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Manufacture of Macroporous Calcium Hydroxyapatite Bioceramics

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Abstract

Trabecular bones of almost all vertebrate organisms basically consist of macroporous (55–70% interconnected porosity) bone mineral, i.e. calcium hydroxyapatite (HA: $Ca_{10}(PO_4)_6(OH)_2$). The macroporosity observed in the trabecular bones then allows the ingrowth of the soft tissues and organic cells into the bone matrix. Sub-micron, chemically uniform, and high phase-purity HA powders produced in our laboratory were mixed, under vigorous ultrasonification, with methyl cellulose of appropriate amounts in the form of an aqueous slurry of proper viscosity and thickness. The ceramic cakes produced in this way were then slowly dried in an oven in the temperature range of 50–90 °C. Dried cakes of porous HA were physically cut into various prismatic shapes. These parts were then slowly heated in an air atmosphere to the optimum sintering temperature of 1250 °C. The HA bioceramic parts obtained by this novel 'foaming technique' were found to have tractable and controllable interconnected porosity in the range of 60–90%, with typical pore sizes ranging from 100–250 microns. Sample characterization was mainly achieved by scanning electron microscopy (SEM) studies and three-point bending tests. © *1999 Elsevier Science Ltd. All rights reserved.*

Keywords: apatite, bioceramics, calcium hydroxyapatite, foams, porosity.

1 Introduction

Ceramics used for the repair and reconstruction of diseased or damaged parts of human body are termed bioceramics. With the growing demands of bioactive materials for orthopaedic as well as maxillofacial surgery, the utilization of calcium hydroxyapatite (HA, with Ca/P = 1·667) and tricalcium phosphate (TCP, with Ca/P = 1·5) as fillers, spacers, and bone graft substitutes has received great attention mainly during the past two decades, primarily because of their biocompatibility, bioactivity, and osteoconduction characteristics with respect to host tissue.[1–3]

In recent years, attention was particularly placed on the fabrication of bioceramics with "porous" configuration because the porous network allows the tissue to infiltrate, which further enhances the implant-tissue attachment.[4–14] In a porous form, hydroxyapatite ceramics can be colonized by bone tissue with the same characteristics as peri-implanted tissues.[15] For colonization of the pores to take place, they must be larger than 50–100 μm[13] or even 250–300 μm according to some researchers.[16–18]

To impart porosity to a ceramic body, various methods were known to be used. They were based mainly on admixing a foreign combustible organic material that burned away during firing, leaving free spaces and voids in the resulting body. These organic powders (such as, polvinyl butyral[9] or amino-acid derivatives[11,12]) in the production of macroporous bioceramics were selected to have lower burn-out temperatures than the ceramic sintering temperature. Such methods, however, were regarded to be unsatisfactory because they did not insure a uniform distribution of pores in the ceramic body, especially when the organic powders reside in discrete pockets after mixing with the ceramic powders.

Ryshkewitch[19] was the first researcher using a ceramic slip of oxide powders and water solution of 0·2% polvinyl alcohol, and mixed this slip with a 4% solution of hydrogen peroxide (H_2O_2) to produce porous alumina and zirconia parts. In this study, porosities (by volume) in the range of 5–60% in the ceramic bodies were obtained after firing at a maximum temperature of 1850 °C. Ryshkewitch's technique was later successfully used by Klein *et al.*[10,20] to produce porous calcium hydroxyapatite (HA) bioceramics.

The present study, to our knowledge, becomes the first attempt which employs the mixing of aqueous methyl cellulose solutions with sub-micron hydroxyapatite powders, under ultrasonic irradiation, to produce porous ceramic cakes, and upon drying and sintering the macroporous HA bioceramic parts.

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2 Experimental Procedure

The hydroxyapatite powders of this study were produced in our laboratory,[21,22] with an average particle size of 0·6 μm, were then used[23] to prepare the HA slurries containing aqueous solutions of methyl cellulose to form sponge-like bioceramic cakes and bodies of differing porosity simulating those of human bones. Solutions containing the HA powders and methyl cellulose were treated with an ultrasonic disruptor (Misonix, Inc., Model: XL2015, NY, USA) to homogenize and degas the slurries.[24] Polymeric slurries were slowly dried in an oven in the temperature range of 50–90 °C. Thus obtained, green cakes were then physically cut into any desired shape, and finally sintered at 1250 °C for 3 h in a stagnant air atmosphere. The green, porous HA parts (typically of cubic or rectangular shapes with the dimensions of 1–2×1–2 ×1–2 cm) were first heated to 250 °C at the rate of 0·5 °C min^{-1}. The total burnout of organic material was achieved during this stage. The parts were then heated to the peak sintering temperature at the rate of 3 °C min^{-1}, and cooled down to RT again at the same rate.

For density (and porosity) measurements, the sintered samples were initially subjected to ultrasonic washing in distilled water for a few minutes. After drying the samples in a stagnant air oven at 90 °C, their dry weights were recorded. The samples were then boiled in distilled water for about 3 h, and allowed to cool in water for 24 h. Wet weight in air and wet weight suspended in water were determined by using an analytical balance (Precisa, 300S, Switzerland). Water absorption, bulk density, apparent porosity and volume fraction of porosity were calculated in our samples by using the below formula;

$$\text{Water absorption} = (W - D)/D \quad (1)$$

$$\text{Bulk density} = D/(W - S) \quad (2)$$

$$\text{Apparent porosity} = (W - D)/(W - S) \quad (3)$$

$$\text{Volume fraction of porosity} = 1 - (\text{Bulk density}/\text{theoretical density}) \quad (4)$$

where W is the wet weight, D is the dry weight and S is the wet weight suspended in water.

In the mercury intrusion porosimetry experiments, which we used to determine the porosities of our samples, liquid mercury was injected into the sintered rectangular pieces at various pressures and total porosity values were determined by using the below formula;

$$V_p = (\text{injected mercury volume at 50 atm} - \text{correction factor at 50 atm}) \times 1{\cdot}02 \quad (5)$$

Total porosity $= (V_p/V_b) \times 100$,

where V_p is the pore volume and V_b is the bulk volume.

Three-point fracture strengths of sintered and porous samples were calculated by using the below formula:

$$S = (3.P.L)/(2.b.d^2) \quad (6)$$

where P is the fracture load (kg), L is the span length (mm), b is the width of the sample (mm) and d is the thickness of the sample (mm).

Scanning electron microscopy (SEM, Jeol Corp., Model: JSM-6400, Tokyo, Japan) was used for the visual characterization of the pore size and morphology distribution in the macroporous HA bioceramic samples.

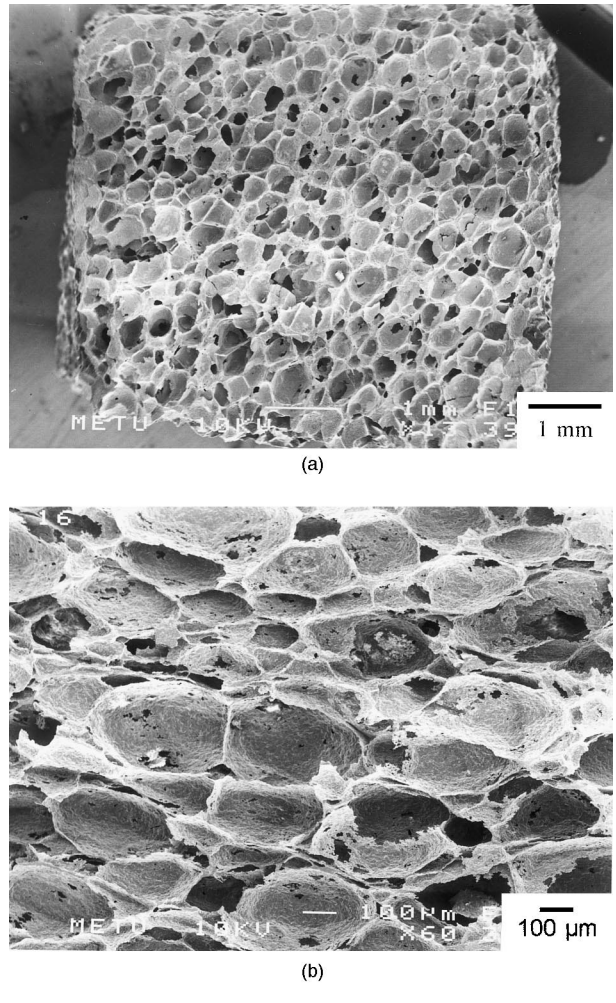

Fig. 1. SEM micrographs of HA bioceramics of 60% porosity.

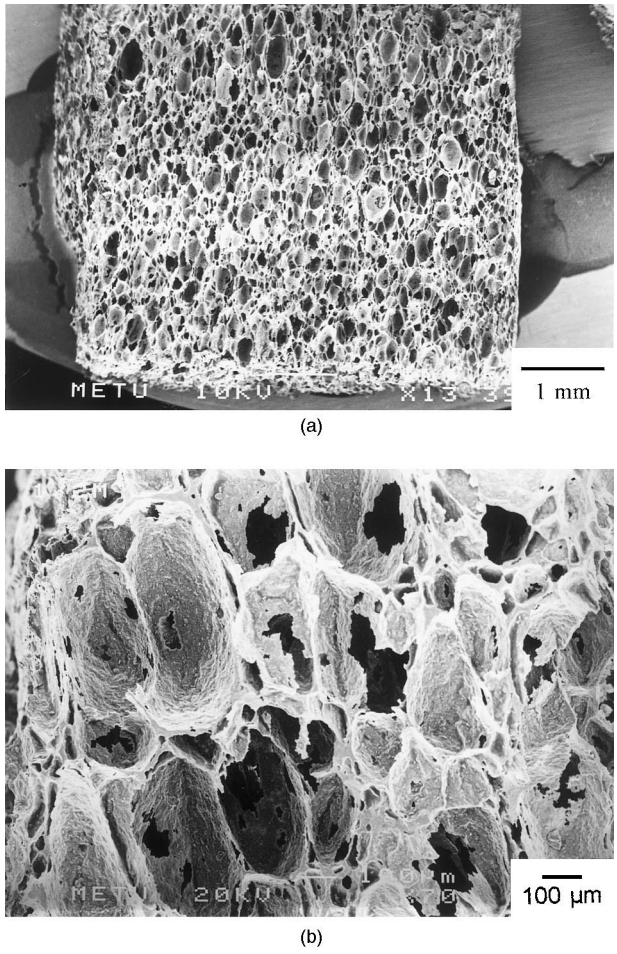

Fig. 2. SEM micrographs of HA bioceramics of 75% porosity.

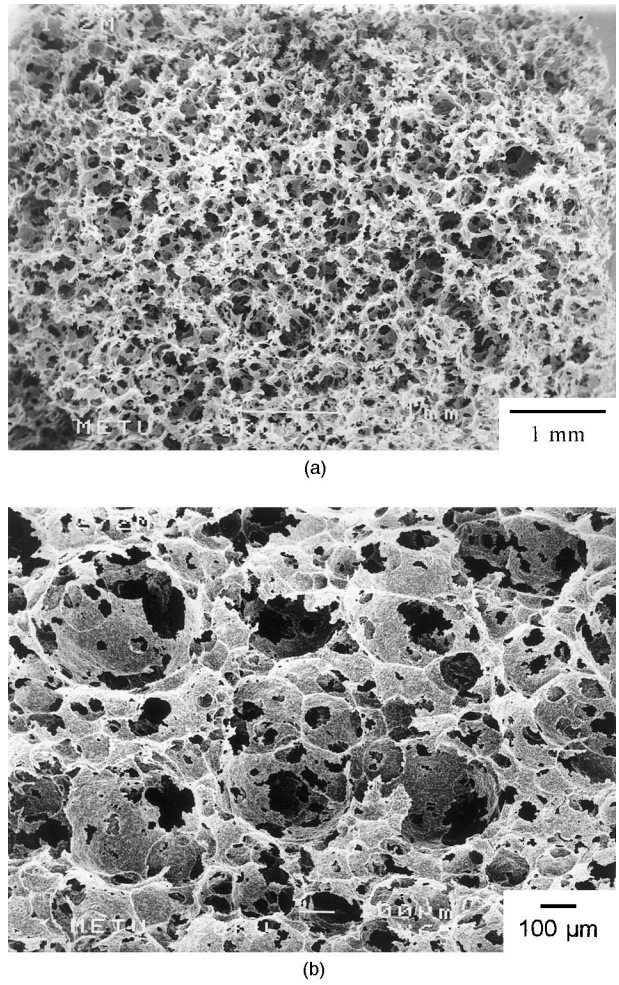

Fig. 3. SEM micrographs of HA bioceramics of 90% porosity.

3 Results and Conclusion

The novel foaming method used in this study,[23,24] to produce macroporous calcium hydroxyapatite bioceramic parts, were shown to be successful in the achievement of total porosity values over the range of 60–90%. The control of the porosity values in the final, sintered HA samples were found to be attained by essentially changing the amounts and concentrations of methyl cellulose, and the intensity of the ultrasonic irradiation used in the processing of the slurries, which yielded the green cakes.

The pore sizes in our HA bioceramics were typically distributed in the range of 100–250 μm. The pores were interconnected. The SEM micrographs given in Figs 1–3 display the microstructures of macroporous HA parts produced in our laboratory with 60, 75 and 90% relative porosity, respectively.

Three-point bending tests performed on rectangular pieces of such samples yielded fracture strengths over the range of 5–10 MPa, for the above-mentioned sequence of porosity values.

In vivo and *in vitro* tests of these HA bioceramics are currently underway.

This technique of porous ceramic manufacturing may easily be used in other ceramic phases and materials, and therefore, has a promising potential for future applications.

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