chromium carbonitride (CrCN)] were deposited by arc evaporative physical vapour deposition on cobalt–chromium–molybdenum heads and cups and tested in a hip simulator. CrN heads were articulated with CrN inserts and CrN heads with CrCN inserts in a $5 \times 10^6$ cycle hip joint simulator study (Leeds Mark II). As illustrated by Fig. 10, the overall wear of the CrN-on-CrN and CrCN-on-CrCN bearing couples was at least 22-fold lower, with overall wear being less than 0.05 mm$^3$ per $10^6$ cycles compared with 1.2 mm$^3$ per $10^6$ cycles for metal-on-metal couples [72, 73].

Wear particles produced in all hip simulator tests were characterized by transmission electron microscopy using a method described by Tipper et al. [74]. All material combinations produced particles generally less than 30 nm in size. The metal ion levels in the serum lubricant were measured; cobalt and chromium ion levels were determined using graphite furnace atomic absorption spectroscopy, and molybdenum ion levels were determined by inductively coupled plasma mass spectrometry. Compared with metal-on-metal uncoated bearings; the cobalt, chromium, and molybdenum ion release was less in all serum samples from the coated bearing couples. The cytotoxicity of the metallic, CrN, and CrCN wear particles was assessed by co-culture with U937 macrophages and L929 fibroblast cells by measuring the effect on cell viability [75]. A significant reduction in the viability of macrophage cells was observed with metallic wear particles at concentrations of 50, 5, and 0.5 μm$^3$ of wear debris per cell compared with the cell only control. The fibroblast cells only showed a significant reduction in viability with 50 μm$^3$ of wear debris per cell. The CrN wear debris at concentrations at 50 and 5 μm$^3$ of wear debris per cell caused a significant reduction in macrophage cell viability, whereas only the highest concentration (50 μm$^3$ of wear debris per cell) caused a significant reduction in the viability of fibroblast cells. The CrCN wear particles were least cytotoxic compared with the other two coatings.

![Fig. 10](image_url) Wear rates of CrN- and CrCN-coated hip replacements compared with metal-on-metal bearings throughout testing. (After Fisher et al. [72, 73])
wear debris caused no significant reduction in cell viability at any concentration tested with either cell type. These initial findings support further development and clinical trials of surface engineered metal-on-metal bearings. In particular, surface engineered coatings may offer an alternative to metal-on-metal surface replacement designs because of the reduced wear and ion release compared with metal-on-metal bearings and increased design flexibility compared with ceramics.

6 CONCLUSIONS

1. Improvements in polyethylene and the introduction of ceramic femoral heads have reduced wear rates but not eliminated the risk of osteolysis.

2. Clinical evidence has indicated that hard-on-hard bearing surfaces are a viable alternative to polyethylene.

3. Laboratory and theoretical evidence indicates that metal-on-metal bearings can produce very low wear volumes when optimized by using components which have large diameters, smooth surfaces, and small diametral clearance between femoral and acetabular components.

4. Ceramic-on-ceramic bearings produce significantly reduced amounts of less reactive wear debris than metal-on-metal components but their use is limited by concerns of component fracture and consequent limitations of applicability in large-diameter thin designs of hip replacement.

5. New technologies such as ceramic-on-metal and surface engineered coatings have been tested and may produce further improvements in terms of performance and applicability of hard-on-hard bearings.

Encouraging short- to medium-term clinical results with second-generation components, and increasing confidence in the quality of preclinical testing in hip joint simulators have led to a significant increase in the use of hard-on-hard bearings. The in vivo performance of bearings using polyethylene can be measured using X-rays. Similar measurements cannot be made with hard-on-hard bearings because the materials are radio-opaque and the volume of wear, as measured by change of dimension, is too small. However, wear can be deduced indirectly for metal-on-metal bearings by measuring the level of metal ions in whole blood, serum, or urine. Only when these studies are complete will it be known whether the latest generation of components are effective in reducing the volume of wear debris, thereby validating the test methods, the data generated, and the designs which resulted.

REFERENCES


Metal-on-metal bearings surfaces  


