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Ceramic-on-Ceramic Bearings in Total Joint Arthroplasty. Part 1

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Summary. *Ceramic bearings were first employed as alternatives to polyethylene (PE) bearings in total joint arthroplasty about a decade after Sir John Charnley introduced the first durable total hip arthroplasty (THA) with a metal-PE articulation. Charnley's approach was based on a metal stem bonded to bone with polymethylmethacrylate (PMMA) and an acetabular component made of ultra-high-molecular-weight PE (UHMWPE). Microscopic particulate debris in the joint space from bearing wear has been shown to lead to periprosthetic inflammation, osteolysis, and implant loosening. Cross-linking can reduce the wear of UHMWPE, but it also compromises UHMWPE's mechanical properties. Accordingly, there are concerns related to potential brittleness if UHMWPE implants are not positioned optimally. Also, the smaller particles generated from cross-linked UHMWPE may present an increased particulate load in vivo. Thus, there is a need for data on the long-term outcomes of cross-linked UHMWPE. Any technology that can reduce bearing wear rates in THA and total knee arthroplasty (TKA) can potentially decrease the morbidity and risks associated with premature revision surgery related to wear. Improved wear resistance also allows the use of large-diameter femoral heads in THA, leading to increased arc of movement and less risk of prosthesis dislocation. The ideal joint bearing for THA and TKA would be able to withstand high cyclic loading for several decades without undergoing corrosion or fretting at modular metal tapers, and would possess proven biocompatibility and material stability in vivo, as well as ultralow wear rates. The search for the ideal total joint bearing has led to the development of ceramic bearings.*

Key words: *total hip arthroplasty; total knee arthroplasty; ceramics; polyethylene; bearing wear.*

Introduction

Bearings made of ceramics (e.g. alumina [aluminum oxide] and zirconia [zirconium oxide]) have been shown to possess extremely low wear properties that make them suitable for both THA and TKA. When compared to the most commonly used bearing couple in joint arthroplasty, which consists of cobalt-chrome (CoCr) metal alloy articulating against UHMWPE, ceramic surfaces offer significant reductions in bearing wear rates.

The superior wear characteristics of ceramic materials have been verified in many clinical and experimental studies. In one study, alumina-alumina articulations in THAs showed less osteolysis in the proximal femur than the metal-UHMWPE controls at 5 years after surgery [1]. Long-term clinical outcomes have shown few, if any problems with alumina total hips, in the absence of confounding variables [2].

Metal-on-metal bearings also reduce THA wear dramatically, but metal wear particles can lead to delayed hypersensitivity reactions, and the long-term effects of systemically dispersed fine metal wear particles remain a matter of speculative concern.

More than half a million total joint arthroplasties are performed annually in the United States, and this number is growing. Worldwide, millions of femoral heads have been implanted. Ceramic bearings have not been as well accepted among US hip surgeons as other bearing types have been, because of concerns related to cost, complexity, lack of familiarity, and problems such as potential catastrophic rupture. At present, ceramic bearings are used in a minority of THAs done in the United States.

Ceramic technology continues to evolve, and new materials based on nonoxide ceramics, composites of existing ceramics, and surface modifications will offer more options to the arthroplasty surgeon. Previous experience has also shown that each new bearing technology applied to total joint replacement can have unforeseen complications [3, 4]. Ceramics will continue to be developed for clinical use as the underlying engineering and material science principles

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of new materials are validated and as clinical data demonstrate their safety and reliability *in vivo*.

Evolution of ceramic bearings

Ceramic-on-ceramic (COC) bearing surfaces have a long history of successful clinical use [5]. Ceramics was first used in hip arthroplasty by Pierre Boutin in 1970 [6] but has evolved to address some of the early limitations, particularly fracture, which occurred as a consequence of the sintering process resulted in large grain size and subsequent ease of crack propagation [7]. Second-generation ceramics treated with hot isostatic pressing had smaller grain sizes and fewer impurities [8]. Zirconia was a second-generation ceramics introduced in 1985 because of its improved fracture toughness and bending strength compared with alumina, but it was subsequently found to have inferior wear characteristics [9]. Zirconia was vulnerable to undergoing transformation at high temperatures and wet environments. This transformation weakened the zirconia and increased its surface roughness [10]. The third-generation ceramics developed in the 1990s was marketed as Alumina Forte (BIOLOX^{s}forte, CeramTec AG, Plochingen, Germany). It showed continued improvements in manufacturing, creating a purer, denser ceramics, but was still vulnerable to rim fracture, particularly of the liner. The development of current fourth-generation Alumina Delta (BIOLOX^{s}delta, CeramTec AG, Plochingen, Germany) has further addressed the limitations of the alumina. This modern ceramics is a compound of zirconia-toughened alumina, strontium, yttria, and chromia (SrO, Y^2O^3, and Cr^2O^3) [11]. The addition of strontium limits crack propagation and, together with the chromia, improves the hardness of the composite. The zirconia improves the toughness and wear characteristics and is stabilized from undergoing transformation by the yttria [11].

Ceramic materials are extremely hard, scratch resistant, and biocompatible, as well as demonstrating a low coefficient of friction. This makes ceramics an ideal bearing material for total hip arthroplasty.

These are formed by fusion of microscopic grains of alumina (Al^2O^3) and/or zirconia (ZrO^2) ceramic powder into a solid phase (Table 1). The process of sintering is “hot isostatic pressing” requiring temperatures exceeding 1400 °C and pressures above 1000 Bars. After sintering, the components are ground and polished to get the finest surface possible. The manufacturing of COC bearings for orthopedics is under strict control (more than 50 checkpoints according to the declaration of CeramTech AG, Plochingen, Germany) and in accordance with international quality standards (ISO 6474). Compared to other currently used bearing couples, modern COC bearings demonstrate the lowest wear rates both *in vitro* and *in vivo*.

Tribological remarks

Current ceramics used for manufacturing bearing surfaces in THA exhibit outstanding tribological properties, the most important of which are hardness and high degree of wetability. Ceramics has a greater hardness than metal and can be polished to a much lower surface roughness, while excellent wetability ensures that the synovial fluid is uniformly distributed between implant surfaces [12]. The former guarantees high resistance to major scratches and undetectable wear rate, while the latter facilitates fluid-film lubrication thus contributing to very low friction between articulating surfaces ($< 1.7x10^{-7}$ mm^3/Nm) [13].

The basic mechanism of wear in COC articulations is intergranular erosion followed by isolated grain pull-out [12]. In fact, hip simulator studies of current COC bearings have shown very low wear rates (less than 0.1 mm^3 per million cycles) [14]. However, measurement of retrieved ceramic implants revealed much higher wear rates than above (≥ 1 mm^3/yr) and a characteristic “ceramic” wear pattern was noted [15-17]. The reason for these differences could lie in different biomechanical conditions *in vitro* and *in vivo*, with the latter being exposed to edge loading, recurrent separation of bearing surfaces and even direct impingement of the ceramic implant on the neck of the stem,

Table 1

Characteristics of ceramic materials used currently in total hip arthroplasty

Type of ceramic material	Grain size (μm)	Density (g/m^3)	Bending strength (MPa)	Fracture toughness (MPa.m$^{1/2}$)	Vickers hardness	Young’s modulus (GPa)
Aluminia (Biolox Forte)	<2	3.98	580	4	20	380
Zirconia	<0.5	n.a.	>900	8	12.5	210
TAMC (Biolox Delta)	<2	4.37	>1380	6.5	19	>350

ZTAMC – Zirconia-toughened alumina matrix composite; n.a. – not available. Sources: CeramTec AG, Plochingen, Germany which increases the total wear of the implant *in vivo*. However, even under microseparation conditions, the wear rates of current alumina and ZTAMC ceramics are lower than highly cross-linked polyethylene (up to 1.8 mm^3/million cycles) [16].

Size of ceramic particles

Ceramic wear particles are continually released into the effective joint space during each step similar to non-COC THA. Depending on the mechanism of wear, ceramic particles are typically generated in smaller numbers and with a bimodal size range involving nanometer size particles (mean 24 nm; range 5 to 90 nm) and larger particles (mean 0.43 μm; range 0.05 to 3.2 μm) probably associated with grain boundary fracture [18, 19]. In addition, even larger ceramic particles are generated during gross damage (catastrophic failure) of the bearing surfaces.

Biological activity of ceramic particles

Prosthetic particles released from artificial joints stimulate periprosthetic cells to produce an inflammatory and pro-osteolytic environment leading eventually to alteration of local bone homeostasis in favour of bone resorption. Generally, the impact of particle load on the extension of bone defects depends at least on the size, amount, origin, and shape of the particles [20].

COC THAs exhibit very low ceramic wear rates and, in addition, ceramic wear particles have much lower specific and functional biological activity than polyethylene particles [21, 22]. Catelas et al. showed that polyethylene particles stimulated greater release of TNF-α when compared to alumina or zirconia [23]. Kubo et al. found much less intense histiocytic response around particles of alumina ceramics (3.9 μm in diameter) than that of UHMWPE (11 μm), stainless steel (3.9 μm), and CoCr (3.9 μm) in a rabbit model [24]. Bos et al. studied macrophages in the pseudo-synovial membrane from well-fixated implants retrieved at autopsy and found the percentage of macrophages was higher in the polyethylene-on-ceramic and metal-on-polyethylene groups (40-60%) than in the ceramic-on-ceramic group (20-40%) [25, 26]. On the other hand, at least one study comparing macrophage apoptosis as a result of stimulation by alumina, zirconia, and PE particles found the response to be size and concentration dependent, rather than particle composition dependent [27]. The overall impression is that ceramic particles are biologically inert, but if released in sufficient numbers (e.g. cases of neck impingement or third body wear), ceramic particles can produce osteolysis similar to that induced by PE particles. In comparable doses, however, the biologic response is less intense with ceramic versus PE particles.

From the above, it could be deduced that osteolysis and aseptic loosening will be obviated in patients with COC THA. Unfortunately, this is controversial because several studies demonstrated periprosthetic osteolysis even in patients with current COC THA [28, 29]. The reason may lie in the multifactorial origin of osteolysis and aseptic loosening when particle related parameters play an important role but not the only pathway inducing these entities [30]. In addition, ceramic bearing surfaces are not the only source of prosthetic particles. In support of this is a histological study of pseudomembranes from loosened alumina cups that suggested that this "unexpected" osteolysis was probably due to metal or cement debris rather than alumina particles. Thus, in terms of biological activity of ceramic particles, the advantages clearly outweigh the disadvantages.

Clinical evidence for ceramic-on-ceramic THA

Assuming that COC bearings offer the lowest wear rates and that ceramic particles induce minimal adverse biological activity, do these facts result in overall improvement in survivorship of THA?

Recent systematic reviews on survivorship of hard-on-hard bearings in THA revealed variable implant longevity and rates of complications in earlier studies (survival rates of 73% to 100% at mean follow up ranging from 31 to 240 months) [31]. Early generations of ceramic-on-ceramic implants were characterized by high failure rates as a result of both component fracture and loosening of the monolithic acetabular component. However, in a recent retrospective study, Petsatodis et al. reported a survivorship of 84.4% of cementless alumina COC prostheses at 20 years follow-up [32]. Others have reported significant differences in survivorship of COC bearings depending on the type of prosthesis and its fixation, especially with respect to cementless and cemented cups [33, 34]. Therefore, the survivorship and rate of complications of ceramic bearing surfaces depend not only on the period of implantation (and therefore the generation of ceramic material) but also on other important factors, e.g. design of the prosthesis, surgical technique and the method of femoral and acetabular fixation.

The new generations of ceramic implants suggest more promising outcomes (Table 2), especially in young and active patients, with survivorship rates (free of revision) between 92% and 99% at ten years of follow-up [35-39]. However, these data are comparable but not better than the best outcomes for both metal-on-metal and metal/ceramic-on-polyethylene articulations (Table 2). In addition, the number of studies and length of follow-up for COC bearings are still insufficient compared to ceramic/metal-on-polyethylene THAs. Finally, the strength of evidence might be further compromised by methodological weaknesses as was reported for other clinical research in orthopedics [40].

Table 2

Summary for a review on ceramic-on-ceramic THA

Parameter	Ceramic-on-Ceramic	Ceramic/Metal-on-Polyethylene	Metal-on-Metal
Wear rate	30.5±7 μm/yr [43]	218.2±13.7μm/yr [43]	20–25 μm/yr [45]
Particle size	0.13–78 μm [45]	30 nm–10 μm [46]	30–100 nm [47]
Cellular response to wear particles	Low	High	High
Hypersensitivity induced by wear debris	No	No	Yes
Tissue necrosis, ALVAL	No or weak	Weak	High grade
Dislocation#	0.78%	0.80%	0.74%
Infection#	0.32%	0.49%	0.53%
Mechanical loosening#	0.39%	0.22%	0.20%
Revision#	1.02%	1.16%	1.12%
Noisy hip	Up to 33%	Rarely	Less frequent
Survivorship, 10 yrs. FU	99% (95% CI; 97-100%) [37]	95.6% (95% CI; 90.1-98.3%) [37]	95.4% (95% CI; 85.8–99.8%) [38]
Survivorship, 20 yrs. FU	84.4% (95% CI; 0.56–1.33) [32]	81.8% (95% CI; 79.0–84.6%)*	84% NA [42]

*# – up to 2 years of follow-up [41]; ALVAL – aseptic lymphocytic vasculitis-associated lesions; FU – follow-up; * for all diagnoses and all reasons for revisions (Swedish Hip Arthroplasty Report 2008); NA – not available*

As a result the conclusion is that the use of highly wear resistant bearing surfaces does not automatically guarantee longer survivorship than the best non-COC THAs. The reason lies at least partially in the occurrence of other unrelated complications (e.g. deep sepsis, instability, periprosthetic fracture, etc.) that require revision surgery and that are not prevented by simple choice of bearing surface. Even aseptic loosening cannot be completely resolved using one specific bearing surface because of its multifactorial etiology [42]. On the other hand, the rate of osteolysis was diminished as a direct consequence of using ceramic bearings. Taken together, combining the best design of THA with COC bearings might improve the long-term outcomes. However, this remains to be demonstrated in well-conducted multicenter studies and/or arthroplasty registries data.

Conflict of interests. The author declares no conflict of interest towards the present article.

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Керамо-керамічні пари тертя в тотальному ендопротезуванні суглобів. Частина 1

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Резюме. *Керамічні поверхні вперше були використані як альтернатива поліетиленовим (PE) поверхням у тотальному ендопротезуванні суглобів приблизно через десять років після того, як сер Джон Чанлі представив вперше тотальне ендопротезування кульшового суглоба (THA) з метало-поліетиленовою парою тертя. Підхід Чанлі був заснований на наявності металевої ніжки, прикріпленої до кістки поліметилметакрилатним кістковим цементом, та ацетабулярному компоненті, виготовленому з поліетилену надвисокої молекулярної маси. Його роботи продемонстрували, що мікроскопічні часточки в суглобовій щілині від зносу поверхонь призводять до перипротезного запалення, остеолізу та розхитування компонентів імплантату. Створення поперечних зв'язків у поліетилені (крос-лінкований поліетилен) може зменшити зношування останнього, але воно також ставить під загрозу механічні властивості поліетилену. Відповідно, існує занепокоєння, пов'язане з потенційною крихкістю, якщо імплантати з поліетилену не розміщені оптимально. Крім того, менші частинки, утворені з крос-лінкованого поліетилену, можуть чинити підвищене навантаження на поверхню імплантату. Будь-яка технологія, яка може знизити швидкість зносу пар тертя при THA та тотальному ендопротезуванні колінного суглоба (TKA), потенційно здатна зменшити захворюваність і ризики, пов'язані з передчасною ревізійною операцією, спричиненої зносом. Покращена зносостійкість також дозволяє використовувати головки стегнової кістки великого діаметра в THA, що приводить до збільшення дуги руху та зменшення ризику вивиху протеза. Ідеальна пара тертя для THA і TKA могла б витримувати високе циклічне навантаження протягом кількох десятиліть, не зазнаючи корозії або деформації на модульних металевих конусах, і мала би доведену біологічну сумісність і стабільність матеріалу in vivo, а також наднизьку швидкість зносу. Пошук ідеальних пар тертя для тотального ендопротеза привів до розробки керамічних компонентів.*

Ключові слова: *тотальне ендопротезування кульшового суглоба; тотальне ендопротезування колінного суглоба; кераміка; поліетилен; тертя поверхонь.*