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Research paper

Porous titanium materials with entangled wire structure for load-bearing biomedical applications

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ABSTRACT

A kind of porous metal-entangled titanium wire material has been investigated in terms of the pore structure (size and distribution), the strength, the elastic modulus, and the mechanical behavior under uniaxial tensile loading. Its functions and potentials for surgical application have been explained. In particular, its advantages over competitors (e.g., conventional porous titanium) have been reviewed. In the study, a group of entangled titanium wire materials with non-woven structure were fabricated by using 12–180 MPa forming pressure, which have porosity in a range of 48%–82%. The pores in the materials are irregular in shape, which have a nearly half-normal distribution in size range. The yield strength, ultimate tensile strength, and elastic modulus are 75 MPa, 108 MPa, and 1.05 GPa, respectively, when its porosity is 44.7%. The mechanical properties decrease significantly as the porosity increases. When the porosity is 57.9%, these values become 24 MPa, 47.5 MPa, and 0.33 GPa, respectively. The low elastic modulus is due to the structural flexibility of the entangled titanium wire materials. For practical reference, a group of detailed data of the porous structure and the mechanical properties are reported. This kind of material is very promising for implant applications because of their very good toughness, perfect flexibility, high strength, adequate elastic modulus, and low cost.

1. Introduction

Metallic biomaterials are preferential choices in load-bearing applications in the field of bone tissue engineering and regenerative medicine due to their outstanding mechanical strength and good toughness. Among them, titanium and its alloys exhibit excellent biocompatibility, good corrosion resistance, and relatively lower stiffness (compared with other metallic biomaterials such as cobalt-based alloys, stainless steel, etc.), which are beneficial to the durability of the load-bearing implants and reduction of the stress-shielding effect. These advantages have facilitated the practical clinical applications of titanium and its alloys, such as artificial titanium alloy joint (Long and Rack, 1998), dental titanium implant (Bréme et al., 1993), coronary titanium–nickel stent (Eigler et al., 1993), etc. In the past decades, attempts were frequently made to seek optimal, mechanical and biocompatible properties for titanium implants (Niinomi, 2008; Geetha et al., 2009), so as to remedy the mismatch of the elastic moduli (between titanium implant and cortical/cancellous bone) and the bioinert properties of the titanium implant surface. The former can lead to stress-shielding over the surround-

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ing bone, which results in resorption of the bone and aseptic loosening of the implant (Jacobs et al., 1993); the latter may cause poor bone–implant interfacial bonding, because the bulk titanium surface develops interfacial fibrous tissue to form encapsulation that isolates the implants from their surroundings (Chang et al., 1996). In order to solve these problems, alloying is a metallurgical method to reduce the elastic modulus via forming full beta structure. So far the elastic modulus as low as thirty GPa has been achieved in some beta titanium alloys (Geetha et al., 2009; Laheurte et al., 2010). But compared with cortical bone (4.4–28.8 GPa) and cancellous bone (0.01–3.0 GPa) (Geetha et al., 2009), the beta titanium alloy is still stiffer. It seems that only a very limited room for reduction of the elastic modulus of the bulk titanium alloy remains. The surface modifications on titanium implants have previously demonstrated that bioinert titanium metal could be converted into bioactive material through specific chemical and thermal treatments (Fujibayashi et al., 2001; Kim et al., 2000; Nishiguchi et al., 1999), by which the smooth titanium surface was processed into a specific porous structure (Fujibayashi et al., 2004; Kim et al., 2000). These previous works implied the superiority of the porous titanium materials as implants. In fact, porous biomaterials have received a lot of attention in past ten years. The porous titanium can provide adequate macro/micro-pores for bone ingrowth, vascularization, and flow transport of nutrients and metabolic waste (Manukyana et al., 2010), which is supposed to be osteoconductive, osteoinductive and capable of osseointegration (Albrektsson and Johansson, 2001; Stevens, 2008).

It is well accepted so far that the titanium material with porous structure, which can imitate cancellous bone in structure and properties, facilitates proliferation of cells into the structure and provides space for bio-factor (cell, gene and/or protein) delivery (Hollister, 2005). Such bone tissue ingrowth through the pores forms stable long-term anchorage for biological fixation of the implant (Garrett et al., 2006; Krishna et al., 2007). It is important that the porous network must interconnect and the appropriate pore size for in vivo bone ingrowth should be in the range of 100–500 μm (Bungo et al., 2006; Freyman et al., 2001; Garrett et al., 2006; Holy et al., 2000; Li et al., 2007). Some investigations validated that the porous titanium implants could be improved in osteoconductive properties as porosity increased, and the amount of new bone growth obviously increased as the pore size increased (Bungo et al., 2006; Li et al., 2007). Another benefit of porous titanium is that the effective elastic modulus can be tuned to match the modulus of bone to reduce the problems associated with stress-shielding. The mechanical properties of porous titanium are dependent on porosity, porous morphology and pore size distribution as these determine the spatial shape and size of the struts or cell-walls (for honeycombed structure) between the pores which are bearing the load. The nature of the struts or cell-walls together with the holistic architecture constructed by the struts is crucial for the macro-mechanical behaviors of the bulk porous titanium. All these key macroscopic parameters are controlled by the fabrication methods of the porous titanium materials.

There are many techniques to produce porous titanium, and various methods may lead to different porosities and morphologies. The earliest metal foaming techniques (Banhart, 2001; Körner and Singer, 2000) such as 'foaming by gas injection' and 'foaming with blowing agents', which have been widely used in the aluminum foaming industry, are inadequate to produce porous titanium for orthopedic applications because only closed-cell structure can be obtained. The conventional metallurgical method by titanium powder sintering was used to fabricate porous titanium successfully (Oh et al., 2003), but the porosity was very limited. When space-holding material was introduced in the powder sintering processing the porosity could been increased up to 80% or even more (Banhart, 2001; Garrett et al., 2006; Wen et al., 2001), and this method could produce bimodal pore sizes by controlling the space-holder size and the titanium particle diameter. Furthermore, a modified powder metallurgical technique called 'sponge replication' (Cachinho and Correia, 2008; Lee et al., 2009; Li et al., 2005) or 'impregnating method' (Zhao et al., 2008) has been used in fabrication of the porous titanium and its alloys, by which the pore size, shape, and distribution can be well-adjusted in a large scale. Some interconnected porous structures that perfectly mimic the spatial morphology of the cancellous bone can be easily reproduced by this method. However, the major drawbacks of the above metallurgical methods are the difficulties to avoid contamination and impurity phases in the titanium materials, which significantly deteriorate the mechanical properties of the titanium struts. In addition, the sintered titanium struts likely contain cracks, unbonded contact nodes between the powder particles, and other metallographic defects, thus, the porous titanium materials cannot bear tensile stress and even exhibit 'brittle' nature. This is the reason that almost all the investigations did not report the tensile strength of the powder sintered porous titanium materials in the literature. This limitation implies a risk of the orthopedic applications of the porous titanium in vivo where a tensile or impact load may be applied on the implant.

In recent years a rapid prototyping technique has been utilized to fabricate porous titanium by using laser power to sinter/remelt titanium powders (Hollander et al., 2006; Krishna et al., 2007, 2008; Mullen et al., 2008; Traini et al., 2008), which is based on a CAD and layer-by-layer manufacturing process. It is of great benefit to accurately controlling the pore shape, size and distribution by beforehand design and programming, so that an ideal porous architecture can be constructed. With alternative fabrication approaches, this technique has been developed into 'direct laser sintering' (Hollander et al., 2006; Krishna et al., 2007; Traini et al., 2008) and 'selective laser melting' (Krishna et al., 2008; Mullen et al., 2008). The rapid prototyping technique provides good feasibility and excellent flexibility for production of custom-built metallic implants with complicated structures. However, the porous titanium production also suffers from impurities, inclusions and other defects associated with the laser sintering or melting processing, which significantly deteriorate the mechanical properties such as tensile-stress resistance and toughness. In the view of physical metallurgy, the struts in the porous titanium with as-sintered (laser sintering) or as-solidified (laser melting) microstructure usually exhibits much lower ductility than that of the wrought or plastically deformed counterparts. It is unlikely that the laser sintered/melted porous titanium implants can stand against tensile and impact stress, though