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Porous bone implants

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Abstract

Porous conical implants made by sintering micro-spheres made from bioactive and bio-compatible glasses have been tested mechanically as well as in vivo by inserting the implant through the cortical bone into the bone marrow. The behaviour is compared to a reference implant made by sintering micro-spheres of metallic titanium. Due to capillary forces the implant pores were filled with bone marrow fluid when inserted. An extended bone ingrowth occurred in the cones of bioactive glass, in the titanium ones only in the cortical area. Further, the factors influencing the formation of a chemical bond between glass and living tissue are discussed.

Keywords: B. Porosity; D. Glass; Bone implants

1. Introduction

Due to the pioneering work of Professor Larry Hench since the 1960s the bioactivity of certain glasses is by now a well established fact [1,2]. Bioactivity is here defined as the ability of tissue to form a chemical bond to an artificial substrate, in the present case to a silicate glass. However, the fine details of the bond formation are still only fragmentarily understood. What is known is that the formation is preceded by a partial dissolution of the glass surface. Further, calcium phosphate is found well within the gel layer [3], but the origin of the calcium and phosphate ions is not known. It has been suggested that they originate from the body fluid [4,5] but, because bioactive glasses are often phase separated, it is equally possible that the deposition is a consequence of a slow kinetic reaction between ions from different phases. Nor is it known whether the phosphate found in the gel is chemically bonded to the gel or just precipitated. The fact that a blocking of the non-bridging oxide ions of the gel prevents apatite formation speaks in favour of the former mechanism, but it may equally well be that calcium silicate bonds are weaker than the aluminium ones [4].

The increased interest in the possible use of glass in implants is due to the fact that the composition can be varied considerably without the glass loosing its bioactivity. In an in vivo study of glasses in the system Na_2O–CaO–P_2O_5–B_2O_3–Al_2O_3–SiO_2, the response in rabbit tibia could be classified on a numerical scale from 1 to 6, 5 and 6 representing good bone bonding [6]. Recently, Brink [7] has shown that the effect of K_2O is roughly the same as that of Na_2O.

The release of calcium from the dissolving glass may also interfere with the phosphate equilibrium over the vesicle wall in the body fluid. A minor increase in calcium concentration may cause precipitation of apatite in the cell. This then leads to cell division [8]. Further it has been shown that close to a particle of a bioactive glass an else inactive silica gel develops an apatite layer on its surface [9]. This indicates that calcium ions dissolving from the bioactive glass lead to a local super saturation and calcium phosphate precipitation. Thus the silica gel as well as calcium ions seem to play an important role in the bonding of glass to bone.

Glass is, however, a brittle material. A more extensive use of bioactive glasses in prosthetic applications has therefore not been possible. In the present paper the glass is used only to enhance the penetration of bone into the implant. The strength is achieved by a rapid formation of new bone along the glass surface. The osteoconductive area is increased by making the implant porous by sintering micro-spheres of a bioactive glass.

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2. Experimental

Two glasses were chosen for the experiments, one was previously defined as bioactive [7], the other as biocompatible [10]. The composition of the two glasses are given in Table 1. The glasses were made from analytical grade raw materials and melted twice in a platinum crucible at 1360°C. The annealed glasses were crushed and spherulized by feeding into a acetylene–oxygen flame as described by Pitkänen et al. [11]. The microspheres were then classified and those within the diameter range 200–300 μm used in this study. These spheres were sintered in a graphite mould at 750°C for 3 min to obtain conical implants with a diameter of about 3 mm. This procedure gave reasonably rigid implants with a porosity of about 40%. The porous implants were cut to a length of about 4 mm, sterilized in ethanol and either immersed in SBF or implanted in rabbit femur.

As reference, metallic titanium micro-spheres of same size were sintered to cones with same porosity. This was done in a vacuum furnace at 1500°C for 2.5 h.

Eight adult ELCO rabbits, NL, were used in evaluating the in vivo behaviour of the porous cones. A set of three different cones were implanted in both femurs of each rabbit. Implantation was done in random circulating order. The conical samples were press-fit to the conical holes drilled in the bone. Due to capillary forces the porous bioactive bodies were on implantation immediately filled with blood containing stem cells from the bone marrow.

3. Results

After 6 weeks the rabbits were killed and the legs X-rayed, (Fig. 1). Next, half of the samples were used for determining the shear strength during push-out, the other half prepared for microscopy as described elsewhere [6]. Each sample was axially sawn into two halves, one for SEM investigation, the other for histomorphometry.

Table 1
Glass compositions (wt.%)

Glass	Na_2O	K_2O	MgO	CaO	B_2O_3	P_2O_5	SiO_2
Bioactive	6	12	5	20		4	53
Biocompatible	25.5			11	1.3	2.5	59.7

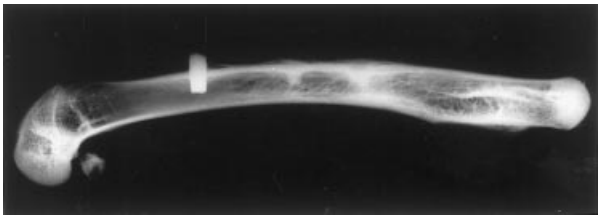

Fig. 1. X-ray picture of rabbit femur after 6 weeks implantation. Cones from left: titanium, bioactive, biocompatible.

In testing the shear strength between implant and cortical bone, the breakage occurred in the bone fingers growing into the implants. The measured strength was 20–30 MPa and there was no significant difference between the glass and titanium cones. However, because of the large contact area of the porous implants, the strength was up to 50% greater than reported by Andersson et al. [12] for smooth surfaced cones of identical size and shape.

The microscopical evaluation revealed an interesting difference between cones of bioactive glass spheres and of titanium. Within the 6 weeks the stem cells in the glass cones had transferred to bone all throughout the cone, while in the titanium implant this was true only in the region surrounded by cortical bone (Fig. 2). The surface of the glass spheres consisted of two layers, a silica gel layer next to the glass and an apatite layer on top of the gel. The contact between sphere and bone was also close, while in the titanium implants, though the osteoconductivity was good, a narrow gap could often be found next to the metal surface.

4. Discussion

A comparison between the bioactive glass cones and the titanium cones in Fig. 2 indicates that in the latter the bone formation takes place only in the cortex region, possibly through ingrowth of bone. On the other hand, in the bioactive glass the behaviour is less straight forward. As seen from Fig. 2, bone formation has taken place also in the region surrounded by marrow. Several interpretations to this phenomenon are possible.

a. The glass surface is more osteoconductive than titanium. It is, however, difficult to explain why in titanium the bone growth extends 2 mm in the direction of cortex, but not at all into the part sticking into the marrow, as is the case with the porous glass implant.

b. The glass surface is osteopromotive. In Fig. 1, it is seen that a bridge of bone grows from the tip of a bioactive cone to the opposite cortex. An osteopromotive property of low silica glasses has been proposed by Hench [2]. The question rises, however, why do only low silica glasses give a bioactive gel, while the bioactivity disappears if the silica content exceeds some 58 wt.%. From the point of glass technology, the only difference is in the solubility. On the other hand a more soluble glass gives more cations to the surroundings. If the materials exchange is mainly governed by diffusion, as one would expect inside a body with pore sizes hardly exceeding 0.3 mm, then the role of these other ions must not be neglected. In an attempt to monitor this, a

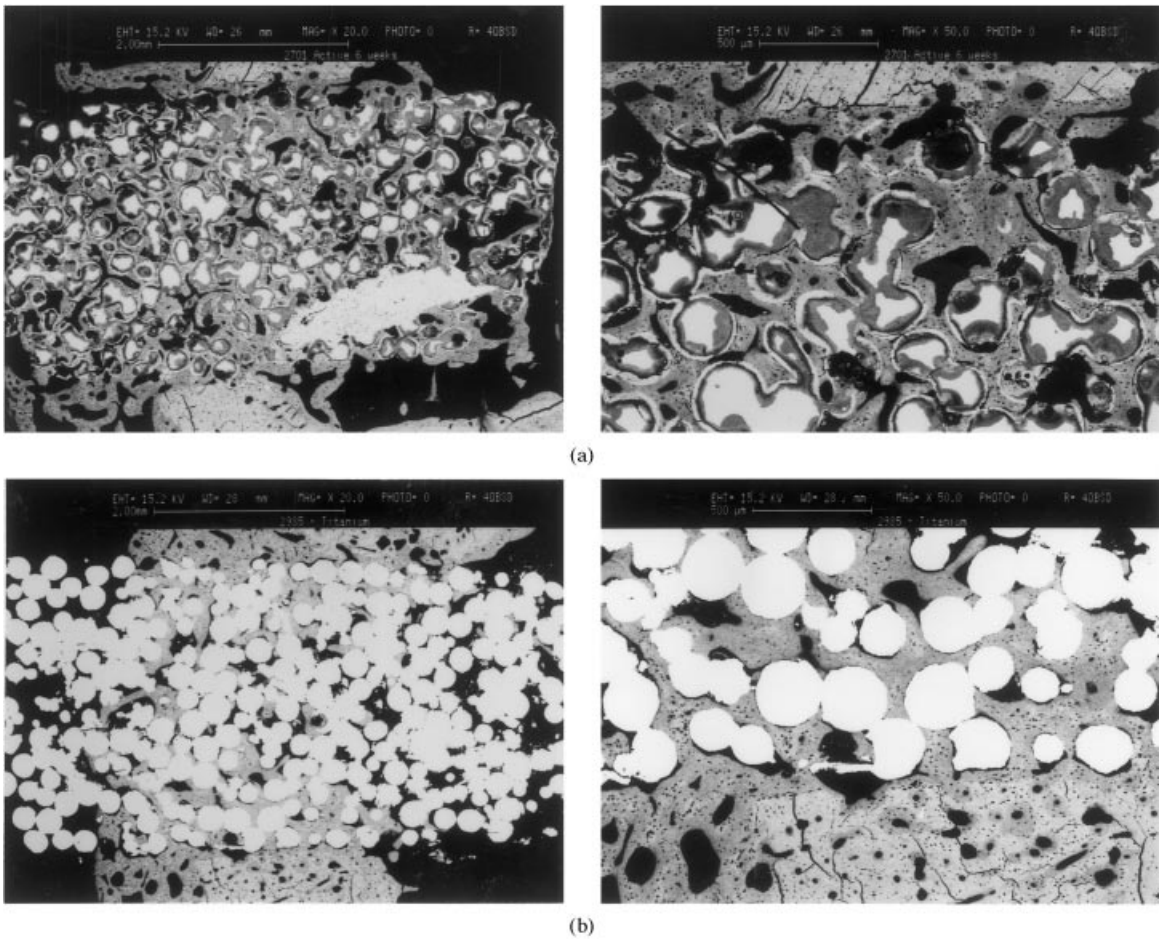

Fig. 2. (a) Bioactive glass cone after 6 weeks in rabbit femur: bar, left = 2 mm, right = 0.5 mm; (b) titanium cone after 6 weeks in rabbit femur: bar, left = 2 mm, right = 0.5 mm.

glass cone was prepared form the biocompatible glass. In the cone, however, one particle was replaced by a bioactive one, Fig. 3. The cone was immersed into SBF for 14 days. Almost all particles had developed a thin gel layer on their surfaces. The bioactive one, however, showed extensive apatite formation outside the gel. Further, apatite was also found in the gel of the next neighbour compatible particles. Thus this gel had captured some ions from the SBF in contact with the active particle. As the calcium content of the glass is much higher than the phosphate content, it is most probable that dissolved Ca^{2+} has caused the solubility product of calcium phosphate to be exceeded in the SBF. The deposition then takes place where enough surface energy is available, i.e. at the gel.

c. The gel formed on a low silica glass is different from one formed on a glass with higher silica content. Designating by Q^n a silica tetrahedron sharing corners with n different tetrahedra, it is possible that the former gel contains a higher proportion of Q^1 tetrahedra than the latter. A gel formed on a bioactive glass would then be much more flexible and able to accommodate apatite, either through a chemical bond [4] or just by providing nucleation sites for apatite crystallization. In favour of the former interpretation is the fact that the bioactivity fades if the alumina content in the glass exceeds some 2%. The complexation ability of alumina with silica is strong and too many Q^1 are occupied.

d. A fourth possible explanation to the ingrowth of bone is that dissolved silicic acid exhibits growth factor properties. Hench has recently discussed this possibility [2]. His argumentation was based on an observation by Keeting et al. [13] that soluble silica seems to have a metabolic role. A support for this was found by Ringbom-Anderson et al. [14] in a systematic investigation of the growth of human osteosarcoma cells in nutrition media containing dissolved silicic acid. The cell growth increased up to 15% with increasing silicic acid content up to 50 ppm. Higher contents, however, lead to a decrease in cell growth and when the concentration exceeded 100 ppm silicic acid, it prevented all cell growth, Fig. 4.

Thus there seems to be several possible explanations to the behaviour of glass in a bone implant. At the present stage is it not possible to make a final conclusion for the

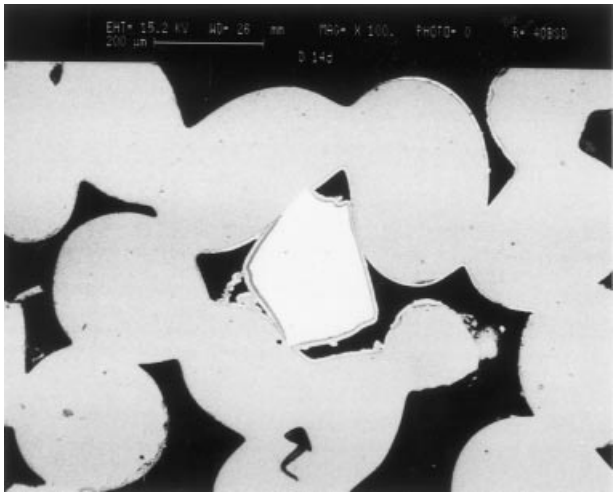

Fig. 3. A bioactive glass particle (white) in a porous body of sintered biocompatible micro-spheres (gray). The dark layer around the light bioactive particle is silica gel, the white layer on top of this as well as on the neighbouring compatible glasses is apatite [9]. Bar = 0.2 mm.

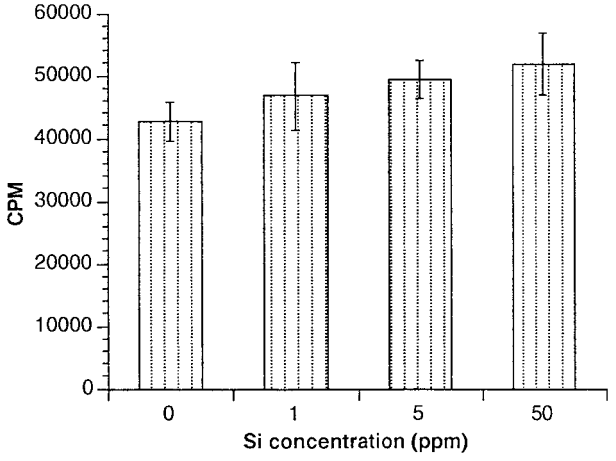

Fig. 4. The effect of dissolved silicic acid on proliferation of human U-2 OS osteosarcoma cells at pH 7.4. 100 ppm Si was toxic to cells [14].

behaviour in such a complex environment as offered by the human body. Any or all the suggested mechanisms may participate in the formation of a chemical bond between bone and glass.

5. Conclusions

Porous implants made from titanium and bioactive glasses show a high osteoconductivity. The question whether glass also is osteopromotive remains, however, still open. By the action of body fluid, bioactive glass particles in a porous implant dissolve and are consecutively replaced by bone. The ultimate state is achieved when all glass is replaced.

Cell differentiation takes place at the surface of the bioactive glass. The structure of the gel seems to be significant. The lower the degree of polymerization, the more active the gel. It is suggested that in biocompatible glasses the dominant silica configuration is Q^2, while bioactivity seems to require primarily Q^1 configurations. There are indications that at least the Q^0 configuration stimulates cell growth.

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