Acta Biomaterialia 7 (2011) 2327–2336

Contents lists available at ScienceDirect

Acta Biomaterialia

journal homepage: www.elsevier.com/locate/actabiomat

Osteoinduction of porous Ti implants with a channel structure fabricated by selective laser melting

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ARTICLE INFO

Article history:
Received 14 September 2010
Received in revised form 22 January 2011
Accepted 27 January 2011
Available online 2 February 2011

Keywords:
Osteoinduction
Interconnective pore size
Channel structure
Bioactive titanium
Selective laser melting

ABSTRACT

Many studies have shown that certain biomaterials with specific porous structures can induce bone formation in non-osseous sites without the need for osteoinductive biomolecules, however, the mechanisms responsible for this phenomenon (intrinsic osteoinduction of biomaterials) remain unclear. In particular, to our knowledge the type of pore structure suitable for osteoinduction has not been reported in detail. In the present study we investigated the effects of interconnective pore size on osteoinductivity and the bone formation processes during osteoinduction. Selective laser melting was employed to fabricate porous Ti implants (diameter 3.3 mm, length 15 mm) with a channel structure comprising four longitudinal square channels, representing pores, of different diagonal widths, 500, 600, 900, and 1200 μm (termed p500, p600, p900, and p1200, respectively). These were then subjected to chemical and heat treatments to induce bioactivity. Significant osteoinduction was observed in p500 and p600, with the highest observed osteoinduction occurring at 5 mm from the end of the implants. A distance of 5 mm probably provides a favorable balance between blood circulation and fluid movement. Thus, the simple architecture of the implants allowed effective investigation of the influence of the interconnective pore size on osteoinduction, as well as the relationship between bone quantity and its location for different pore sizes.

1. Introduction

Bioactive materials such as bioglass, hydroxyapatite (HA), other calcium phosphate-based biomaterials, and A–W glass ceramic can directly bond to living bones [1] via an apatite layer. Furthermore, when these bioactive materials have a specific porous structure they sometimes become osteoinductive within soft tissues even without the addition of osteogenic cells or bone morphogenetic protein [2–9]. We have previously shown that even porous Ti containing no calcium phosphate can become osteoinductive when it has a complex interconnecting porous structure and bioactive surfaces activated by simple chemical and thermal treatments [10]. In these studies we found that complex macroporous structures, microrough surfaces, and apatite-forming abilities are prerequisites for osteoinduction. Modification of the physico-chemical or surface properties appears to be an attractive method for improving the osteogenic capacity of synthetic materials. Although the clinical utility of osteoinductivity in biomaterials remains controversial, some reports have suggested that this property is a distinct advantage for biomaterials intended for use as bone substitutes [11–14]. Therefore, we believe that it is important in the development of porous biomaterials to tailor osteoinductivity to the specific purpose.

For macroporous structures in general it is recognized that well-defined concavities are important. However, the most suitable pore structure has not been clearly identified. In previous studies we employed plasma sprayed or powder sintered porous bioactive Ti in order to investigate the influence of different porous structures on osteoinduction [10,15]. However, the conventional manufacturing methods employed in our previous studies did not allow precise control over porosity, pore size, and interconnectivity. Furthermore, other osteoinductive materials, such as porous biphasic calcium phosphate [16,17] and porous β-tricalcium phosphate [18,19], are unsuitable for the investigation of macroporous structures because their pore structures change over the implantation period.

Since the 1980s rapid prototyping (RP) technology has emerged as a revolutionary manufacturing process with inherent capabilities for the rapid fabrication of objects of virtually any shape. Using this technology it has become possible to automatically generate three-dimensional (3D) objects by combining computer-aided

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doi:10.1016/j.actbio.2011.01.037

design (CAD) data and computer-aided manufacturing (CAM), and such objects have been used to fabricate scaffolds or orthopedic implants [20–22]. Furthermore, when Ti is used in RP to produce biomaterials it must be non-resorbable and structurally invariant for accurate evaluation of the pore structure. Among the several RP techniques available we adopted the selective laser melting (SLM) [20–22] process, because it is compatible with Ti and can be used to control the configurations directly with dimensional accuracies of the order of several hundred micrometers. SLM is a powder-based additive manufacturing technique that is capable of producing parts in a layer by layer fashion from a 3D CAD model by employing a high energy laser beam that fuses the metal powders present in its focal zone. SLM is now considered a promising fabrication technique, and we intend to apply it to orthopedic implants under load-bearing conditions.

SLM allows the manufacture of implants with an irregular structure that mimics that of human cancellous bone, which thus far had not been possible with commercially pure Ti (cp-Ti). We reported the tensile strength, surface structure, and apatite forming ability of such devices after chemical and thermal treatment [23]. Furthermore, in our preliminary study these cancellous bone mimicking Ti implants were found to be effective bone substitutes in vivo because of the osteoconductivity and osteoinductivity induced by our chemical and thermal treatments.

In the present study, we evaluated the effects of the interconnective pore size on osteoinductivity (speed, amount, and location of bone formation) to establish basic data for developing a porous osteoinductive biomaterial. At the same time we expect to acquire new knowledge about the mechanisms behind osteoinduction.

For this purpose we used SLM to manufacture porous Ti implants, referred to as channel implants, each with four square longitudinal channels, representing interconnected pores, of different diagonal widths, 500, 600, 900, and 1200 μm. This pore design allowed a comparison of osteoinductivity for different pore sizes in identical environments. These implants were subsequently chemically and thermally treated to induce bioactivity [12] and then implanted into the back muscles of mature beagle dogs.

2. Materials and methods

2.1. Materials

Cylindrical channel implants (diameter 3.3 mm, length 15 mm) were designed using a CAD program (Magics®, Materialise, Belgium) (Fig. 1A) and the design data stored in STL file format, commonly used in stereolithography. All the cylinders had four longitudinal square channels, acting as pores, of different diagonal widths, 500, 600, 900, and 1200 μm. In the present paper these cylinders are referred to as p500, p600, p900, and p1200, respectively. As outlined above, we set this range of target channel sizes on the basis of the fact that an SLM machine can only efficiently produce features larger than 500 μm across and the reported upper limit for osteoconduction is around 1000 μm [24–27]. The STL data were converted into slice data that defined the path for the laser scan inside the cross-sectional contours of the produced shapes. The channel implants were fabricated by an EOSINT M270 SLM machine (Electro Optical Systems GmbH, Germany) using cp-Ti powder (>99.5% pure) with particle diameters of less than 45 μm (Osaka Titanium Technologies, Japan).

2.2. Manufacturing porous Ti implants

After the slice data had been obtained the porous Ti powder was melted using an Yb fiber laser beam in an argon gas atmosphere and then the selected slice of the product was solidified. Then the top of the previously melted surface was recoated with a layer (thickness 30 μm) of fresh Ti powder using a recoater blade. Subsequently, selective irradiation was again carried out using the laser beam. These steps were repeated and as the layers were stacked on top of each other the final geometry was achieved. The process

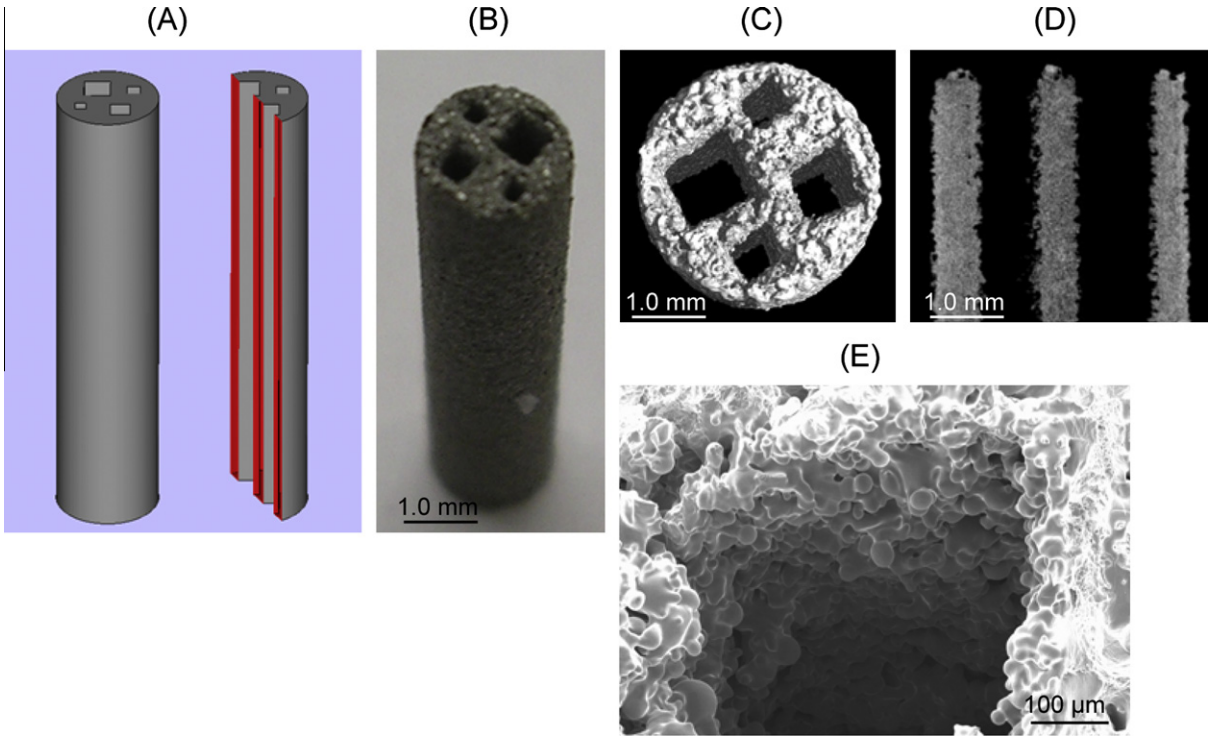

Fig. 1. (A) Computer-aided design (CAD) of a channel implant with four square channels (diagonal widths 500, 600, 900, and 1200 μm). (B) Manufactured channel implant. (C, D) Microcomputed tomography (μCT) of the rapid prototyped channel implant. (E) SEM image of a cross-section of after heat treatment at 1300 °C. Microporous structure was observed on the surface of the smallest of the four pores (500 μm).

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