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Hierarchical tailoring of strut architecture to control permeability of additive manufactured titanium implants

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ABSTRACT

Porous titanium implants are a common choice for bone augmentation. Implants for spinal fusion and repair of non-union fractures must encourage blood flow after implantation so that there is sufficient cell migration, nutrient and growth factor transport to stimulate bone ingrowth. Additive manufacturing techniques allow a large number of pore network designs. This study investigates how the design factors offered by selective laser melting technique can be used to alter the implant architecture on multiple length scales to control and even tailor the flow. Permeability is a convenient parameter that characterises flow, correlating to structure openness (interconnectivity and pore window size), tortuosity and hence flow shear rates. Using experimentally validated computational simulations, we demonstrate how additive manufacturing can be used to tailor implant properties by controlling surface roughness at a microstructual level (microns), and by altering the strut ordering and density at a mesoscopic level (millimetre).

1. Introduction

Titanium (Ti) open-cell foams are popular materials for bone augmentation because they facilitate bone ingrowth and mechanical integration [1–4]. Open-cell foams improve bone integration and implant life time over monolithic materials because: control of the porosity of the foams allows better matching of moduli between the bone and implant, reducing stress shielding; interconnected pore networks (with interconnect size in excess of 100 μm) provide space for cell migration, matrix production and blood vessel ingrowth; and the foams have a high specific surface area which promotes cell attachment [5,6]. As reviewed by Singh et al. [4], there are several methods for producing suitable porous structures, including: metallisation of sacrificial reticulated polymer scaffolds [7]; electrochemical reduction of Ti foams [8]; sacrificial space holder spheres [9]; and high energy laser/electron beam powder melting [3,10–12]. These processes have the ability to produce highly porous structures (porosity >60%) which can be designed to replicate the trabecular bone into which the porous structure eventually integrates, and are an improvement over previous methods, such the straight sintering of Ti particles [13], which yield lower levels of porosity. However, only additive manufacturing (AM) techniques such as Selective Laser Melting (SLM) [3,12] can provide complete control over both percentage porosity and the interconnected network of the structure. In this paper, open-cell Ti foams were produced from commercial purity titanium (CP-Ti) powder using SLM. Many prior studies have demonstrated that the mechanical properties are affected by porosity (e.g. Singh et al. and Mullen et al. [3,12,14]), and the study by Mullen et al. [12] illustrated that properties can also be influenced by the arrangement of struts, even at the same porosity. For example by going from a regular strut alignment to one that is randomised by 30%, the mean compressive strength goes from 49 to 56 MPa.

Flow of blood is required to transport cells, nutrients and growth factors into the porous implant after implantation. Permeability is a measure of the ease at which liquid flows through a porous structure under a pressure gradient and is a convenient way to characterise the bulk flow. Permeability influences vascular invasion and the supply of nutrients required to sustain cell growth and also provides an outlet for the removal of cell debris, thereby increasing its osteoconductive potential [15]. Hui et al. [16] highlighted the importance of implant permeability, reporting results that show high implant permeability enhances the integration of the implant with the host bone. In addition to flow, permeability can also be related to the internal topology of foams, the porosity, and the surface roughness of the foam struts. Moreover, flow through porous foams with different internal structures will result in tortuous paths [17], which impact the fluid shear distribution which can stimulate cell growth [18–21] and the motion of cells along preferential routes.

At low Reynolds numbers, i.e. low fluid velocities, the permeate flux across trabecular bone implants is found to be well predicted by Darcy's Law (Eq. (1)) [16,22,23]:

$$Q = R/A = K\Delta P/\mu L \tag{1}$$

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where R is the volumetric flow rate, A the cross sectional area of the porous structure, Q the permeate flux across the test structure, K the permeability, μ the fluid dynamic viscosity and L the length over which a pressure difference, ΔP, is measured. Typically during cell culture studies using bioreactors, flow rates through the porous structures are low and generally do not exceed 1 ml min^{-1} [24], remaining in the low Reynolds number, or creeping flow regime. However, when experimentally measuring permeability, flow rates of this order are impractically low for easy measurements and much higher flow rates are often employed, thereby creating a situation which shifts the flow from laminar flow to transitional or even turbulent flow. These regions are defined in terms of the Reynolds number (Re) as laminar flow ($Re < 10$), intermediate flow ($10 < Re < 300$) and turbulent flow ($Re > 300$) [25], where Re is defined as:

$$Re = \rho v d / \mu(1-\varepsilon) \quad (2)$$

where ρ is the density of the fluid, v the superficial velocity of the fluid, d the length in the flow direction and ε the porosity of the object that the fluid flows through.

Permeability measurements have been performed by various groups of researchers on bone and other biomaterials [26–29], demonstrating when measuring the permeability using water, foams of nominal characteristics (porosity > 60% and pore size range 100–500 μm) show transitional or turbulent behaviour at flow rates greater than 10^{-2} ms^{-1}. When this is the case, the Dupuit–Forchheimer modification of Darcy's Law must be used [23]:

$$-dP/dL = (\mu/K)v + \rho C v^2 \quad (3)$$

where C is a characteristics coefficient of the fluid.

The permeability of bone depends very much on its structure and density. Kohles et al. [28], working with cubic bovine cancellous bone samples of unit side 1.59 cm, measured intrinsic permeabilities in the range 2.33×10^{-10} m^2 to 4.65×10^{-10} m^2, varying with both location and direction. Grimm and Williams [30] measured the permeability of human trabecular bone from calcanei in the mediolateral direction to be in the range 0.4×10^{-9} m^2 to 11×10^{-9} m^2. Ideally implant structures could be tuned to match this order of magnitude range in values.

Although prior Ti foams have not been designed explicitly for matched permeability, a number of authors have measured this property for different foam types. For example Shimko et al. [26] found their tantalum porous structures to have permeabilities ranging from 2.1×10^{-10} m^2 to 4.8×10^{-10} m^2, and Despois et al. [23] found the permeability increased from 1.23×10^{-12} m^2 to 3.56×10^{-10} m^2 as they reduced the density, keeping pore size constant at 75 μm. They also tried changing the pore size from 75 to 400 μm, which increased the permeability from 4.0×10^{-10} m^2 to 7.64×10^{-10} m^2. Since a larger value for permeability means easier flow, reducing density (large flow passages) or having bigger pores (less surface area) should increase permeability as observed. Singh et al. [31], characterised Ti foams with 65% porosity, and found their structures to have permeability values in the range 1.17×10^{-10} m^2 to 1.63×10^{-10} m^2. In summary, common metal foams being considered for implants were experimentally found to have permeability values in the range 10^{-12} m^2 to 10^{-8} m^2.

Performing computational modelling of fluid flow using computational fluid dynamics (CFD) codes overcomes many experimental difficulties and allows direct design and tailoring of permeability. Two-dimensional (2D) numerical approaches to calculate permeability using the Navier–Stokes equation as discussed later were reported by Nagelhout et al. and Fuloria et al. [32,33] on fluid flow through a columnar dendritic network. Other authors [34,35] have used the boundary element method to calculate permeabilities for a variety of packing of fibrous materials. High resolution X-ray micro- tomography (μCT) imaging techniques allow the reconstruction of the three-dimensional (3D) morphology of the structure [36]. Its advantageous non-destructive 3D imaging properties allow the direct computation of permeability from real 3D scaffold volumes. Bernard et al. presented several studies on the 3D simulation of permeability in Al-Cu alloys based on μCT data [37–40]. Singh et al. [31] studied the permeability of Ti foams produced by the space-holder technique for spinal fusion devices using μCT data. Darcy's Law was applied to calculate the permeability after numerically solving the Navier–Stokes equation. However, the impact of implant design factors on permeability and how they may be tailored has not yet been studied previously.

The aim of this study is primarily to determine how each SLM variable, such as randomness of the pore structure, percentage porosity and surface topography, affects permeability. The goal is to derive relationships which allow the independent tailoring of the permeability to optimise the design of implants. First, a dual computational and experimental approach for the evaluation of permeability was applied to a range of samples produced by the SLM process with structures modified during their manufacture to validate the models. Then structural modifications were made on the computer design of Ti structures at both microscopic (surface roughness) and macroscopic (strut architecture) levels.

2. Materials and methods

2.1. Ti foam preparation

Samples were produced from commercially pure Ti (Grade 1) metal powder by selective laser melting (SLM) using an MCP Realizer 2, 250 SLM system (MTT, UK). The full manufacturing process has been detailed by Mullen et al. [3,12]. The samples were constructed using the Unit Cell (UC) Approach [3] which comprises the following sequences:

> A computer-aided design (CAD) model of the porous structure was built using the Manipulator© software suite (University of Liverpool, UK). The 3D geometries were completely filled with cubic unit cells of a single defined edge length. All UCs were filled with a connected lattice structure by joining sets of points and vectors in 3D space to form regular octahedron (this geometrical shape allows tessellation).

To produce the ‘irregular’ structures, a randomisation process was used based upon perturbation of the Cartesian coordinates of the vertices of the UC with respect to the regular structure by a specific percentage of the UC size while maintaining connectivity [12]. (Note that the structures produced are in fact pseudo-random as the randomness is fully reproducible.)

Both the regular and irregular structures form completely open porosity that is fully connected to the surface.

The points and vectors were trimmed to the CAD file of the cylinder, and then sliced (50 μm intervals) to produce a set of points on all layers, de-noting the firing position of the scanning laser. These locations, together with the laser properties (beam diameter, dwell time and energy), determine the strut diameter (references [3 and 12] provide details of the SLM process). Upon completion, the parts were removed from the argon atmosphere build chamber, wire electro-discharge machined (EDM) from the substrate plate and cleaned of all un-fused powder, after which a sintering operation was carried out under vacuum at 1400° C for 3 h to heat treat the parts.

In this study, cylindrical samples of 2 different sizes (equal diameter and height of 4 and 10 mm) with 4 levels of randomness were produced. Samples evaluated were nominally 65% porous, with a UC size of 600 μm and were produced with 0, 10, 20 and 30% randomisation, with the sequential operation being illustrated in Fig. 1.