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Review

Topological design and additive manufacturing of porous metals for bone scaffolds and orthopaedic implants: A review

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ABSTRACT

One of the critical issues in orthopaedic regenerative medicine is the design of bone scaffolds and implants that replicate the biomechanical properties of the host bones. Porous metals have found themselves to be suitable candidates for repairing or replacing the damaged bones since their stiffness and porosity can be adjusted on demands. Another advantage of porous metals lies in their open space for the in-growth of bone tissue, hence accelerating the osseointegration process. The fabrication of porous metals has been extensively explored over decades, however only limited controls over the internal architecture can be achieved by the conventional processes. Recent advances in additive manufacturing have provided unprecedented opportunities for producing complex structures to meet the increasing demands for implants with customized mechanical performance. At the same time, topology optimization techniques have been developed to enable the internal architecture of porous metals to be designed to achieve specified mechanical properties at will. Thus implants designed via the topology optimization approach and produced by additive manufacturing are of great interest. This paper reviews the state-of-the-art of topological design and manufacturing processes of various types of porous metals, in particular for titanium alloys, biodegradable metals and shape memory alloys. This review also identifies the limitations of current techniques and addresses the directions for future investigations.

1. Introduction

Bone is a complex tissue that continually undergoes dynamic biological remodelling, i.e., the coupled process whereby osteoclasts resorb mature bone tissue followed by osteoblasts that generate new bone to maintain healthy homeostasis of bone [1]. This unique feature of bone underpins its ability to remodel itself to repair damage. However, when a bone defect exceeds a critical non-healable size, external intervention is required to supplement self-healing if the defect is to be bridged [2]. Despite recent advances in biomaterials and tissue engineering, repair of such a critical-sized bone defect still remains a challenge. The optimal choice is to use autograft (patients' own tissue) [3]. However, harvesting autograft tissue creates the morbidity associated with a second surgical site. An alternative choice is allograft tissue (taken from another person), which carries the risk of transmissible disease and depends on logistic circumstances (limited availability). The insufficiencies of application of autograft and allograft tissue have led to greater research efforts to identify biomimetic materials and structures that are suitable for skeletal repair without the inherent problems.

Metals and alloys have a long history of application as bone implants [4–7]. Among them, the use of stainless steels, cobalt (Co) based alloys (CoCrMo), and titanium (Ti) and its alloys are well established due to their good biocompatibility, satisfactory mechanical strength and superior corrosion resistance [5]. However, implants made of these materials are usually much stiffer than natural bones, leading to stress shielding - a major source for bone resorption and eventual failure of such implants [5]. Cortical bone (compact bone) has elastic moduli ranging from 3 to 30 GPa, while trabecular or cancellous bone has significantly lower elastic moduli of 0.02–2 GPa. Most current implant materials have much higher

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moduli than those of bones, e.g., Ti6Al4V has a modulus of around 110 GPa and CoCrMo alloys have a modulus of around 210 GPa [5,6,8]. Therefore, to avoid stress shielding at the bone-implant interface, the equivalent Young's modulus and yield stress have to be adjusted when using these bulk materials. An effective method is to introduce adjustable porosity or relative density as proposed by Gibson and Ashby [9] for isotropic materials.

Traditional methods for fabricating open-cell porous metals include liquid state processing (direct foaming, spray foaming, etc.), solid state processing (powder metallurgy, sintering of powders and fibres, etc.), electro-deposition and vapour deposition [10,11]. Although the shape and size of the pores can be adjusted by changing the parameters of these manufacturing processes, only a randomly organized porous structure can be achievable. However, additive manufacturing (AM) technologies can fabricate porous metals with predefined external shape and internal architecture [2,12–14]. Metal-based additive manufacturing (MAM) techniques, such as selective laser melting (SLM) and electron beam melting (EBM), are computer controlled fabrication process based on layer-wise manufacturing principles. SLM [15–17] and EBM [18,19] are increasingly used for the fabrication of porous metals with complex architecture. Instead of using electron beam as the energy source in EBM, the SLM technology uses laser beam with adjustable wavelength. Therefore, EBM can only process conductive metals whereas SLM can process polymer or ceramics as well as metal. Furthermore, due to more diffuse energy (larger heat-affected zone), EBM process has larger minimum feature size, median powder particle size, layer thickness, resolution and surface finish [20]. The robust application of MAM technologies requires extensive material, process and design knowledge, specific to each MAM technology [21]. MAM system behaviour is subject to significant stochastic error and experimental uncertainties, requiring that “assumptions are necessary to simplify the problem” [22]. Sources of error include: complex and transient heat transfer phenomena [23], geometric effects [24] with poorly defined powder thermal properties [25]. MAM prediction error can lead to excess melt pool temperature [26], resulting in undesirable microstructure, residual stress, local porosity, and surface roughness. Understanding the effects of design decisions on temperature related process defects is critically important to the process control. Comprehensive reviews of AM technologies can be found elsewhere [27,28].

Recent successes in orthopaedic regenerative medicine have promised an exciting future of AM technology. The world's first additively manufactured mandible was implanted in a patient by Dr. Jules Poukens and his team in 2012 in Belgium [29]. A full lower jaw implant (mandible in Fig. 1) was coated with hydroxyapatite and implanted in an 83 year old lady. The porous implant was slightly heavier than a natural jaw, and provided robust attachment of muscles and sufficient space for nerves [29]. Skull reconstructions with AM parts have been performed successfully by using digital design and AM. Mertens et al. [30] successfully reconstructed a class III defect using AM manufactured titanium implants, which provided both midfacial support and a graft fixture (midface defect in Fig. 1). Jardini et al. [31] in Brazil designed and AM fabricated a customized implant for the surgical reconstruction of a large cranial defect. In 2014, Prof. Peter Choong, an Australian surgeon from St Vincent's Hospital, together with scientists from the Commonwealth Scientific and Industrial Research Organization (CSIRO) and Anatomics, successfully implanted the world's first 3D-printed titanium heel bone into a patient [32].

Typical design and application approaches of porous metallic implants normally include the design of scaffold, AM and post-processing (heat-treatment and surface modification) as illustrated in Fig. 1. This review aims to identify the current status and the future directions of design-oriented AM technology in producing porous metallic structures for bone tissue repair, with a particular emphasis on topological design of internal architecture of porous metals for bone implants.

2. Structure and properties of bone

2.1. Structure of bone

Bone is a natural composite containing both organic components (mainly type-I collagen, but also type-III, type-IV collagen and fibrillin) and inorganic crystalline mineral (e.g., hydroxyapatite, HA) [1,22,33,34], as illustrated in Fig. 2. The structure of bone is similar to reinforced concrete that is used in the building industry. The function of HA crystals and collagen molecules are like the steel rod and cement to concrete: one part provides flexibility and the other provides strength and toughness. Type-I collagen is a triple helix of ~1.5 nm in diameter and ~300 nm in length. It is the primary organic components of bone. Other non-collagenous proteins include glycoproteins and bone specific proteoglycans [1]. Hydroxyapatite is the inorganic component of bone and is plate-shaped of 50 $\times$ 25 nm in size and 1.5–4 nm thick [35]. The HA crystals are oriented in a periodic array in the fibrils, preferentially with their *c* axis parallel to the collagen fibrils [35]. These two phases account for about 95 wt. % of the dry bone.

Bone has a hierarchical structure. Each level performs diverse mechanical, biological and chemical functions. The hierarchical levels of bone include macroscale, microscale, sub-microscale, nanoscale, and sub-nanoscale (Fig. 2). The macroscale level represents the overall shape of the bone. Bone can be classified as compact bone (cortical bone), and trabecular bone (cancellous bone). Compact bone is almost solid, with only ~3–5% spaces for osteocytes, canaliculi, blood vessels, and erosion cavities etc. There are large spaces in trabecular bone. The pores in trabecular bone are filled with bone marrow, and the porosity varies between 50 and 90% [1,36–38]. The building block of compact bone is the osteons, which are of the size ranging from 10 to 500 μm, whereas the trabecular bone is made of a porous network of trabeculae. At the micron- and nano-scales, aggregated type-I collagen and HA form the collagen fibril. The reinforced collagen fibre is a universal building element for both compact and trabecular bones.

2.2. Mechanical properties of bone

Mechanical properties of bone vary significantly with age, anatomical site and bone quality. It continues to be a major scientific challenge to fully understand the mechanics of living bones [36–38]. Among the various biomechanical properties of bone (stiffness, strength, creep and fatigue), elastic modulus has attracted the most research interest because of its critical importance for characterizing various bone pathologies and guiding artificial implant design. The elastic modulus and strength of bone are anisotropic. Compact bone is both stronger and stiffer when loaded longitudinally along the diaphyseal axis than the radial transverse directions (Table 1). It is also stronger in compression than in tension. Trabecular bone is an anisotropic and porous composite. Like many biological materials, trabecular bone displays time-dependent behaviour as well as damage susceptibility during cyclic loading [41]. The mechanical properties of trabecular bone depend on not only the porosity, but also the architectural arrangement of the individual trabeculae. The physical and mechanical properties of human bone are summarized in Table 1 (values are averaged from reported data) [1,26,34,36–38,42,43].