Ceramics are non-metallic inorganic materials with a broad range of composition. They are usually processed by mixing the particulates of the material together with water and an organic binder. The mixture is then pressed into a mould to obtain the desired shape, dried to evaporate the water and the binder burned out by thermal treatment. Firing or sintering at a very much higher temperature then densifies the residual material. The final microstructure of the ceramic is greatly dependent on the thermal process applied, the maximal temperature reached and the duration of the thermal steps. There are five types: glass, plasma-sprayed polycrystalline ceramic, vitrified ceramic, solid-state sintered ceramic and polycrystalline glass-ceramic. Other factors such as the purity of the powder, the size and distribution of the grains, and the porosity are important in determining the mechanical and biological properties. The extended use of bioceramics in medicine is related to their excellent biocompatibility as a result of their high level of oxidation.

Ceramics used in orthopaedic surgery are classified as bioactive or inert according to the tissue response when implanted in an osseous environment. The bioactivity of a material can be defined as its ability to bond biologically to bone. An inert ceramic merely elicits a minor fibrous reaction. In clinical practice, inert fully-dense ceramics are used as bearings in total joint replacements because of their exceptional resistance to wear and their tribological properties. By contrast, bioactive ceramics are employed as coatings to enhance the fixation of a device or as bone-graft substitutes because of their osteoconductive properties.

This review discusses the clinical applications of ceramics in orthopaedic surgery and the future trends in their design.

Sliding ceramics. The most widely used bearing couple in total joint replacements remains metal-on-polyethylene. The long-term survival of the artificial joint, however, is impaired by the wear of its components which ultimately leads to osteolysis around the implant secondary to an inflammatory response induced by wear debris occurring from both the articulating and non-articulating surfaces. Of the different types of particle found in the membranes surrounding the loosened components, polyethylene debris has been identified as a major factor inducing activation of macrophages which leads to the production of bone-resorbing cytokines with resultant loss of bone stock, especially in the young and active population. Pierre Boutin first introduced ceramics in orthopaedics in the early 1970s in order to eliminate or reduce complications related to polyethylene. They are mainly used in total hip arthroplasty as femoral heads articulating against polyethylene, and as cups in the alumina-on-alumina combination.

Because of their relatively brittle nature, fracture of ceramic femoral heads has been, along with cost, the main limitation to their expanded use world-wide. The risk of fracture, however, has been virtually eliminated because of a great improvement in the manufacturing process with increased purity and density, improvement in the size and distribution of the grains and better quality control. Accurate fixation of the ceramic ball to the femoral stem through a well-designed Morse taper avoids critical stresses in the head and a better surgical technique has been developed. It is difficult to determine the actual incidence of fracture, the frequency of which is very variable. When the above conditions are optimised, however, the rate has been evaluated as 0.02% for alumina heads and 0.03% for those of zirconia. These figures indicate that fracture of the ceramic head is no longer an important issue, and in our opinion should not be a relevant argument against its use.

Alumina ceramic. Dense alumina of surgical grade is obtained by sintering alumina powder at temperatures between 1600 and 1800°C. The resultant material is in its highest state of oxidation, allowing thermodynamic stability, chemical inertness and excellent resistance to corrosion. Improvement in the manufacturing process has lowered the size and distribution of the grains, which are major factors in avoiding the propagation of cracks and fracture. Alumina
is a brittle material with excellent compression strength but the bending strength is limited. The Young’s modulus is 300 times greater than that of cancellous bone, and 190 times higher than polymethylmethacrylate (PMMA). Alumina has been a standardised material since 1984 (International Standard Organisation, ISO 6474). The properties of dense alumina of surgical grade are summarised in Table I.

The tribological properties of alumina ceramic against itself are outstanding with a linear wear rate 4000 times lower than that of metal-on-polyethylene. The excellent frictional characteristics are due in part to a high wettability because of the hydrophilic surface and fluid film lubrication which minimises adhesive wear. These properties, demonstrated both in vitro and on analysis of retrieved implants, are responsible for the limited amount of wear particles produced and the subsequent moderate biological reaction to ceramic wear debris. The clearance between the two components in the case of the alumina-on-alumina combination should be around 50 μm to avoid Hertz stresses at the surface of the alumina which may result in detachment of grains and third-body wear.

Clinical experience over a 20-year period at the authors’ institution using the alumina-on-alumina combination has given favourable results depending upon the method of fixation used for the socket. With cemented alumina cups, the rate of survival at ten years was 88.6% when aseptic loosening was considered as the end-point. Patients younger than 50 years of age had a survival of 94% at ten years. With bulk alumina press-fit sockets, the rate of survival was 94.6% at six years. The main cause for revision was aseptic loosening of the acetabular component. Osteolysis around the loosened components was limited, and during most revision procedures bone graft was not needed. We have recently reviewed at 20 years a consecutive series of 118 hips with 86 cemented sockets and 32 cementless acetabular components with three pegs. The cementless sockets performed significantly better than the cemented with a survival free from revision of 88.5%. No wear was measurable on plain anteroposterior radiographs of the pelvis (Fig. 1). Boehler et al in Austria have reported similar encouraging results using cementless sockets. Experience in the USA with the Mittlemeier alumina-on-alumina prosthesis has not been so favourable with revision as high as 27% at 26 months. It is likely that the poor quality of the alumina and inadequate design of the implant are responsible for this unacceptable rate of failure. Proper quality of design and manufacturing is the best warranty of a low rate of wear and limited osteolysis. We currently use press-fit titanium shells with a hydroxyapatite (HA) coating and a modular alumina liner snapped into the shell at the time of operation.

Alumina-on-polyethylene has also been evaluated, but few results are available in the literature. Studies comparing alumina with metal articulating against polyethylene suggest that polyethylene wear is reduced two-fold, but interpretation of these studies is difficult since the metal head usually measures 32 mm, which is a major limiting factor of the metal-polyethylene combination. Wroblewski, Siney and Fleming recently described the results of 22 mm alumina femoral heads articulating against cross-linked polyethylene in a ten-year follow-up. A running-in rate of penetration was noted which then decreased to 0.022 mm/year after the first 18 months. It is difficult, however, to conclude from this study whether the alumina head, the small diameter of the femoral head, or the cross-linked nature of the polyethylene cups was responsible for the low rate of wear observed.

<table>
<thead>
<tr>
<th>Property</th>
<th>Alumina</th>
<th>Zirconia</th>
</tr>
</thead>
<tbody>
<tr>
<td>Purity (%)</td>
<td>&gt;99.8</td>
<td>97.0</td>
</tr>
<tr>
<td>Density (g/cm³)</td>
<td>3.98</td>
<td>6.05</td>
</tr>
<tr>
<td>Grain size (μm)</td>
<td>3.6</td>
<td>0.2 to 0.4</td>
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<tr>
<td>Surface finish (Ra, μm)</td>
<td>0.02</td>
<td>0.008</td>
</tr>
<tr>
<td>Bending strength (MPa)</td>
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<td>1000</td>
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<tr>
<td>Compressive strength (MPa)</td>
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<td>2000</td>
</tr>
<tr>
<td>Young’s modulus (GPa)</td>
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<td>210</td>
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<tr>
<td>Hardness (Vickers hardness number)</td>
<td>2000</td>
<td>1200</td>
</tr>
<tr>
<td>Fracture toughness KIC (MN/m³/²)</td>
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</tbody>
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Fig. 1
A cementless alumina-on-alumina total hip replacement at 20 years. No wear of either component is measurable and no osteolysis is present. (From P. Boutin’s collection.)
Zirconia ceramic. Zirconia ceramic was introduced in the manufacture of femoral heads for total hip replacements because of its higher strength and toughness which would reduce the risk of fracture. Pure zirconia is an unstable material showing three different crystalline phases: monoclinic, tetragonal and cubic. The phase changes result in a large variation in volume and significantly decrease the mechanical properties of the material due to the production of cracks. Stabilisation of zirconia by adding oxides to maintain the tetragonal phase has therefore been undertaken. Yttrium-stabilised tetragonal polycrystalline zirconia (Y-TZP) has a fine grain size and offers the best mechanical properties (Table I). This material was standardised in 1997 (International Standard Organisation, ISO 13356). Zirconia femoral heads should articulate only against polyethylene sockets since both zirconia against alumina and zirconia against zirconia have been shown to produce catastrophic rates of wear in vitro. Zirconia-on-polyethylene has demonstrated similar rates of wear as alumina-on-polyethylene in vitro, but in vivo the results have not been so favourable. Allain et al recently described a consecutive series of 78 hips using a zirconia femoral head and a polyethylene cup. Complete radiolucent lines were observed around the cup in 23% of the hips and 17% of the femoral implants had radiolucency greater than 1 mm. Survival at eight years was 63%. These worrying results were confirmed by Hernigou and Babrami in a study comparing wear of the cup and osteolysis in 40 hips over a period of ten years. Two comparable groups of 20 hips each had either a 32 mm alumina or a 28 mm zirconia femoral head. During the first five years, the zirconia group had a lower rate of wear of 0.04 mm/year compared with 0.08 mm/year, and osteolysis on the calcar measured in square millimetres was similar in both groups. Between five and ten years, however, the rate of wear increased dramatically in the zirconia group to 0.15 mm/year at ten years as opposed to 0.07 mm/year in the alumina group. Osteolysis of the calcar was significantly greater in the zirconia group at 135 mm² compared with 65 mm². These results are of concern. The long-term performance of zirconia ceramic may well be altered by degradation in vivo with transformation of the material into its monoclinic unstable phase. Another explanation was suggested by Lu and McKellop, who measured the frictional heating of polyethylene cups in a hip joint simulator. The steady-state temperature of the polyethylene reached 99°C with heads of zirconia ceramic compared with 45°C with alumina prostheses. This may account for the long-term wear because of the consequent structural changes and may also produce precipitation of lubricant proteins. Better results are apparent in current clinical trials.

Mixed-oxide ceramics. A new class of materials has been developed recently to combine the tribological properties of alumina and the mechanical characteristics of yttrium-stabilised zirconia. These mixed-oxide ceramics containing 40% to 80% zirconia have shown rates of wear in vitro comparable to those of alumina ceramic. Preliminary results in hip joint simulators have been promising, but further investigations are needed to assess their long-term performance.

Bioactive ceramics. Bioactive ceramics are osteoconductive, acting as a scaffold to enhance bone formation on their surface, and are used either as a coating on various substrates or to fill bone defects. An osteoconductive material can only elicit bone formation in an osseous environment, whereas an osteoinductive substance can promote bone formation even in an extraosseous situation.

Calcium phosphate ceramics. Two bioceramics belonging to the calcium phosphate family have had extensive evaluation as orthopaedic implants, namely HA and tricalcium phosphate (TCP). Stochiometric synthetic HA (Ca₁₀(OPO₄)₆(OH)₂), with a calcium-to-phosphate atomic ratio...
ratio of 1.67, was introduced as a bone-graft substitute because its formula is similar to that of the inorganic mineral phase of bone. Biological HA, however, is Ca-deficient and a carbonated apatite. The bonding mechanism of HA to bone, although not completely understood, seems to be due to the attachment at the surface of the HA of osteogenically-competent cells which differentiate into osteoblasts. A cellular bone matrix is then formed at the surface of the HA. An amorphous area is present between the surface and the bone tissue containing thin apatite crystals. As maturation occurs, this bonding zone shrinks and HA becomes attached to bone through a thin epitaxial layer, resulting in a strong interface with no layer of fibrous tissue interposed between the bone and HA. Bone formation grows from the surface of the HA towards the centre of the pores.

HA coating is widely used on femoral prostheses and on sockets as a means of fixation in order to avoid complications related to the use of PMMA. It is usually applied by plasma spray. An American multicentre study has reported excellent results, with a rate of femoral revision of 0.3% at a mean follow-up of 8.1 years, with one case of loosening out of 324 implants. Similar results have been reported by Geesink and Hoefnagels with a rate of survival of 98% for the femoral component at a mean follow-up of eight years in a consecutive series of 118 hip replacements. It was concluded from these studies that bone ingrowth or ongrowth was enhanced by HA. The incidence of femoral osteolysis was not increased and the long-term resorption of the HA coating did not seem to alter the clinical and radiological results. Complications such as third-body wear due to HA particles or increased osteolysis have been reported. The results from controlled studies comparing HA with other non-cemented methods of fixation are contradictory, some showing advantages for HA-coated implants over non-coated prostheses and others not. It has not yet been clearly shown that HA offers improved fixation when compared with bone cement. The thickness of the coating, the chemical composition of the material and the roughness and nature of the metal substrate seem to be key factors in ensuring good results.

The main disadvantages which have limited the clinical application of HA as a bone-graft substitute are related to the brittle nature and poor tensile strength of the material. Consequently, information on the clinical use of ceramic bone-graft substitutes is scarce. Oonishi et al. have described their experience with particulate HA in 40 acetabular reconstructions of failed aseptic total hip replacements, followed up for between four and ten years. Radiolucent lines at the HA-bone interface disappeared within three months. Three sockets migrated in hips with combined cavitary and segmental defects. This material is an interesting alternative to allograft bone in these complex situations.

The senior author (LS) has used particulate HA with βTCP in conjunction with morsellised allograft bone in femoral reconstructions employing the impaction grafting technique. No failure has been observed in this limited series.

Tricalcium phosphate (Ca₃(PO₄)₂) exists in either α- or β-crystalline forms. The β-form is the most stable with a calcium-to-phosphate atomic ratio of 1.5. These ceramics are considered to be resorbable since the rate of biodegradation is higher when compared with HA. Degradability occurs by combined dissolution and osteoclastic resorption. TCP has been evaluated in spinal fusion with results comparable to those with autogenous bone.

**Bioactive glasses.** Bioactive glasses were first developed by Hench and Wilson and have a vitreous structure. They bond chemically to bone. The model in this class of materials is Bioglass 45S5 of which the composition in weight % is: 45% SiO₂, 24.5% CaO, 6% P₂O₅ and 24.5% Na₂O. The bonding mechanism of silicate bioactive glasses to bone has been attributed to a series of surface reactions ultimately leading to the formation of a hydroxy carbonate apatite layer at the glass surface. The critical element necessary for the formation of this is the production of a layer of porous silica gel with a high surface area (Fig. 2). Greater production of bone has been demonstrated with Bioglass 45S5 when compared with HA, but due to its poor mechanical properties this material has not been used in load-bearing applications.

Recently, in order to improve the reactivity of the material, sol-gel processed glasses, hydrolysed at ambient temperatures, have been developed to obtain bioactive gel-glasses in the SiO₂-CaO-P₂O₅ system, with an initial high specific surface area. These materials have similar osteo-conductive properties to melt-derived glasses, but have an improved degradability. The low temperatures used to produce sol-gel glasses allow them to be used as a coating on alumina substrates. When implanted in an animal model, sol-gel glass-coated alumina has demonstrated the ability to form an interface mainly composed of newly-formed bone by 24 weeks (Fig. 3).

In this class of material, apatite wollastonite (CaO·SiO₂) glass ceramic developed by Kokubo et al. has osteo-conductive properties similar to Bioglass 45S5 but increased mechanical strength. It has been used as a spacer at the iliac crest, for vertebral prostheses and as a shelf in procedures about the shoulder with favourable results.

**Bioactive bone cements.** Bioactive bone cements have been explored as an alternative to PMMA in order to avoid complications related to PMMA debris and to enhance fixation of the prosthesis. These materials have undergone extensive basic research. They include calcium-phosphate-based bone cement and glass-ceramic bone cement. A strong cement-bone interface is obtained by the formation of HA at the surface of the cement. Moreover, calcium phosphate cements are resorbable and are progressively replaced by newly-formed bone. To our knowledge, however, clinical trials of such materials have not yet been undertaken.
Conclusions
The excellent biocompatibility and outstanding tribological properties of inert ceramics have encouraged their use as bearings in total joint replacements. Improvement in the manufacturing process has allowed the production of reliable materials. They are best used in young and active patients who have a high risk of loosening and osteolysis in the mid to long term. Ceramic coatings provide an attractive alternative for biological fixation. In the near future, ceramic substitutes for bone grafts will probably be used in association with osteoinductive materials such as bone morphogenetic proteins or mesenchyal stem cells to accelerate bone formation further.

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References