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Bearing Surface Options for Total Hip Replacement in Young Patients

BY CHRISTIAN HEISEL, MD, MAURICIO SILVA, MD, AND THOMAS P. SCHMALZRIED, MD

An Instructional Course Lecture, American Academy of Orthopaedic Surgeons

Total hip arthroplasty is one of the most successful and cost-effective surgical interventions in medicine[1] and is the most effective treatment for osteoarthritis of the hip joint. Long-term studies of selected patient cohorts[2-4] and the Scandinavian hip registries[5,6] have demonstrated high survivorship rates after more than twenty years. On the basis of this success, total hip replacement is being performed on increasingly younger and more active patients. However, there are at least two problems that a young or active patient faces with regard to the prosthetic joint. First, the use of the implant is more intense in proportion to their physical activities[7]. Second, the patient's life expectancy is longer and the potential total number of loading cycles is increased proportionally.

Patient-related factors contribute to implant wear regardless of the type of bearing[8]. Higher patient activity results in higher wear rates[8]. Follow-up studies of young patients have demonstrated a relationship between the amount of wear and the age of the patient[9,10], the revision rate[11], osteolysis[9,12,13], and aseptic loosening[12]. The overall rate of survival of total hip arthroplasty implants in young patients is reduced compared with that in average patient groups[3]. The survival rate of artificial joints in patients younger than fifty years of age is approximately 80% after ten years or more, regardless of the fixation technique and bearing combination[11,12,14-18]. To our knowledge, only one recent study, by Kim et al.[10], demonstrated a survival rate of 99% after ten years in patients less than fifty years of age.

In chronological order, the categorical factors limiting the function and longevity of a total hip prosthesis are the surgical technique, fixation of the implant to the bone, osteolysis (often associated with wear of the bearing), fatigue failure of the implants, and long-term skeletal remodeling. No implant system can overcome inadequate surgical technique. A sound biomechanical construct is the foundation of a well-functioning prosthesis. The next challenge is to obtain and maintain satisfactory fixation. As physical activity increases, the stresses on the fixation interfaces and the implants also increase. The durability of implant fixation in young patients has been improved by cementless fixation[19]. Osteolysis associated with polyethylene wear has become the limiting factor[20,21] (Fig. 1). Except in cases where the bearing actually wears through, wear is clinically important only if it induces progressive osteolysis. Hips are generally not revised as a result of wear; they are revised because of osteolysis associated with wear (and the generation of wear particles).

Polyethylene, ceramic, and metal wear particles incite an inflammatory response that can result in periprosthetic bone resorption (osteolysis)[22-26]. Bearing wear, osteolysis, and aseptic loosening can limit the durability of a prosthetic hip joint, irrespective of the combination of materials used[27,28], given

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that the biomechanical reconstruction and fixation are satisfactory.

New bearings for total hip arthroplasty have been introduced with the aim of reducing the number of biologically active wear particles. There are two approaches: one is to improve the wear resistance of polyethylene through cross-linking[29] and the other is to avoid polyethylene and utilize alternative bearings. The latter approach has fueled the development and reintroduction of new ceramic-on-ceramic and metal-on-metal bearings. The goal of all bearing combinations is to reduce wear to less than a clinically relevant level—that is, a level that does not induce osteolysis or another outcome that necessitates revision surgery. Although the use of surrogate variables such as wear rates to predict the outcome of a total joint replacement may be helpful as a prognostic tool, this should be done with caution.

Tribology—Wear and Lubrication

Tribology is defined as the science of surfaces interacting under an applied load and in relative motion (as in bearings or gears). It includes the study of friction, lubrication, and wear. Wear is the removal of material, with the generation of wear particles, that results from relative motion between two apposed surfaces under load. The primary wear mechanisms are adhesion, abrasion, and fatigue. Adhesion involves bonding of the surfaces when they are pressed together under load. Sufficient relative motion results in material being pulled away from one or more surfaces, usually from the weaker material. Abrasion is a mechanical process in which asperities on the harder surface cut and plough through the softer surface, resulting in removal of material. When local stresses exceed the fatigue strength of a material, that material fails after a certain number of loading cycles, with release of material from the surface[28].

The conditions under which the prosthesis was functioning when the wear occurred have been termed the wear modes[30]. Mode-1 wear results from the motion of two primary bearing surfaces against each other, as intended. Mode 2 refers to the condition of a primary bearing surface moving against a secondary surface that was not intended to come into contact with the first. Usually, this mode of wear occurs after excessive wear in Mode 1. An example would be when a femoral component penetrates through a modular polyethylene liner and articulates with the metal backing. Mode 3 refers to the condition of the primary surfaces moving against each other, but with third-body particles interposed. In Mode 3, the contaminant particles directly abrade one or both of the primary bearing surfaces. This is known as three-body abrasion or three-body wear. The primary bearing surfaces may be transiently or permanently roughened by this interaction, leading to a higher Mode-1 wear rate. Mode-4 wear refers to two secondary (nonprimary) surfaces rubbing together. Examples of Mode-4 wear include wear due to relative motion of the outer surface of a modular polyethylene component against the metal support, so-called backside wear, fretting between a metallic substrate and a fixation screw, or fretting and corrosion of modular taper connections and extra-articular sources. Particles produced by Mode-4 wear can migrate to the primary bearing surfaces, inducing three-body wear (Mode 3)[30].

Lubrication has a major influence on the amount of abrasive and especially adhesive wear. The tribological performance of a joint depends on the fluid film covering its surfaces[31]. A high ratio of fluid-film thickness to surface roughness (λ ratio) is desirable in order to reduce friction and wear. A λ ratio equal to or less than unity describes boundary lubrication. With an increasing λ ratio, friction is reduced and the bearing reaches a state of mixed lubrication. A value of >3 represents fluid-film lubrication[32,33]. Fluid-film lubrication completely separates the surfaces of a bearing. This occurs when the lubricating film is thicker than the height of the asperities on the bearing surfaces. In this situation, the load is carried by the fluid, and wear of the bearing is minimal. Mixed film lubrication separates the surfaces only partially and is represented by a λ ratio of >1 and < 3. For a given load and sliding velocity, fluid-film thickness is dependent on the properties of the fluid, the bearing materials, the macrogeometry of the bearing (which is a function of diameter and radial clearance), and the surface microtopography (surface finish)[34].

Cross-Linked Polyethylene Acetabular Bearings

Ultra-high molecular weight polyethylene has been the preferred acetabular bearing material for more than thirty years. The aggregate clinical experience

TABLE I Alternate Bearing Combinations for Total Hip Arthroplasty

Bearing Material	Benefits	Risks
Cross-linked polyethylene	High wear resistance, no toxicity, relatively low cost, multiple liner options (elevated rim, etc.)	Reduction in other material properties (gross material failure), increased bioactivity of wear particles
Metal-on-metal	Very high wear resistance, favors larger diameters (lowers wear), long in vivo experience	Increased ion levels, delayed-type hypersensitivity, carcinogenesis
Ceramic-on-ceramic	Highest wear resistance, no toxicity, long in vivo experience	Position sensitivity, liner chipping, fracture risk

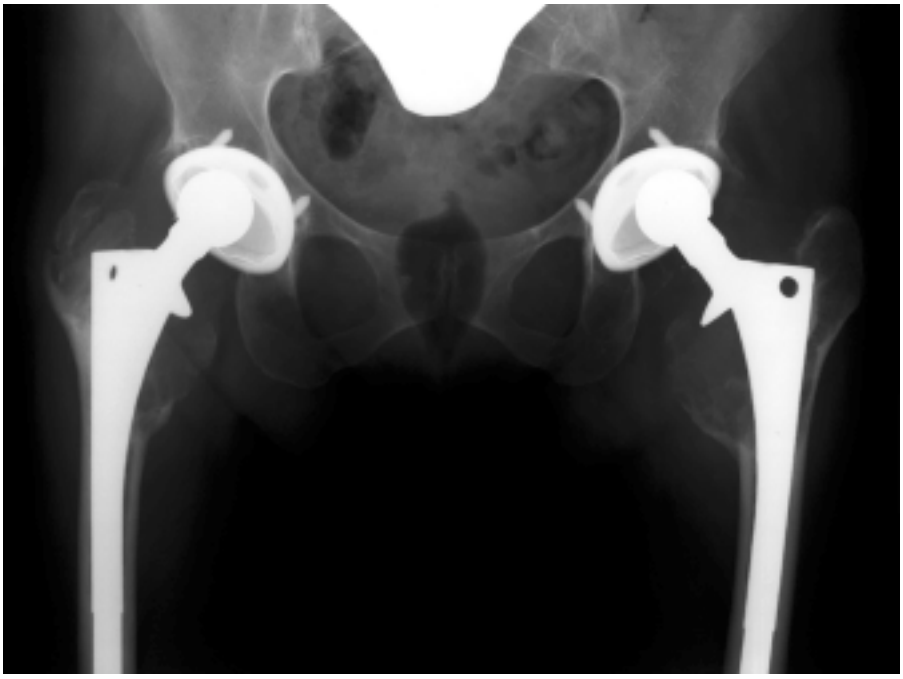

Fig. 1
Radiograph made eight years after total hip replacement in a fifty-year-old woman. A high degree of polyethylene wear and associated osteolysis are seen.

indicates a low probability of gross material failure of this application and, despite evidence of systemic distribution, there are no clinically apparent systemic consequences. The fundamental limitation is wear resistance.

Ethylene is a gaseous hydrocarbon composed of two carbon atoms and four hydrogen atoms: C_2H_4. Polyethylene is a long-chain polymer of ethylene molecules in which all of the carbon atoms are linked, each of them holding its two hydrogen atoms[35]. The mechanical properties of ultra-high molecular weight polyethylene are strongly related to its chemical structure, molecular weight, crystalline organization, and thermal history[36].

The microstructure of ultra-high molecular weight polyethylene is a two-phase viscoplastic solid consisting of crystalline domains embedded within an amorphous matrix[36,37]. Connecting the crystalline domains are bridging tie molecules that provide improved stress transfer and physical strength[37]. Ultra-high molecular weight polyethylene is defined as polyethylene with an average molecular weight of greater than 3 million g/mol[36]. The ultra-high molecular weight polyethylene currently used in orthopaedic applications[36,38] has a molecular weight of 3 to 6 million g/mol, a melting point of 125°C to 145°C, and a density of 0.930 to 0.945 g/cm^3. Ticona (Summit, New Jersey) and Basell Polyolefins (Wilmington, Delaware) supply ultra-high molecular weight polyethylene resins to orthopaedic manufacturers. Calcium stearate is an additive in the manufacturing process of many polyethylene resins; it acts as a corrosion inhibitor[36,39], whitening agent[38], and lubricant to facilitate the extrusion process[36,39,40]. In general, both Ticona and Basell resin powders consist of numerous fused, spheroidal ultra-high molecular weight polyethylene particles, but a fine network of submicrometer-sized fibrils that interconnect the microscopic spheroids characterizes Ticona resins. Ticona resins have a mean particle size of approximately 140 µm, whereas Basell resins have a mean particle size of approximately 300 µm[36].

Cross-linking has been utilized to improve the wear resistance of polyethylene and can be accomplished with use of peroxide chemistry, variable-dose ionizing radiation, or electron beam irradiation[41]. Cross-linking occurs when free radicals, located on the amorphous regions of polyethylene molecules, react to form a covalent bond between adjacent polyethylene molecules. It is believed that cross-linking of the polyethylene molecules resists intermolecular mobility, making the polyethylene more resistant to deformation and wear in the plane perpendicular to the primary molecular axis. This has been demonstrated to dramatically reduce wear from crossing-path motion, as occurs in acetabular components[42,43]. Cross-linking has a detrimental effect on yield strength, ultimate tensile strength, and elongation to break[29]. The decrease in these properties is proportional to the degree of cross-linking. This fact has generated debates on the optimal degree of cross-linking. Hip-simulator studies have indicated that cross-linking can reduce the type of wear that occurs in acetabular components by >95%[29,44,45] (Fig. 2).

Clinical and laboratory research has revealed that sterilization methods can dramatically affect the in vivo performance of a polyethylene component[46] (Fig. 3). Polyethylene components can be sterilized with gamma irradiation, gas plasma, or ethylene oxide. Gamma irradiation in an air environment was the industry standard since the early 1970s; the doses range between 2.5 and 4 Mrad (1 Mrad = 10^6 radiation absorbed dose = 10^4 Gy) and are most commonly between 3.0 and 3.5 Mrad (30,000 to 35,000 Gy). Gamma radiation breaks covalent bonds, including those in the polyethylene molecules. This produces unpaired electrons from the broken covalent bonds, called free radicals. These highly reactive moieties can combine with oxygen (if present) during the irradiation process, during shelf-storage, and in vivo.

Oxidation of the polyethylene molecule is a chemical reaction that results in chain scission (fragmentation and shortening of the large polymer chains) and introduction of oxygen into the polymer[38]. The net result lowers the molecular weight of the polymer; reduces its yield strength, ultimate tensile strength, elongation to break (makes it

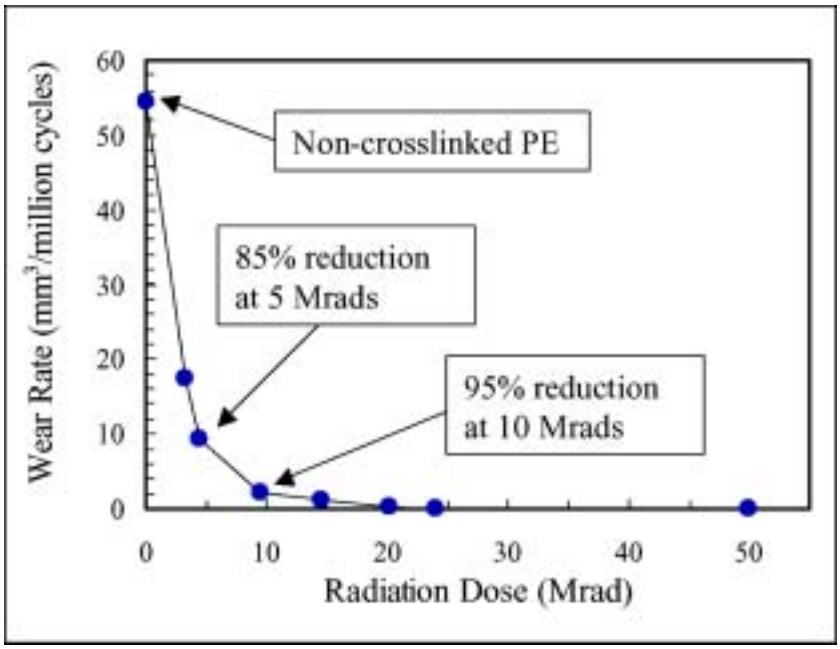

Fig. 2
Wear rate of polyethylene (PE) acetabular components in a hip simulator as a function of radiation dose[29] (1 Mrad = 10,000 Gy).

more brittle), and toughness; and increases its density (lowers its volume)[47-50].

In general, oxidation and cross-linking are competing reactions. As cross-linking increases, oxidation decreases and vice versa[51,52]. In components subjected to gamma irradiation in air, the relative amount of oxidation and cross-linking varies with the depth from the surface of the component[52]. This results in a corresponding variation in the wear resistance of the material as a function of the depth from the surface[51]. Once implanted, the component is exposed to dissolved oxygen in body fluids. Free radicals in the polyethylene will react with the available oxygen over time. Relatively little is known about the rate of oxidation of polyethylene in vivo. It appears that it is lower than that in vitro, but there is debate about how much lower and it is likely that several factors affect the rate of oxidation.

Methods have been developed to produce components with increased wear resistance due to cross-linking that do not oxidize on the shelf or in the body. Free radicals created in polyethylene by ionizing radiation can be driven to a cross-linking reaction by heating the polymer to above the melting temperature (125°C to 135°C)[29]. Components made from such remelted material have no residual free radicals; thus, there is no potential for oxidation when the component is subsequently sterilized by ethylene oxide or gas plasma. Remelting does, however, induce changes in the crystalline structure of the material that are associated with a decrease in some material properties. This fact has resulted in controversy regarding the relative detriment of remelting compared with that of retention of some residual free radicals.

The manufacturing processes of the currently available products—Marathon (DePuy, Warsaw, Indiana), Longevity (Zimmer, Warsaw, Indiana), Durasul (Centerpulse Orthopedics, Austin, Texas), Crossfire (Stryker Howmedica Osteonics, Allendale, New Jersey), and XLPE (Smith and Nephew Orthopaedics, Memphis, Tennessee)—differ with regard to the dose and type of irradiation (gamma or electron beam), thermal stabilization (remelting or annealing), machining, and final sterilization[53]. For this reason, each material should be considered separately, and the specific wear characteristics of each

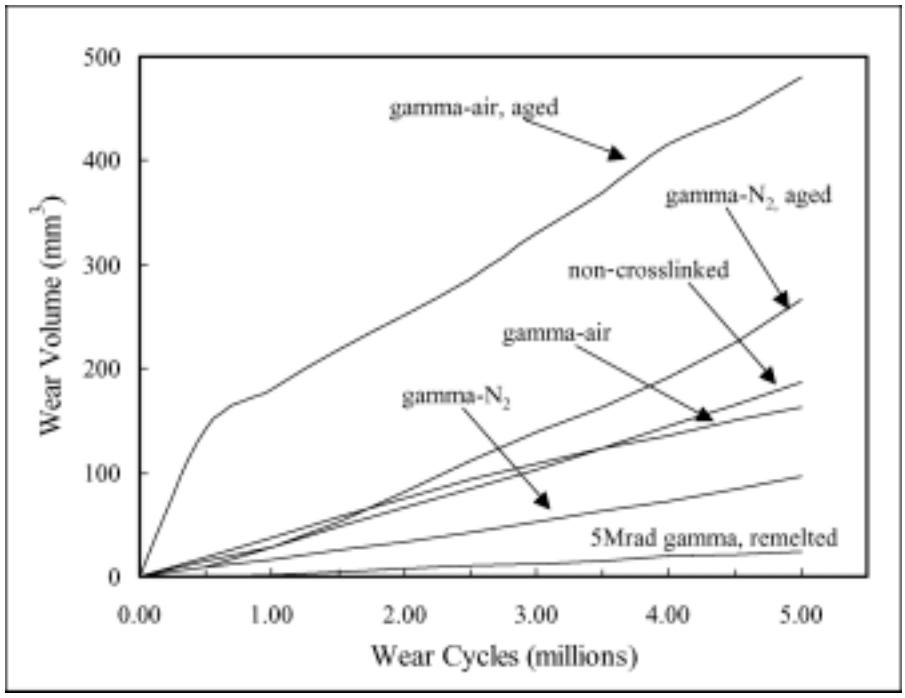

Fig. 3
Wear of polyethylene acetabular components in a hip simulator as a function of manufacturing and sterilization method. Wear rate increases with higher levels of oxidation[46]. Gamma = gamma irradiation, and 1 Mrad = 10,000 Gy.

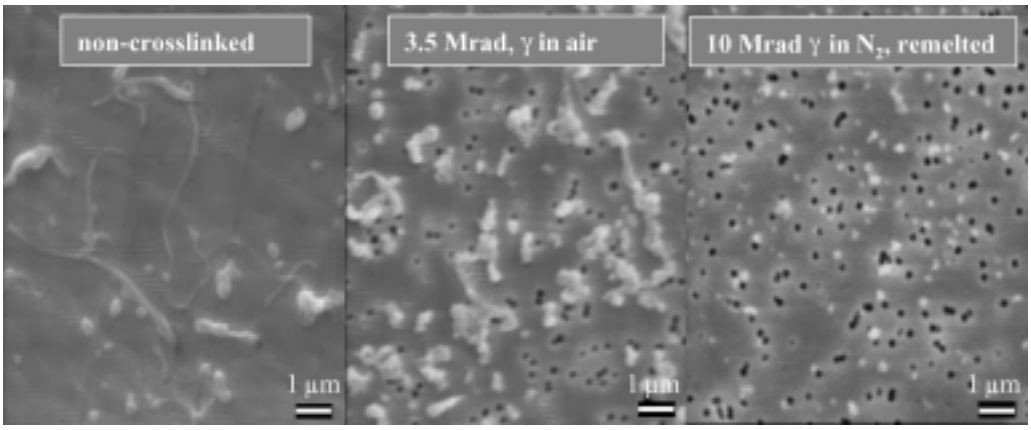

Fig. 4

Shape and size of polyethylene particles isolated from hip-simulator studies. Non-cross-linked polyethylene produces a large amount of particles that are elongated fibrils. A substantial reduction in particle size is seen between the non-cross-linked polyethylene (left) and the partially cross-linked polyethylene (3.5 Mrad [35,000 Gy]) (center). Partially cross-linked polyethylene produces mostly round particles of submicrometer size and fibrils of up to a few micrometers in size. Highly cross-linked polyethylene (10 Mrad [100,000 Gy]) (right) shows predominantly round particles in the submicrometer size range (×10,000).

should be established through clinical studies.

A reoperation for any reason is the primary definition of failure of a total hip arthroplasty. Unfortunately, often a long follow-up period is required to demonstrate statistical and practical differences between implant systems. Shorter-term in vivo wear studies may help to predict long-term outcomes. Increased volumetric wear has been associated with component loosening and osteolysis[54-58]. The association between volumetric wear and periprosthetic bone resorption appears to be related to the number and size of polyethylene wear particles that are generated and are released into the effective joint space[59]. On this basis, a lower wear rate may not necessarily be clinically preferred if a higher number of biologically active wear particles are generated.

Penetration of the femoral head into the acetabular polyethylene is due to a combination of creep and wear. Because of creep, short-term linear penetration rates tend to be higher than those seen over the longer term. Because creep decreases exponentially with time, it is generally accepted that the majority of the linear penetration that occurs after the first one or two years is due to wear[60,61].

Data from clinical trials with small patient groups have shown a reduction in wear rate associated with cross-linking[60,62-64]. Martell et al.[65] evaluated seventy-four patients at a minimum of two years postoperatively. Thirty-five patients had a polyethylene acetabular liner that had been gamma irradiated in air (the historical standard), and thirty-nine had a liner that had been cross-linked with 3 Mrad (30,000 Gy) of gamma radiation in nitrogen and then heat annealed. Radiographs were analyzed with a computer-assisted, two-dimensional digital edge detection technique[66]. The hips treated with the historical standard had a mean volumetric wear rate of 94 ± 78 mm^3/yr compared with a mean of 54 ± 70 mm^3/yr in the hips with the cross-linked polyethylene ($p < 0.05$). The degree of clinical wear reduction associated with the intentionally cross-linked material was very similar to the degree of wear reduction seen in a hip-wear-simulator study comparing these same acetabular components[65]. In another clinical study, Martell and Incavo[67] compared twenty-four liners made of highly cross-linked polyethylene (Crossfire; gamma irradiation to 7.5 Mrad [75,000 Gy], heat-annealed at 120°C, and sterilized with 2.5 to 3.5 Mrad [25,000 to 35,000 Gy] of gamma radiation while packaged in nitrogen) with twenty-five standard polyethylene liners that had been sterilized in the same manner. After two years of follow-up, the cross-linked polyethylene showed a significant (53%) reduction in linear wear (0.094 compared with 0.202 mm/yr; $p = 0.008$).

Digas et al.[68] initiated a prospective study comparing a cemented, highly cross-linked polyethylene acetabular component (Durasul; electron beam radiation to 9.5 Mrad [95,000 Gy] at 125°C, remelted at 150°C for two hours, and sterilized with ethylene oxide) with a cemented polyethylene component that had been sterilized with gamma irradiation in nitrogen. The in vivo wear in thirty-three patients followed for a minimum of two years was measured with radiostereometric analysis. The fifteen patients with the cross-linked polyethylene had less three-dimensional femoral head penetration (0.18 compared with 0.20 mm/yr) but the difference was not significant. The fact that there was not more of a difference in the linear penetration in this short-term study is not surprising given that the creep rates of the two polymers are about the same[68]. Another conclusion that could be reached on the basis of those data is that the clinical performance of moderately cross-linked polyethylene with little oxidation (the "standard" polyethylene) is quite good.

In another study[69], twenty-four hips received a modular polyethylene liner sterilized with gamma radiation in air (Enduron) in combination with a cementless acetabular component, and thirty hips received the same acetabular component with a cross-linked polyethylene liner (Marathon; gamma irradiation to 5 Mrad [50,000 Gy], remelted at 155°C for twenty-four hours, machined from the center of a ram-extruded bar, and sterilized with gas plasma). The patients treated with the cross-linked polyethylene were younger and more active than the patients treated with the conventional polyethylene (mean ages, fifty-nine and seventy-four years, $p < 0.0001$). After a minimum of two years of follow-up, the conventional polyethylene liners had a mean volumetric wear rate of 88 ± 79 mm^3/yr, with 104 mm^3/yr in the men and 74 mm^3/yr in the women. The cross-linked liners had a mean volumetric wear rate of 21 ± 23 mm^3/yr, with 28 mm^3/yr in the men and 15 mm^3/yr in the women. The difference between the conventional and cross-linked liners was significant ($p = 0.0001$). Some of the reduction in linear penetration observed in this study may have been due to a reduction in conformational change (backside bedding-in) between the modular acetabular liner and the metal shell. Adjusted for the measured activity, the volumetric wear rate per million cycles was 48 mm^3 for the conventional polyethylene liners and 10 mm^3 for the cross-linked polyethylene liners ($p = 0.0004$). Hip-simulator studies with those materials have shown wear rates of 36.8 mm^3 and 5 mm^3, respectively, per million cycles[43,46]. The reductions of wear of cross-linked polyethylene per million cycles in vivo (79%) and in vitro (86%) were similar. The increased wear resistance of cross-linked polyethylene is fueling an increase in the use of larger-diameter heads. This trend is driving debates on the minimum thickness needed for cross-linked polyethylene components and the degree of increase in volumetric wear with larger-diameter heads.

Polyethylene Wear Particles

The number, shape, and size of polyethylene wear particles are multifactorial: they are a function of the modes and mechanisms of wear that produce them, the stresses on the bearing surface, the motions, and the polyethylene molecular orientation. Most of the polyethylene wear particles produced in a prosthetic joint are micrometers to submicrometers in size and are produced in Mode 1, in very large quantities, by well-functioning joints[30]. The predominant wear mechanisms appear to involve microadhesion and microabrasion with the generation of many polyethylene particles of <1 µm in length. The resultant wear damage is predominately burnishing and scratching[30].

Techniques have been developed to isolate and analyze wear particles generated in vivo by retrieving them from periprosthetic tissues[30,59,70-74]. The concentration of debris particles from prosthetic joints is directly related to the duration of implantation[75] and can extend into the billions per gram of tissue[71-73,76]. So far, these data are available only for conventional polyethylene because of the limited number of retrieved samples of cross-linked polyethylene[27,77,78].

Substantial differences between the wear particles from cross-linked and non-cross-linked polyethylenes have been found in vitro (Fig. 4). Cross-linked polyethylenes releases a relatively high number of submicrometer and nanometer-sized polyethylene particles and relatively fewer particles that are several micrometers in dimension[79-82]. These submicrometer particles induce a greater inflammatory response in vitro than do larger particles[80-83]. Additionally, the cellular response is dependent on the shape of the particles: elongated particles generate a more severe inflammatory reaction than do globular particles[84].

Illgen et al.[81] tried to correlate volumetric wear with biologic activity in vitro. They compared the wear of a cross-linked polyethylene (Longevity; electron beam irradiation to 9 Mrad [90,000 Gy] and gas plasma sterilization) with that of a conventional polyethylene (gamma irradiation in nitrogen), as measured in a hip simulator, and then tested the biologic activity of the iso-

Fig. 5
A modern metal-on-metal modular articulation. Modularity increases reconstructive options but can also be a source of metal particles and ions. The larger-diameter articulation (36 mm in this example) reduces wear and increases the range of motion.

lated particles in cell cultures. They found a reduced relative biologic activity of the cross-linked polyethylene particles. As the number, size, and shape of the particles released by the cross-linked polyethylene liners depend on the material used[83], the mode of cross-linking[79], and patient-related wear factors, only clinical studies of each specific cross-linked polyethylene can answer the question of whether cross-linked polyethylene offers a favorable benefit-to-risk ratio.

Although the short-term clinical data on cross-linked polyethylene are encouraging, it will be possible to draw stronger conclusions after minimum five-year clinical data have been compiled and retrieval data have become available. The central issue is not linear penetration or wear rate, but the development of osteolysis, loosening, or the need for revision surgery for any reason related to the bearing.

Ceramic Femoral Heads

Another approach to reducing polyethylene wear is to improve the wear characteristics of the femoral head. In a hip-simulator study, McKellop et al.[43] demonstrated that decreased surface roughness reduces polyethylene wear. As an alternative to metallic (cobalt-chromium) heads, ceramic heads are manufactured in many variations and sizes. The ceramic materials are much harder and can be polished to a lower surface roughness (made smoother) than metal heads. Alumina (Al_2O_3) or zirconia (ZrO_2) heads have both a high hardness and a high strength, which make them more difficult to scratch, and this can reduce abrasive wear[31,85,86]. Another important issue is the better wettability of the material. Ceramics are more hydrophilic and have improved lubrication and lower friction. Hip-simulator and clinical studies have indicated that the wear of a ceramic-on-polyethylene bearing is at least equivalent to[61,87-89] or less than[60,90-92] that of a metal-on-polyethylene bearing. Wear reduction of up to 50% has been reported[31,85,90-92].

Ceramics are brittle materials, creating the possibility of a fracture of a ceramic head. A review of more than 500,000 current-generation alumina femoral heads indicated a fracture rate of 0.004% (4:100,000)[93]. Even if the number of unreported cases is assumed to be three times higher[93], the fracture rate of ceramic heads is still much lower than that of femoral stems, which is approximately 0.27% (270:100,000)[94]. Following a specific change in their manufacturing process in 1998, zirconia heads (Prozyr) from one manufacturer (Saint-Gobain Céramiques Avancées Desmarquest, Vincennes CEDEX, France, www.prozyr.com) had an increased rate of fracture. It is important to recognize that the risk of fracture of alumina and zirconia heads from other manufacturers was not affected by this change.

Zirconia has a higher hardness and burst strength than alumina, but it is not thermostable. It can undergo phase transformation, probably as a result of its reduced heat conductivity[53,95]. A practical effect of this material property is that zirconia femoral heads should not be sterilized in an autoclave. Usually yttrium oxide (Y_2O_3) is added to improve the material properties of zirconia. A new approach is the use of so-called "zirconia toughened aluminas"[96,97]. Mixing of the two materials achieves a composite with the high strength of zirconia and the thermal stability of alumina. Additional studies must be performed to evaluate the possible benefits of these composites.

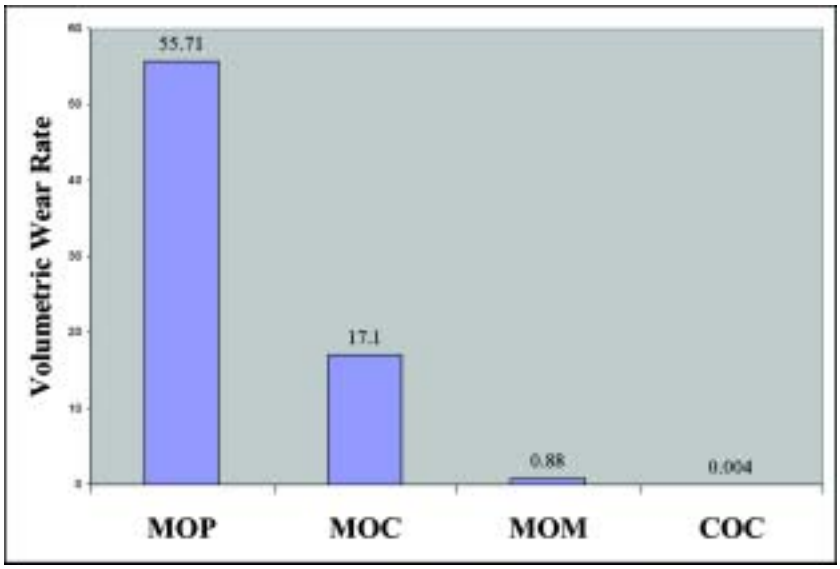

Fig. 6
Wear rates (in cubic millimeters per year) of different bearing combinations tested in a hip simulator[116]. MOP = metal-on-polyethylene, MOC = metal-on-ceramic, MOM = metal-on-metal, and COC = ceramic-on-ceramic.

Metal-on-Metal Bearings

Early loosening of prostheses with a metal-on-metal bearing was initially assumed to be due to the bearing but has now been recognized as being due to suboptimal implant design, inconsistent manufacturing, and surgical technique[53,98-101]. A review of the fifteen to twenty-year results showed that the survivorship of metal-on-metal hip prostheses is comparable with that of Charnley and other metal-on-polyethylene prostheses[102]. The failures were not due to the wear properties of the bearing[98,99,103-109]. Retrieval studies have indicated that the metal-on-metal McKee-Farrar prostheses produced substantially less wear than the conventional metal-on-polyethylene bearings did[32,53,110]. Hip-simulator studies of metal-on-metal bearings have shown a substantial (up to 200-fold) reduction in volumetric wear rates compared with those of conventional polyethylene articulations[111-118]. Consequently, there is renewed interest in metal-on-metal bearings for total hip arthroplasty, and

there has been a revival of research on and development of metal-on-metal bearings, initially in Europe[119-121] and later in the United States[122,123].

In 1988, Müller and Weber reintroduced the metal-on-metal bearing, and the development of this chromium-cobalt-alloy bearing was sold under the brand name Metasul (Centerpulse Orthopedics). With more than a decade of experience with second-generation metal-on-metal bearings, over 160,000 Metasul bearings have been implanted, and this bearing technology has also been extended to large-diameter surface-replacement components[100,124].

The interplay of materials, macrogeometry (diameter and radial clearance), microgeometry (surface topography), and lubrication influences the wear of metal-on-metal bearings to a far greater degree than it influences the wear of metal-on-polyethylene bearings[125]. Mixed film lubrication appears to be the operative mechanism in most metal-on-metal hip joints. Fluid-film lubrication is encouraged by making the femoral head as large as practically possible (doing so increases the sliding velocity and pulls more fluid into the articulation), the clearance as small as practically possible, and the surface as smooth as practically possible. With metal-on-metal bearings, in contrast to polyethylene bearings, a larger-diameter bearing actually produces lower wear rates than does a smaller-diameter bearing with similar manufacturing parameters[33,126].

The clinical outcomes associated with contemporary total hip systems with metal-on-metal bearings have generally been good[120-122]. We are not aware of any reports of reoperations for a problem directly attributable to the metal-on-metal articulation, we know of no evidence of run-away wear, and few metal particles have been seen in histological sections[120,121,124]. There have, however, been reoperations as a result of infection, heterotopic ossification, instability, impingement, and aseptic loosening. Impingement wear can be a source of metallosis, especially if a titanium-alloy neck impinges on a cobalt-chromium acetabular articulation[127]. Larger-diameter bearings have a greater arc of motion, which decreases the risk of impingement (Fig. 5).

Sieber et al.[128] reported on 118 Metasul components (sixty-five heads and fifty-three cups) retrieved because of dislocation (24%), loosening of the stem (17%), loosening of the cup (28%), or other reasons such as heterotopic ossification or infection (31%). None were revised because of osteolysis. The mean time to revision was twenty-two months (range, two to ninety-eight months). An update on this experience included 297 retrieved heads or cups[129]. The time between implantation and revision in this group ranged from one to 117 months, and the distribution of indications for the revisions was similar (dislocation in 21%, loosening of any component in 39%, and other reasons in 40%). The mean annual linear wear rates in the two studies were found to decrease with the time from the insertion of the implant: they were 25 and 35 µm in the running-in phase, decreasing to a steady state of about 5 µm after the third year in both studies. The volumetric wear rate after the running-in period was estimated to be 0.3 mm^3/yr, leading to the conclusion that these metal-on-metal bearings have a volumetric wear rate more than 100 times lower than that of conventional polyethylene bearings.

Osteolysis has been rare in clinical reports on hips with second-generation metal-on-metal bearings followed for 2.2 to five years[120-122,124]. Beaulé et al.[130], however, reported a case of progressive diaphyseal osteolysis occurring within two years postoperatively in a patient with a well-fixed cementless total hip prosthesis with a Metasul bearing. Histological analysis showed minimal wear of the bearing surface and only a small number of inflammatory cells in the tissues. As there was no evidence of a foreign-body reaction, it was hypothesized that the osteolysis was secondary to transmission of joint fluid pressure rather than induced by particles[131].

In the initial United States experience, seventy-four Metasul bearings, in Weber cemented cups, were implanted with a variety of femoral components. After follow-up periods of up to four years (average, 2.2 years), the clinical results were good to excellent and no hip prosthesis had loosened. Twenty-seven of the patients had a metal-on-polyethylene-bearing prosthesis of similar design in the contralateral hip, and none of those patients could detect a difference between the two hips[122]. Complete clinical and radiographic data on fifty-six patients (fifty-six hips), followed for four to 6.8 years (average, 5.2 years), have also been reported[132]. Good to excellent clinical results were found in 99%. One patient required acetabular revision because of loosening secondary to suboptimal cementing technique. There were no loose or revised femoral components or radiographically apparent osteolysis[132].

Metal Wear Particles and Ion Release

Wear particles from metal-on-metal bearings measure nanometers in the linear dimension and are substantially smaller than polyethylene wear particles[25,133]. The size of metal particles, as demonstrated by scanning electron microscopy studies, ranges from 0.1 to 5 µm. Scanning electron microscopy studies suggested that large metallic particles observed with light microscopy were agglomerates of smaller particles[133,134].

Little is known about the rates of metallic particle production in vivo, lymphatic transport of metallic particles from the joint, or systemic dissemination[135,136]. On the basis of information on volumetric wear rates and average particle size, it has been estimated that 6.7×10^{12} to 2.5×10^{14} metal particles are produced per year, which is thirteen to 500 times the number of polyethylene particles produced per year by a typical metal-on-polyethylene joint[133]. The aggregate surface area of these metal wear particles is substantial and may have both local and systemic effects. Surface area has been identified as a variable affecting the macrophage response to particles[137]. However, the local tissue reaction around a metal-on-metal prosthesis, indicated by the number of histiocytes, is about one grade

lower than that around a metal-on-polyethylene prosthesis[133,135]. A number of hypotheses have been proposed to explain this discrepancy[133]. Since metal particles are considerably smaller than polyethylene particles, histiocytes can store a larger number of metal particles; therefore, the total number of histiocytes required to store the metal particles is lower. Very small particles may enter macrophages by pinocytosis instead of phagocytosis, which may alter the cellular response to the particles. There may be a difference between metal wear particles and polyethylene wear particles with regard to the relative proportion that are retained locally as opposed to distributed systemically. Dissolution of metal particles results in elevation of the cobalt and chromium ion concentrations in erythrocytes, serum, and urine[138].

It is important to recognize that there may be several sources of metal particle and ion generation in modern total hip replacements. Studies have shown systemic dissemination of soluble and particulate corrosion products from modular junctions, resulting in the presence of metallic particles in the lymph nodes, liver, and spleen[139-143]. In subjects without a metallic implant, the levels of serum and urine cobalt and chromium are undetectable or nearly undetectable[140], whereas the levels of metal ions in erythrocytes, serum, and urine are elevated in patients with a metal-on-metal bearing. It appears that the ion levels are higher in the short term and decrease over time. This finding is consistent with a conditioning phase or running-in period of the bearing[144]. Since wear of a metal-on-metal bearing cannot generally be measured on a radiograph, erythrocyte, serum, and urine metal-ion concentrations may be useful indicators of patient activity and the tribological performance of these bearings. Unfortunately, the toxicological importance of these elevations in trace metal levels has not been established yet.

Delayed-type hypersensitivity, an immune response resulting from exposure to metal ions such as nickel, chromium, and cobalt, may develop in a small number of susceptible patients. Recently, some groups have described specific histological changes in the tissues around revised metal-on-metal prostheses[145-148]. They found lymphocytic infiltrations in the subsurface layer of the lining tissues, which were either diffuse or aggregated around small postcapillary vessels. The tissues from patients with a metal-on-metal implant also showed ulcerations of the pseudosynovial surface compared with those from patients with a metal-on-polyethylene implant. Interestingly, all of these changes were less obvious in tissues retrieved from patients with a McKee-Farrar or loose cemented curved cobalt-chromium stem than they were in tissues retrieved from patients with a modern metal-on-metal bearing[146]. There seems to be no correlation between the amount of metal debris and the occurrence or extent of the immunological reaction[148]. These immunological reactions are termed "aseptic lymphocytic vasculitis-associated lesions."[148] The clinical relevance of these findings is not clear yet because only a small number of patients with a metal-on-metal bearing have had to have a revision so far and only a fraction of the revised cases have shown these histological changes.

Clinically, delayed-type hypersensitivity may present as unexplained pain associated with aseptic effusions and interfacial loosening. It is unclear whether delayed-type hypersensitivity contributes to aseptic loosening or whether implant loosening contributes to delayed-type hypersensitivity[136,145,149-155]. In vitro studies have also demonstrated that polyethylene particles cause a greater inflammatory response in general but cobalt-chromium particles have higher toxicity[136,156-158].

There remains a theoretical increase in the risk of cancer with metal-on-metal bearings[143,159-166]. The aggregate clinical data have not indicated such an increase in risk, but the majority of patients in the reports presenting those data were followed for less than ten years. The latency period of known carcinogens, such as tobacco, asbestos, and ionizing radiation, is several decades. Longer follow-up of large groups of patients is needed to better assess the risk of cancer with any implant system[167]. Since the goal of more wear-resistant bearings is to reduce the need for a reoperation, theoretical risks should be weighed against the known risks of revision total hip replacement. In the Medicare population, the ninety-day mortality rate following revision total hip arthroplasty was reported to be 2.6%, which is substantially higher than that following primary total hip arthroplasty and is directly related to the revision procedure[168].

Ceramic-on-Ceramic Bearings

Ceramic-on-ceramic bearings have demonstrated the lowest in vivo wear rates to date of any bearing combination[113,169] (Fig. 6). The same principles of friction and lubrication reported for metal-on-metal bearings apply to ceramic-on-ceramic bearings. However, ceramics have two important properties that make them an outstanding material with regard to friction and wear. First, ceramics are hydrophilic, permitting a better wettability of the surface. This ensures that the synovial fluid film is uniformly distributed over the whole bearing surface area. Second, ceramic has a greater hardness than metal and can be polished to a much lower surface roughness. Although the better wettability results in a fluid film that is slightly thinner than that found with metal-on-metal bearings, that is compensated for by the reduced size of the asperities on the surface. Overall, this results in a favorable higher λ ratio and in a reduced coefficient of friction. This bearing combination is the most likely to achieve true fluid-film lubrication[33]. However, because of the hardness of ceramics, the wear characteristics are sensitive to design, manufacturing, and implantation variables. Rapid wear has also been observed, generally associated with suboptimal positioning of the implants[170,171].

Ceramic-on-ceramic bearings currently in clinical use are made of alumina. Developments in the production process (sintering) have improved the quality of the material[172]. Modern alu-

mina ceramics have a low porosity, low grain size, high density, and high purity. Thus, hardness, fracture toughness, and burst strength are increased[172-174]. There have been in vitro tests utilizing zirconia and alumina-zirconia composites in order to improve wear characteristics[96,97], but these mixed oxides must be studied further before clinical trials can be conducted.

The United States experience with ceramic-on-ceramic bearings was initially limited to the Autophor/Xenophor prostheses, which were conceived and introduced in Europe by Mittelmeier[175,176]. The clinical results with the Autophor prosthesis were generally less satisfactory than those with established metal-on-polyethylene designs, and ceramic-on-ceramic implants were never widely used in the United States[177]. Previous studies (mostly from Europe) showed prosthetic survival rates of 75% to 84% after ten years[171,178,179] and 68% after twenty years[18]. For patients fifty years of age and younger, the prosthetic survival rates were 84% after ten years, 80% after fifteen years[179], and 61% after twenty years[18].

Similar to the situation with metal-on-metal bearings, the perception of which was based on the clinical performance of the McKee-Farrar prosthesis[98], the perception of the performance of ceramic-on-ceramic bearings has been complicated by the fact that the Autophor stem and socket had features that are now recognized as suboptimal[170]. Follow-up studies showed very low in vivo wear rates[18,180,181], but failures occurred as a result of inferior implant design and fixation technique[174,182]. The current generation of ceramic-on-ceramic bearings is frequently being utilized in implant systems that have demonstrated long-term successful fixation and excellent clinical results with a metal-on-polyethylene bearing.

Two prospective, randomized multicenter trials are being performed in the United States, with more than 300 patients enrolled in each study. Garino[183] reported the experience with the Transcend system (Wright Medical Technology, Arlington, Tennessee). A modular cementless acetabular component with either a cemented or a cementless stem was implanted in 333 hips. After a duration of follow-up of eighteen to thirty-six months (mean, twenty-two months), 98.8% of the implants were still in situ. In the second trial, the ABC System (Stryker Howmedica Osteonics) was implanted in 349 hips[184,185]. D'Antonio and Capello[184] evaluated the most recent results of the six participating surgeons with the highest number of enrolled patients. This subgroup consisted of 207 patients with 222 hips followed for a mean of forty-eight months. Five hips had been revised, and 97.7% of the implants were in place. An additional, nonrandomized arm of this multicenter study consisted of 209 patients treated with the Trident system (Stryker Howmedica Osteonics)[186]. One hundred and seventy-five of them had been followed for a minimum of two years, and the revision rate in that group was 1.7% (three revisions).

One potential complication with these implants that a surgeon should recognize is chipping of the liner during insertion. This happened in three cases (1%)[183] treated with the Transcend system and in nine (2.6%)[186] treated with the ABC system. The Trident bearing has a metal-backed ceramic insert with an elevated titanium liner rim, and no intraoperative chipping of that liner was reported[184].

Including the original study groups and additional implantations, at the time of this writing 1361 ceramic inserts have been implanted in these studies and there have been no failures due to the bearing[183,184]. No fractures of the implanted liners or ceramic heads have been reported. The incidence of fractures of current-generation ceramic heads[93] is four in 100,000. It is too early to make a comparable statement about the acetabular inserts as more data on a higher number of implants are needed. Results from the multicenter studies are encouraging, with no liner fractures reported to date.

Ceramic Wear Particles

Ceramic materials may have better biocompatibility than metal alloys[187], but the relative size, shape, number, reactivity, and distribution (local compared with systemic) of the respective wear particles have not been fully determined. Hatton et al.[188] reported a bimodal size range of particles isolated from tissue around failed ceramic-on-ceramic total hip replacements. They found a large amount of small particles between 5 and 90 nm (mean, 24 nm) but also larger particles between 0.05 and 3.2 µm. Ceramic debris may not be bioinert as initially assumed because osteolysis has been described in some patients with a ceramic-on-ceramic bearing[26,189]. Recently, some studies have demonstrated inflammatory and cytotoxic reactions on the cellular level, but the relationship to material, size, and particle number remains uncertain[188,190-192]. It seems that there is less inflammatory reaction than that found with metal-on-metal or metal-on-polyethylene bearings in well-functioning prostheses[193]. Ion toxicity is not an issue with ceramics because of their high corrosion resistance[193].

Overview

Cross-linked polyethylene, metal-on-metal, and ceramic-on-ceramic bearings have all demonstrated lower in vivo wear rates than conventional metal-on-polyethylene couples. The degree of wear reduction is promising, but it may not directly translate into greater longevity of a total hip replacement in all patients. Continued close follow-up is needed to demonstrate a favorable benefit-to-risk ratio based on a reduction in the number of revision operations (Table I). The use of any of these bearings has specific benefits and risks that should be considered on a patient-by-patient basis.

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