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Definition and Classification

Ceramic is defined as “synthesized inorganic, solid, crystalline materials, excluding metals”. Ceramics used as biomaterials to fill up defects in tooth and bone, to fix bone graft, fracture or prosthesis to bone, and to replace diseased tissue, are called bioceramics. They must be highly biocompatible and antithrombogenic, and should not be toxic, allergenic, carcinogenic nor teratogenic. Bioceramics can be classified into three groups: (1) bioinert ceramics, (2) bioactive ceramics, and (3) bioresorbable ceramics. Bioinert ceramics have a high chemical stability *in vivo* as well as a high mechanical strength as a rule, and when they are implanted in living bone, they are incorporated into the bone tissue in accordance with the pattern of “contact osteogenesis”. On the other hand, bioactive ceramics have a character of osteoconduction and a capability of chemical bonding with living bone tissue. In other words, when bioactive ceramics are implanted in living bone, they are incorporated into the bone tissue in

accordance with the pattern of “bonding osteogenesis”. Mechanical strength of bioactive ceramics is generally lower than that of bioinert ceramics. Bioresorbable ceramics have a character of being gradually absorbed *in vivo* and replaced by bone in the bone tissue. The pattern of their incorporation into the bone tissue is considered similar to bonding osteogenesis, although the interface between bioresorbable ceramics and bone is not stable as that observed with bioactive ceramics.

Bioinert Ceramics

In 1969, Benson [1] predicted that carbon ceramic will be brought into clinical application as a biomaterial in the near future, as it has an excellent biocompatibility, a high compressive strength, and a reasonable elastic modulus. When carbon fibers were used as an artificial ligament, however, it tended to undergo fragmentation. Recently such mechanically stronger carbons as low temperature isotropic carbon (LTI carbon) and carbon fiber reinforced carbon (CFRC) have been developed, but their clinical application has not yet been brought to realization.

Bioinert ceramics such as alumina ceramic (Al_2O_3) and zirconia ceramic (ZrO_2) have a higher compressive and bending strength and better biocompatibility than stainless steel (SUS 316 L) or Co-Cr alloy. Alumina ceramic

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particularly, therefore, had been attempted to use as osteosynthetic devices (alumina monocrystal) or to fabricate bone and joint prostheses (alumina monocrystal + polycrystal) in 1980s [2]. Recently, however, due to their brittleness and too high elastic modulus as compared to those of human bone, they are used very little for above purposes. On the other hand, it has been known that a ball made of alumina or zirconia exhibits a wear resistant character when its surface is polished to an average surface roughness of $0.02\ \mu\text{m}$. At present, therefore, clinical application of alumina and zirconia is almost solely limited to the bearing surface of joint prosthesis.

It is well known that one of important factors causing loosening of joint prosthesis is the periprosthetic osteolysis which is due mainly to excessive macrophage activities against wear debris, particularly of polyethylene (PE), around the prosthesis. Even after the introduction of highly cross-linked polyethylene (XLPE) which is much wear resistant, there have been various attempts to reduce the number of wear debris by changing the bearing surface of prosthesis from PE-on-metal combination to metal-on-metal, ceramic-on-ceramic, or PE-on-ceramic combination. In 1970, Boutin [3] started to use an alumina-on-alumina combination for the bearing surface of hip prosthesis. According to Sedel [4], alumina ceramic used for the hip prosthesis between 1970 and 1979 had a mean crystal grain size of $7\ \mu\text{m}$ and the linear wear of its bearing surface was 5 to $9\ \mu\text{m}$ per year. While alumina ceramic currently used has a grain size of $2\ \mu\text{m}$ that means less brittle than the old type, and the linear wear is in the order of $3\ \mu\text{m}$ per year. Sedel further described that the overall wear of the currently used alumina-on-alumina hip prosthesis, calculated by the weight of debris generated, was approximately 1,000 times less than for metal-on-polyethylene and 40 times less than for metal-on-metal joint, if all requirements for alumina quality, sphericity, circularity, and clearance of the bearing components are met. Therefore, in spite of the fact that the alumina-on-alumina hip prosthesis used in 1970s did not show significantly better 10–15 year results as

compared to those of Charnley hip prosthesis, the current alumina-on-alumina hip prosthesis is expected to bring about much better long-term results and less breakage than those used in 1970s.

On the other hand, it is a well known rule for hip prosthesis that the smaller the head size the less the volumetric wear of the bearing surface. This rule has been much accounted of the PE-on-ceramic hip prosthesis in attempts to reduce the volumetric wear of polyethylene. The diameter of most alumina femoral head of hip prosthesis has been limited to 26–32 mm even with new alumina ceramic, because it exhibits only moderate bending strength and toughness. If it is attempted to reduce the head size to 22 mm with absolute safety against breakage of the component, zirconia ceramic is naturally taken into consideration as a constituent material, as it has an advantage over alumina of higher bending, compressive, and impact strength, higher fracture toughness, and lower elastic modulus. The reasons why zirconia ceramic had not been brought into clinical application until recently were that the zirconia synthesized in 1980s was abnormally radioactive [5] and tended to biodegrade *in vivo*. Modern technology, however, made it possible to synthesize a new zirconia which is not abnormally radioactive and is stable *in vivo*. This has been accomplished mainly by developing a refining technique to obtain pure zirconium from a raw ore and by adding chemical stabilizers such as yttrium oxide or cerium oxide during the sintering process of zirconia. The estimated amount of radioactivity exhibited from the zirconia, prepared by us in Kobe Steel Company since 1993, was $1.152\ \mu\text{R}$, while normal background of radioactivity amounts to about 100 mR [6]. Thus, the radioactivity of new zirconia is considered negligible.

Concerning crystallographical stability, alumina ceramic (usually α -alumina) is entirely consisted of hexagonal crystals and hence it is chemically very stable *in vivo*. On the other hand, zirconia ceramic is usually consisted of three crystallographical phases; cubic, tetragonal, and monoclinic, and transformation of the phase

takes place under various conditions such as change of temperature, mechanical stress, and humidity. The phase transformation often results in self destruction of the ceramic. Until 1980s, such a crystallographical unstableness was one of reasons why zirconia had not been used as a constituent material for the bearing component of joint prosthesis in which high stress concentration may be created on the ceramic surface by repeated loading under wet condition. In 1990s, however, new sintering methods have been introduced to prepare crystallographically stable zirconia ceramics, by adding such chemical stabilizers as Y_2O_3 , CeO_2 , and MgO in the sintering process. They are called partially stabilized zirconia (PSZ). For example, to prepare a zirconia femoral head of 22 mm in outer diameter, zirconia powder with a grain size of less than 1 μm is mixed with chemical stabilizers (Table 3.1), and is moulded into a ball by rubber pressing at room temperature. The ball is then sintered for 2 h at 1,500 °C. The sintered zirconia ball undergoes machining to shape a precise spherical ball with an outer diameter of 22 mm and a tapering pit (Fig. 3.1). The ball is finally polished to obtain an average surface roughness of less than 0.02 μm . Thus made zirconia femoral head is consisted mainly of tetragonal phase and 1–2 % of monoclinic and cubic phase.

Mechanical properties of PSZs are compared with those of new alumina ceramic with grain size of less than 2 μm in Table 3.2. PSZs have significantly higher bending strength, compressive strength, fracture toughness, and impact strength, but have a lower Vickers hardness and elastic modulus than the alumina ceramic, although they are slightly different depending on the grain size and kind of chemical

stabilizers used. Among them, yttrium oxide PSZ (Y-PSZ) has the highest bending strength and fracture toughness followed by cerium oxide PSZ (Ce-PSZ). Breaking tests for Y-PSZ and alumina femoral heads, 22 mm in outer diameter, were performed by static loading over a polyethylene liner which was set against the ceramic head. The alumina heads were broken by loads of 2,400–3,400 kg (average 2,800 kg), while Y-PSZ heads were broken by loads of 2,770–4,480 kg (average 3,700 kg). Thus, Y-PSZ head was significantly stronger than the alumina head against breakage [7]. Fatigue test was performed on eight Y-PSZ femoral heads on a hip simulator in physiological saline at 37 °C, by applying 10^7 cycles of repeated loading with 450 kg. This loading is considered to correspond approximately to 20 years of a person walking. After the test, no breakage was observed in all the eight Y-PSZ heads.

Wear tests for the polyethylene liner against the Y-PSZ, alumina and stainless steel head, all 22 mm in outer diameter, were performed using a hip simulator in physiological saline at 37 °C by applying a load of 450 kg at 1 Hz. After 5×10^5 cycles of loading, the polyethylene liner against the stainless steel head showed significant wear, while those against the Y-PSZ head and alumina head did not show any measurable wear, even after 2×10^6 loading [7].

Thus, alumina is chemically more stable than PSZ *in vivo*, while PSZ is mechanically stronger than alumina, and both of them exhibit much better wear-resistant character comparing the stainless steel or Co-Cr alloy as assessed in a form of bearing components of hip prosthesis. For these reasons, alumina is used to fabricate a ceramic-on-ceramic hip prosthesis where head size is not a key issue, while PSZ is used to fabricate a PE-on-ceramic hip prosthesis where the head size must be made reasonably small. One of reasons why the zirconia-on-zirconia or alumina-on-zirconia hip prosthesis is not yet brought to the market is that, even with the PSZ, its crystallographical stability *in vivo* in a long term has not been confirmed.

Table 3.1 Chemical composition of zirconia ceramic

	Weight %
$SiO_2 + M_2O$ ($M_2O: Na_2O, K_2O$ etc)	0.1
Fe_2O_3	0.1
Al_2O_3	0.5
Y_2O_3	4.8 ± 0.7
ZrO_2	Remainder

On the other hand, recently a combined ceramic (Zirconia 20 % and Alumina 80 %) is brought into clinical use as a XLPE-on-Ceramic hip prosthesis. This combination of Zirconia and

Alumina is to aim at covering the weak points each other. It is said, however, when a proportion of Zirconia exceeds 6 %, effect of phase transformation can not be neglected in vivo. Therefore,

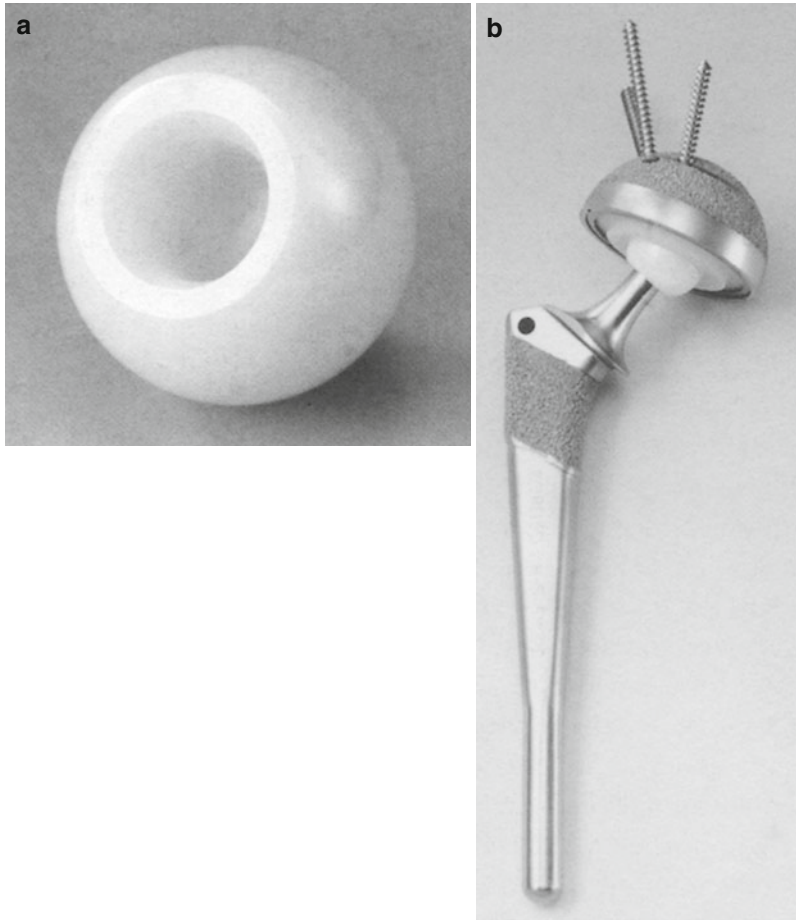


Fig. 3.1 (a) A zirconia made femoral head with an outer diameter of 22 mm. (b) A cementless hip prosthesis made of titanium alloy with a combination of zirconia head and polyethylene socket for the bearing component

Table 3.2 Mechanical properties of bioinert ceramics

		Zirconia	Alumina
Bending strength	(kgf/mm ²)	170	>40.8
Compressive strength	(kgf/mm ²)	500	408
Fracture toughness	(MPa,m ^{1/2})	5.2	3.4
Impact strength	(kg/mm ² or kJ/m ²)	14	4
Vickers hardness	(HV kg/mm ²)	1270	2300
Elastic modulus	(kgf/mm ²)	20500	>38800
Density	(g/cm ³)	6.05	>3.9
Crystal size	(μm)	0.2	<7

clinical follow-up longer than 10 years is required to confirm the long term results in this combination as well.

A new technology developed by Smith and Nephew Co. in 1998 using Zirconium-Niobium alloy made it possible to solve the problem of phase transformation of zirconia ceramic in vivo. When a high temperature is applied on Zr-Nb alloy, its surface transforms to a monoclinic zirconia layer in about 5 μm thick. The layer is called Oxinium. Thus made surface monoclinic zirconia layer (Oxinium) is a gradient material made from Zr-Nb alloy. When a femoral head of hip prosthesis is made by the use of the above technique, its surface is not affected by phase transformation in vivo as it is made of monoclinic zirconia, and, in addition, the femoral head is not broken as it is made of Zr-Nb alloy, a metal. For these reasons, Oxinium-on Oxinium hip prosthesis is considered reasonable theoretically.

Bioactive Ceramics

Bioactive ceramics include glasses, glass-ceramics, and ceramics that elicit a specific biological response at the interface between the material and the bone tissue which results in the formation of a bond between them. The first evidence of direct bone bonding to a glass implant was discovered by Hench et al. in 1970 [8]. Since then, some other glasses, glass-ceramics, and ceramics had been proved to have a bone bonding capability. Among them, Bioglass®, apatite- and wollastonite-containing glass-ceramic (AW-GC) and synthetic hydroxyapatite (HA) are representative materials currently used for clinical application.

In 1970, Hench et al. [8] synthesized a bioactive glass by a chemical composition of SiO_2 45, CaO 24.5, P_2O_5 6, Na_2O 24.5 (wt%). This glass is called 45S5 Bioglass® and known to exhibit the strongest bioactivity among hitherto developed bioactive ceramics. Wilson et al. [9] proved that when the implant-tissue interface was immobilized, collagen fibers of the soft tissue became embedded and bonded within the growing silica-rich and hydroxy-carbonate apatite layer on the 45S5 Bioglass®. Such soft-tissue bonding has never been observed with other bioactive ceramics or glass-ceramics. However, as Bioglass® is mechanically much weaker than the human cortical bone, it cannot be used as a weight bearing bone prosthesis. In stead, it has been used as a bone void filler in a form of granule, coating material on metallic prostheses, and to fabricate a middle ear prosthesis.

Aoki et al. [10] in 1966 and Jarcho et al. [11] in 1976 separately developed a process for producing dense hydroxyapatite implants with considerably high mechanical strength. Synthetic hydroxyapatite ($\text{Ca}_{10}(\text{PO}_4)_6(\text{OH})_2$) has a capability of chemical bonding with the living bone tissue, but it takes much longer time than Bioglass® for bone bonding. Its mechanical property is shown in Table 3.3 in comparison with that of the natural bone and AW-GC. The bending strength of HA is lower than that of the natural cortical bone, and hence HA cannot be used to fabricate a weight bearing bone prosthesis with absolute safety against breakage *in vivo*. It has been used as a bone void filler in a form of granule with various particle size (Fig. 3.2), coating material on metallic prostheses, and to fabricate an iliac crest prosthesis and a laminoplasty spacer in

Table 3.3 Mechanical property of natural bone and bioactive ceramics

	Bending strength (Mpa)	Compressive strength (Mpa)	Elastic modulus (Gpa)
Natural bone	30–190	90–230	3.8–17
Synthesized Hydroxyapatite	110–170	500–900	35–120
A-W Glass-Ceramic	220	1000	120

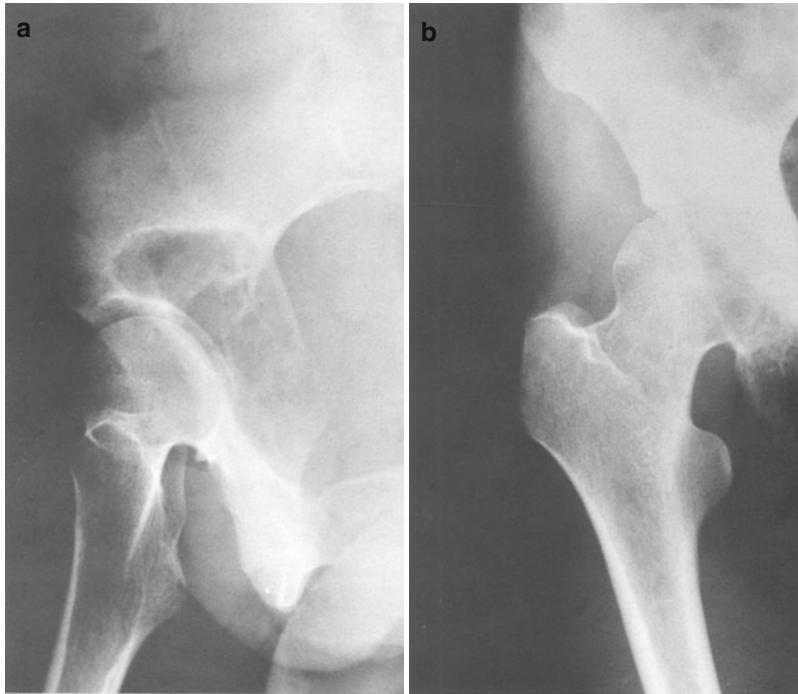


Fig. 3.2 (a) A giant cell tumor developed in the right ilium and ischium of a 27 year old female. The tumor was excised and the remaining large bone defect was filled with a mixture of autogenous bone chips, HA granules

and fibrin glue. (b) 20 years postoperatively, HA has been well incorporated into the surrounding bone and the patient has no symptoms

which high mechanical strength is not required. In 1987, Geesink et al. [12] developed a HA coated hip prosthesis and reported an excellent 10 year clinical result in a large number of patient (Fig. 3.3). At present, HA is the bioactive ceramic most widely used for clinical application as a bone void filler and a coating material for hip prostheses which are employed in cementless hip replacement.

The HA coating over the prosthesis is usually performed by plasma splay coating, and thus made HA coated layer is consisted of more than 50 % of amorphous apatite. As the amorphous apatite is absorbed by osteoclasts in a few years, HA coating only cannot maintain the firm connection between the prosthesis and bone for longer than 20 years. For this reason, modern

cementless hip prosthesis has a HA coating over the porous Ti coating.

Aiming at producing a mechanically stronger bioactive material, Kokubo et al. [13] in 1982 developed apatite-and wollastonite-containing glass-ceramic (AW-GC or Cerabone AW®) by a chemical composition of SiO_2 34.0, CaO 44.7, P_2O_5 16.2, MgO 4.6, CaF_2 0.5(wt%). As shown in Table 3-3, AW-GC has a significantly greater bending and compressive strength than the human cortical bone and dense HA. Bioactivity was compared among Bioglass®, HA and AW-GC by implanting them into the living bone tissue and carrying out detaching tests in different postimplantation periods. It was demonstrated that the bone bonding occurred earliest with Bioglass®

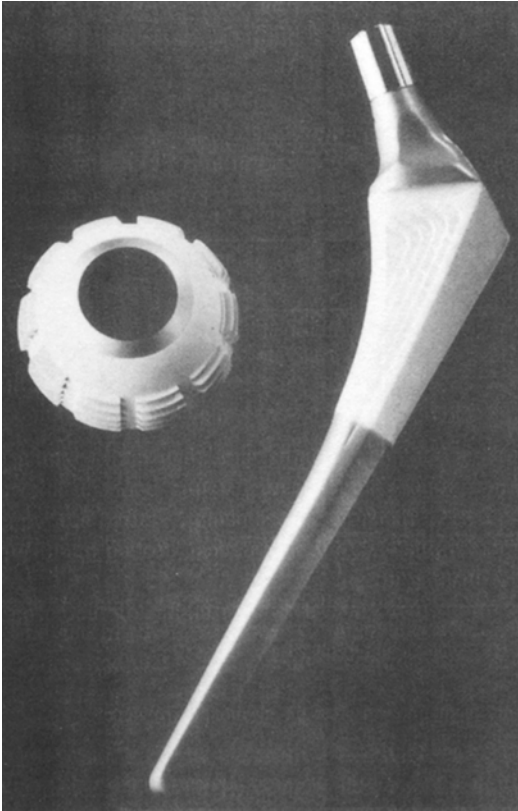


Fig. 3.3 HA coated hip prosthesis developed by Geesink et al.

followed by AW-GC and then HA. The essential mechanism of bone bonding for Bioglass® and AW-GC is considered similar, that is the formation of an apatite layer on the implant surface in the body environment. This surface apatite formation takes place by a chemical reaction of Ca^{2+} and HSiO_3^- ions dissolved from the implant surface. This apatite is called chemical apatite (Fig. 3.4). At the same time, on the cut surface of bone, an apatite layer accompanied by collagen fibers is formed by the activity of osteoblasts. This apatite layer is called biological apatite. Neo et al. [14] observed under transmission electron

microscope that the chemical apatite and the biological apatite were intermingled at the bone bonding interface (Fig. 3.5). The HA implant also showed the similar bone bonding morphology under transmission electron microscope, but HA took longer time than Bioglass® or AW-GC for bone bonding. This may presumably be due to the fact that HA is solely consisted of crystals, while others contain the glass phase which is dissolved faster than the crystal and dissolved HSiO_3^- ions might provide favorable sites for nucleation of the apatite [15].

Yamamuro et al. [16] replaced vertebral bodies of sheep with a vertebral prosthesis made of AW-GC and found that the prosthesis bonded directly to the adjacent vertebrae within about 1 year (Fig. 3.6). Then, by the use of AW-GC, various bone prostheses were fabricated such as iliac crest prosthesis, vertebral prosthesis, intervertebral spacer, and laminoplasty spacer (Fig. 3.7) [17]. The iliac crest spacer is used to substitute a bone defect remaining after harvesting a large bone graft from the iliac crest in various orthopaedic operations. The vertebral prosthesis is used to substitute for vertebral bodies suffering from benign and malignant tumors, compression fracture and burst fracture (Fig. 3.8). The intervertebral spacer is used for interbody fusion through either anterior or posterior approach (Fig. 3.9). The laminoplasty spacer is used to maintain bilateral laminae opened after surgical enlargement of the cervical spinal canal in degenerative spondylosis and ossification of the posterior longitudinal ligament (Fig. 3.10). AW-GC has also been used as a bone substitute in forms of either block (Fig. 3.11) or granule. When bioactive ceramic granule is used as a bone void filler together with fibrin glue, osteoconduction and bone bonding are accelerated. AW-GC is also widely used for bioactive coating of hip prosthesis. Its details are described in the Chapter of Ceramic Coating.

Fig. 3.4 Schematic representation of apatite formation on AW-GC

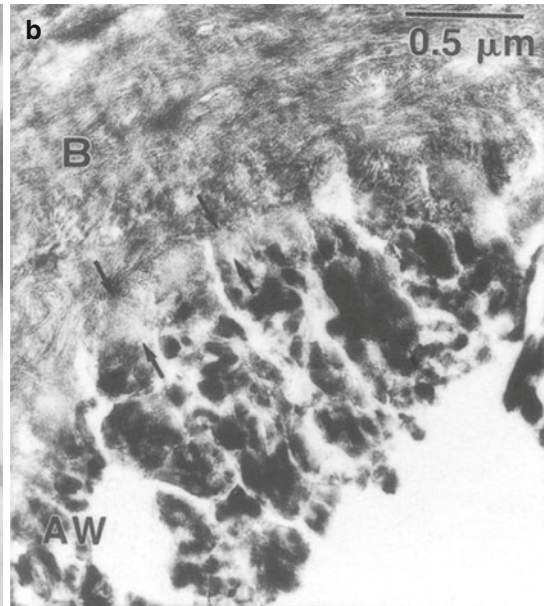
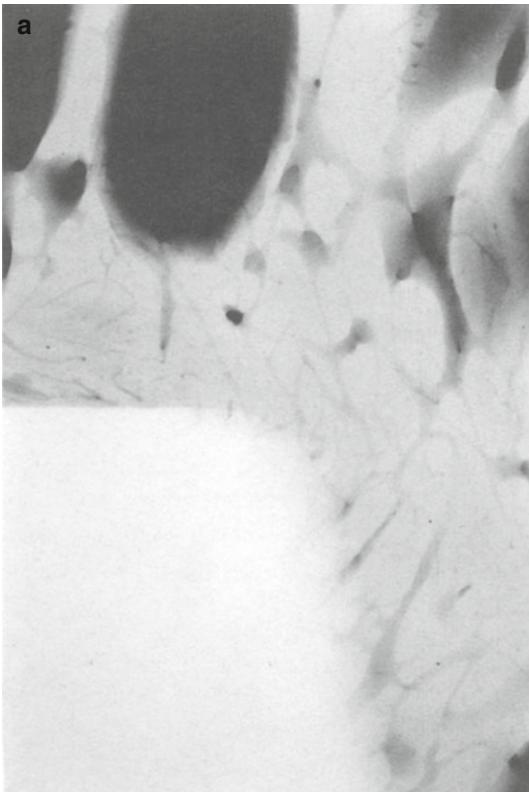
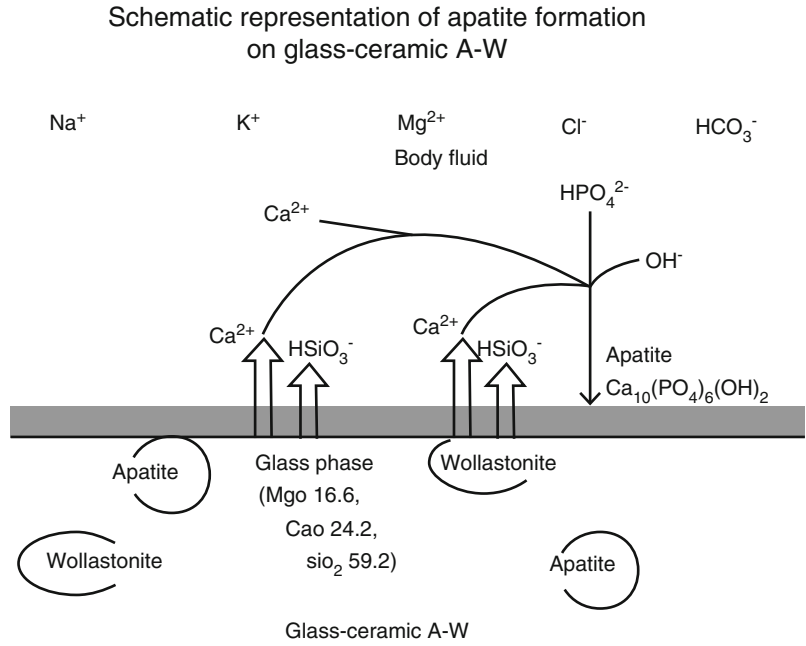


Fig. 3.5 (a) A contact micro-radiograph showing the bone bonding between AW-GC implant and newly formed bone. (b) A transmission electron micrograph showing the

bonding interface between AW-GC (AW) and bone B (arrows). AW AW-GC crystals, B bone tissue

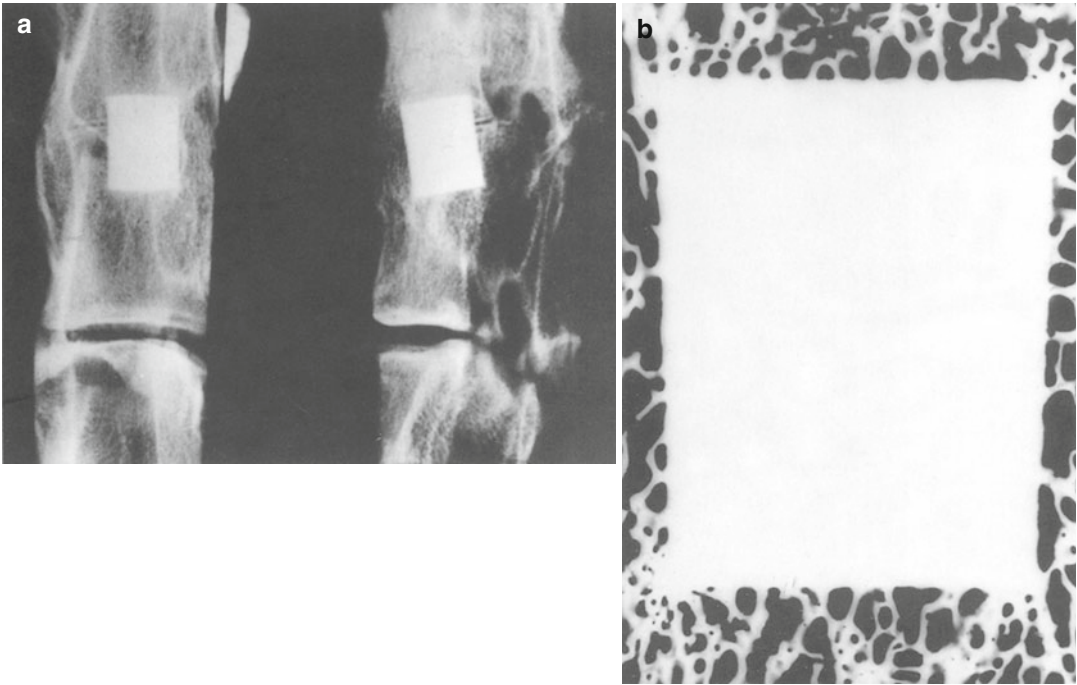


Fig. 3.6 (a) A X-ray picture showing an AW-GC implant used for interbody fusion of the lumbar vertebrae of a sheep. (b) A contact micro-radiograph demonstrating direct bonding between the AW-GC implant and bone trabeculae of the lumbar vertebrae, 1 year postimplantation

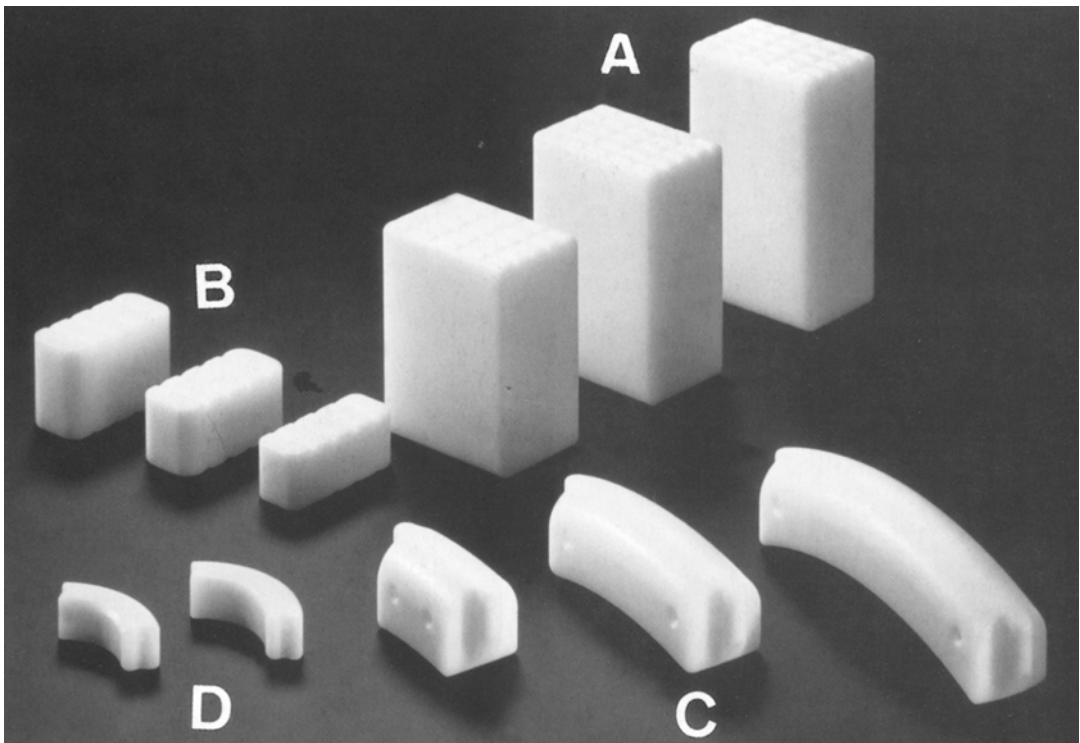


Fig. 3.7 Various bone prostheses. A: vertebral prosthesis, B:intervertebral spacer, C:iliac crest prosthesis, D:laminoplasty spacer

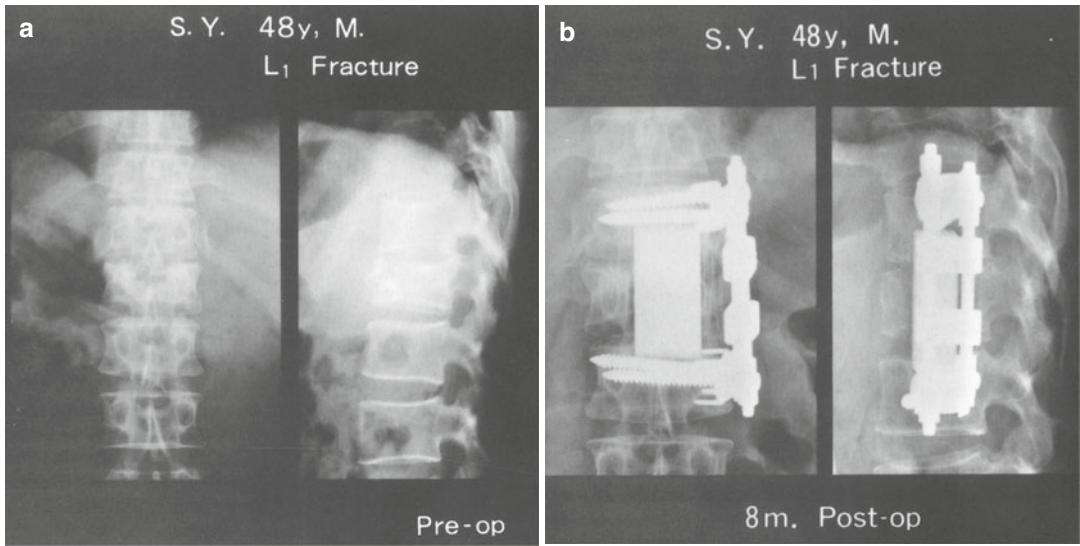


Fig. 3.8 (a) A x-ray picture demonstrating L₁ burst fracture associated with paraplegia developed in a 48 year old male. (b): A postoperative X-ray picture showing a vertebral prosthesis used for the reconstruction of the lumbar spine

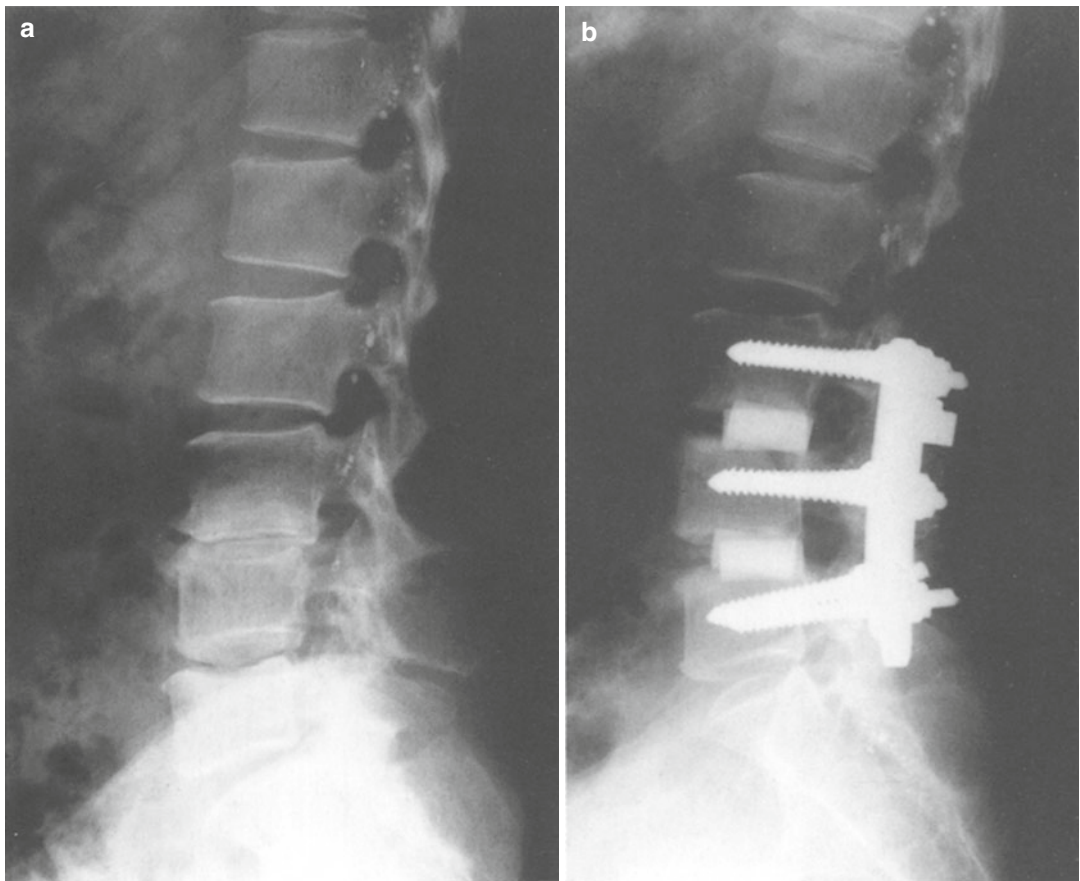


Fig. 3.9 (a) A case of multiple degenerative spondylosis of the lumbar spine developed in a 58 year old female. (b) A postoperative X-ray picture showing the results of postero-lateral interbody fusion using intervertebral spacers made of AW-GC

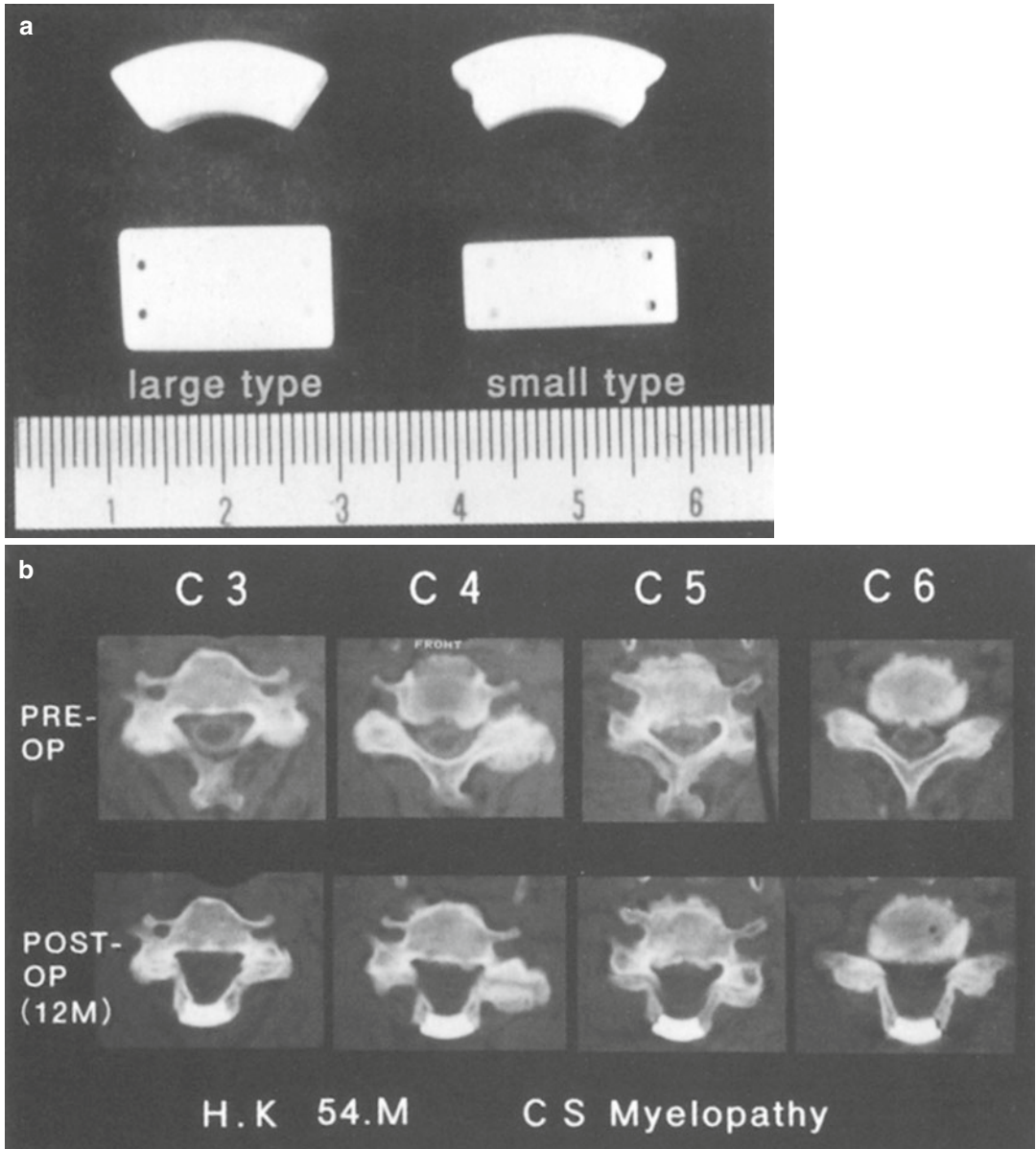


Fig. 3.10 (a) Laminoplasty spacers made of AW-GC. (b) CT images of a 54 year old male suffering from cervical myelopathy due to spondylosis (*upper row*). Enlargement and reconstruction of the spinal canal was performed by the use of laminoplasty spacers in four levels (*lower row*)

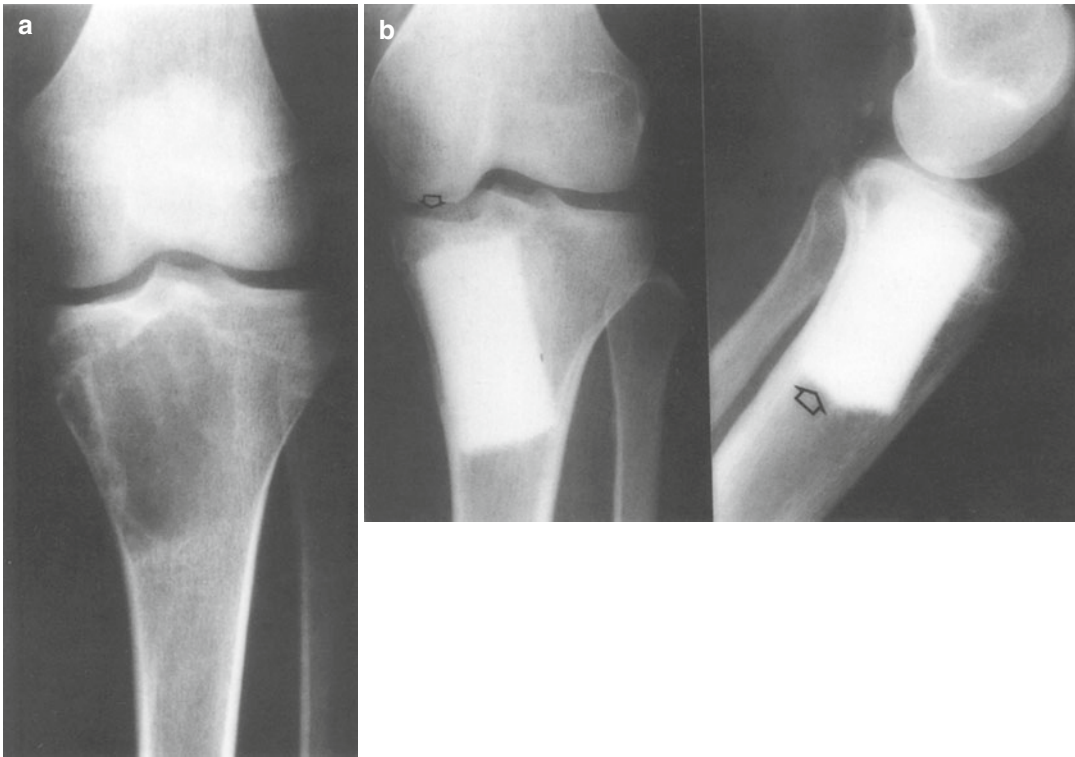


Fig. 3.11 (a) A giant cell tumor developed in the proximal tibia of a 15 year old female. (b) 10 years postimplantation of an AW-GC block that was used in combination of

autogenous bone chips to replace the tumor. No clinical symptoms postoperatively

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