



Fatigue crack propagation in additively manufactured porous biomaterials



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ABSTRACT

Additively manufactured porous titanium implants, in addition to preserving the excellent biocompatible properties of titanium, have very small stiffness values comparable to those of natural bones. Although usually loaded in compression, biomedical implants can also be under tensional, shear, and bending loads which leads to crack initiation and propagation in their critical points. In this study, the static and fatigue crack propagation in additively manufactured porous biomaterials with porosities between 66% and 84% is investigated using compact-tension (CT) samples. The samples were made using selective laser melting from Ti-6Al-4V and were loaded in tension (in static study) and tension-tension (in fatigue study) loadings. The results showed that displacement accumulation diagram obtained for different CT samples under cyclic loading had several similarities with the corresponding diagrams obtained for cylindrical samples under compression-compression cyclic loadings (in particular, it showed a two-stage behavior). For a load level equaling 50% of the yield load, both the CT specimens studied here and the cylindrical samples we had tested under compression-compression cyclic loading elsewhere exhibited similar fatigue lives of around 10^4 cycles. The test results also showed that for the same load level of $0.5 F_y$, the lower density porous structures demonstrate relatively longer lives than the higher-density ones. This is because the high bending stresses in high-density porous structures gives rise to local Mode-I crack opening in the rough external surface of the struts which leads to quicker formation and propagation of the cracks. Under both the static and cyclic loading, all the samples showed crack pathways which were not parallel to but made 45° angles with respect to the notch direction. This is due to the fact that in the rhombic dodecahedron unit cell, the weakest struts are located in 45° direction with respect to the notch direction.

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1. Introduction

Autologous and allogeneic bone grafts from iliac crest are known as the gold standard for treating large bone defects [1]. However, problems such as donor morbidity, low bone stock availability, and post-surgery complexions has limited their use and given rise to use of synthetic materials [2]. Titanium and its alloys are biocompatible and corrosion resistant materials that have been long used as parts of orthopedic implants. The much higher stiffness of titanium compared to bone, however, unloads the natural bone around the implant and causes bone resorption. Porous titanium implants, while keeping the excellent biocompatible properties of titanium, have much smaller stiffness values that can be as low as those of natural bone [3]. The interconnected hollow space inside the porous materials also allows for easy body fluid transport and therefore subsequent bone growth inside the porous structure [4,5].

Additive manufacturing techniques such as selective laser melting (SLM) [6], selective laser sintering (SLS) [7], fused deposition modeling (FDM) [5,8], and selective electron beam melting (SEBM) [9] have several advantages over traditional manufacturing techniques. These advantages include precise control over the unit cell size and shape, providing the possibility of integrating the solid and porous parts inside an implant, and creating patient-specific implant. While the static properties of additively manufactured porous biomaterials have been usually found in the range of natural bone [5,10–13], their fatigue properties are more difficult to determine due to their high porosities as well as the defects created during manufacturing processes. Since implants are usually loaded several times in daily activities, understanding their fatigue properties is of importance [14]. Since the porous biomaterials are usually loaded in compression, the few recent studies on their fatigue behavior is dedicated to their fatigue mechanical properties under compression-compression loads [14–17]. Although the weight-bearing implants are usually loaded in compression (to carry body weight), depending on their shape, current position, and direction and magnitude of other external loads, the implants can also be loaded in

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bending, shearing, and tension which leads to mode-I fatigue crack initiation and propagation. Indeed, even when the porous structure as a whole is under compression, (parts of) individual struts might be under tension.

While no study has been carried out on the crack propagation in the additively manufactured porous structures, there are a few studies in which the fracture toughness and crack propagation properties of traditional foams have been studied. The fracture toughness measurements of Olurin et al. [18] on compact tension (CT) specimens showed an R-curve behavior. Tests on specimens with sharp notches showed that their fracture behavior is notch insensitive but is very dependent on the development of crack bridging ligaments behind the crack tip. Kabir et al. [19] studied the effect of various parameters such as loading rate, specimen size, crack length, foam density, cell orientation (flow and rise-direction), cross-linking, and solid material properties on the fracture behavior of polymeric foams. They found foam density and the mechanical properties of the solid material the most important parameters in determining the fracture behavior of polymeric foams. More recently, Kashef et al. investigated the fracture toughness [20] and fatigue crack growth [21] behavior of open-cell titanium foams manufactured by space-holder method for medical applications. They observed that compared to solid titanium, titanium foams with 60% porosity had a significantly higher Paris exponent, which can be explained by crack bridging and crack closure [21]. They also showed that the fatigue crack grows through the weakest path throughout the foam.

In this study, the static and fatigue crack propagation behavior of additively manufactured porous biomaterials with porosities between 66% and 84% is investigated using CT samples. The samples were made using SLM technique and were tested under tension (in static study) and tension-tension (in fatigue study) loadings. The failure mechanisms were investigated using displacement increment data and damage propagation pathways. To better observe the stress distribution in the samples, finite element (FE) models based on beam elements were also developed and solved.

2. Methodology

2.1. Experimental tests

Four types of samples were manufactured by selective laser melting method (Layerwise NV., Belgium). All specimens had the same general dimensions of 30 mm × 31.25 mm × 6 mm. The other dimensions of the specimens are shown in Fig. 1 that are selected according to standard ASTM E647 – 13a. The specimens had strut cross-section dimensions between 140 and 251 μm and pore sizes between 486 and 608 μm (Table 1). The above-mentioned strut and pore dimensions led to relative densities between 16.3% and 33.6% (Table 1 and Fig. 2). The specimens were processed on top of a solid substrate in inert

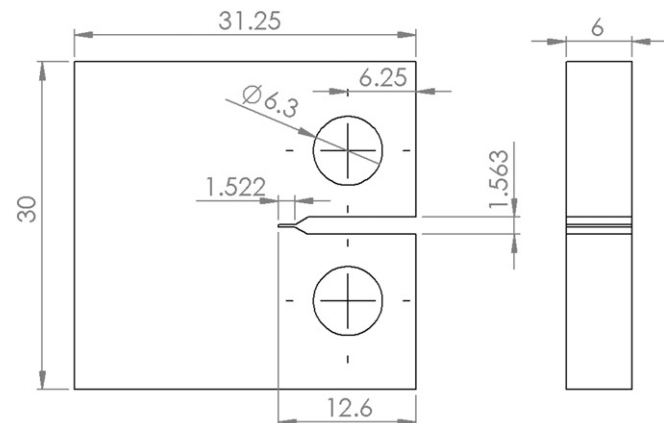


Fig. 1. Dimensions of the manufactured samples.

Table 1

The porosity, pore sizes, and strut sizes of the four samples tested in this study [14].

	Ti 120–500	Ti 170–500	Ti 170–450	Ti 230–500
Porosity, dry weighing (%)	83.7 ± 0.3	77.1 ± 0.5	70.1 ± 0.3	66.4 ± 0.3
Porosity, Archimedes (%)	83.6 ± 0.3	76.8 ± 0.5	69.7 ± 0.3	65.8 ± 0.3
Porosity, micro-CT (%)	84.22	77.68	71.20	68.45
Pore size, nominal (μm)	500	500	450	500
Pore size-micro-CT (μm)	560 ± 173	608 ± 182	486 ± 162	560 ± 186
Strut size, nominal (μm)	120	170	170	230
Strut size, micro-CT (μm)	140 ± 38	218 ± 62	216 ± 64	251 ± 76

atmosphere and were made of the biomedical Ti-6Al-4V ELI titanium alloy spherical powders according to ASTM B348. All the specimens were designed using the same type of repeating unit cell, i.e. rhombic dodecahedron. Each specimen type was first tested under static loading (Instron Electropuls e10000 with 10 kN load cell) using a displacement rate of 1 mm/min. The load in the end of the elastic region in the load-displacement curve was measured for each specimen type and a value about half of that value was used as the maximum stress in the cyclic loading. The load ratio (the ratio of the maximum to the minimum load in each loading cycle) was set to $R = 0.1$. All the specimens were tested under tension-tension fatigue. The loading-unloading frequency was set to 15 Hz in all the tests.

2.2. Numerical modeling

To better observe stress distribution in the struts of the specimens, FE models (Fig. 3) were made and simulated using a commercial package (i.e. ANSYS static standard solver). The FE models had general dimensions similar to those of actual specimens. For each specimen type, the strut size was set to the strut size of the corresponding specimen. Due to presence of irregularities in the cross-section area of the struts in the additively manufactured specimens, the struts of the FE model were discretized using the Timoshenko beam elements with different diameters. Timoshenko beam elements in ANSYS uses linear interpolation and takes transverse shear deformation into account. The strut diameter at each location of the FE model were taken from a Gaussian distribution function whose mean value was taken from the mean value of the manufactured struts (Table 1) and its standard deviation was set to the values measured experimentally (Table 1). The solid part of the manufactured specimens was also created in the FE model and discretized using cubic elements. The DOFs of the neighbor nodes located in the interface between the solid and porous regions in the neighborhood of the grips were coupled to each other. The nodes belonging to the volumetric elements around the lower grip were constrained in all the directions. The nodes belonging to the volumetric elements around the top grip were constrained in X and Z directions and were displaced in Y direction. The displacement values of $d = 0.1, 0.2, 0.3, 0.4$, and 0.5 mm were considered. $d = 0.5$ mm was the maximum displacement below which the experimental load-displacement curve was linear for all the test samples. The mechanical properties of bulk Ti-6Al-4V (Table 2) were used for modeling the mechanical behavior of the struts as well as the solid portions of the specimens.

3. Results

3.1. Experimental tests

Under isostatic loading, all the specimens showed an initial linear part in their load-displacement curve (Fig. 4). In this regime, no local damage was observed in the struts. After observing some initial damages in the notch tip of the specimens, the load-displacement curve slope decreased very rapidly and the load-displacement curve reached a maximum point (at displacement range around 0.6–0.8 mm, Fig. 4).

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