Perspectives on Additive Manufacturing Enabled Beta-Titanium Alloys for Biomedical Applications

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Abstract: “Stress shielding” caused by the mismatch of modulus between the implant and natural bones, is one of the major problems faced by current commercially used biomedical materials. Beta-titanium (β-Ti) alloys are a class of materials that have received increased interest in the biomedical field due to their relatively low elastic modulus and excellent biocompatibility. Due to their lower modulus, β-Ti alloys have the potential to reduce “stress shielding.” Powder bed fusion (PBF), a category of additive manufacturing, or more commonly known as 3D printing techniques, has been used to process β-Ti alloys. In this perspective article, the emerging research of PBF of β-Ti alloys is covered. The potential and limitations of using PBF for these materials in biomedical applications are also elucidated with focus on the perspectives from processes, materials, and designs. Finally, future trends and potential research topics are highlighted.

Keywords: Additive manufacturing; 3D printing; Powder bed fusion; Selective laser melting; Electron beam melting; Titanium

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1. Introduction

Powder bed fusion (PBF) is a group of additive manufacturing (AM) or three-dimensional (3D) printing techniques. When equipped with lasers as energy sources, the processes are also known as laser powder bed fusion (L-PBF). L-PBF is also commercially known as selective laser melting (SLM) or direct metal laser melting1,2. Another type of PBF process uses electron beam as the energy source and thus is known as electron beam powder bed fusion (EB-PBF) or commercially as selective electron beam melting. These manufacturing processes have shown successes in processing alloys3 and even ceramics4,5. Like any other AM techniques, PBF has the capability to fabricate functional parts with complex geometry due to its freeform fabrication capabilities. The process starts from designing of the parts using a computer-aided design (CAD) software and exporting the data files into the PBF equipment. The data files are also first input with process parameters and then sliced into two-dimensional cross sections. The PBF process itself involves a cycle of depositing powder layers onto the build platform or previously processed layers, then the melting of the powder selectively using laser or electron beam. The areas that are melted follow the cross sections from the sliced CAD data file. After this step, the build platform is then lowered and a new powder layer is deposited. The cycle repeats until the full 3D components are fabricated6. As a result of the cyclic process and repeated thermal cycles, the materials undergo solid-liquid-solid phase transformations7. These unique physical phenomena bring about the microstructural changes which affect the mechanical properties of the materials. The detailed PBF process is also described elsewhere8,11. A schematic of the PBF process is shown in Figure 1.

Most of the current materials used for biomedical implants commercially, such as 316L stainless steel, CoCrMo, and even Ti6Al4V, have a problem which is the...
mismatch of their modulus to that of natural bones. While the elastic modulus of human cortical bone ranges from 17.6 to 28 GPa, that of CoCrMo and 316L stainless steel is more than 200 GPa[12] and even Ti6Al4V has elastic modulus of approximate 130 GPa[13]. These are many times higher than that of human cortical bone even when they are commonly used[14]. Osteolysis can be attributed to this modulus mismatch as it lessens the loading on the natural bones that are neighboring the implant, leading to bone resorption. Finally, implant loosening would occur.

Recently, due to their higher strength, lower modulus and better corrosion resistance, beta-titanium (β-Ti) alloys have been identified as potential materials to improve implants quality[7,15,16]. β-Ti alloys are titanium alloys where the β-phase is significantly retained in equilibrium or at least on quenching from the β-phase without transformation into martensite or α-phase[7]. Furthermore, β-Ti alloys usually consist of non-toxic elements such as tantalum, niobium, molybdenum, tin, and zirconium. To obtain β-Ti alloys that are biocompatible, niobium and tantalum which are β-phase stabilizers for titanium are the common choices for alloying elements due to their high biocompatibility. The other elements such as zirconium, molybdenum, and tin are added to further modify phases and microstructures of the β-Ti alloys.

In this article, an overview of the processing, microstructure, and properties of β-Ti alloys processed by PBF that can be used in biomedical applications is discussed. The potential and limitations of using PBF for these materials in biomedical applications are also elucidated with focus on the perspectives from processes, materials, and designs. Finally, future trends and potential research topics are highlighted.

2. β-Titanium alloys by powder bed fusion

2.1. Powder bed fusion

It is important to understand the physical phenomena that occur during the PBF process to obtain parts with good quality, that is, parts that are defect free. For functional applications of PBF parts, defects can have detrimental effects on wear and corrosion resistance which have to be avoided for biomedical applications. The complex PBF process involves multitude of physical phenomena, such as thermal energy absorption, reflection, and transfers. Phase transformations such as solid to liquid and then back to solid also occur[17,19].

For low modulus β-Ti alloys that are often metastable, a high cooling rate during the manufacturing process is required to retain the β-phase[19]. As such, PBF is inherently designed for this due to the rapid solidification and cooling cycles that occur in the process[20]. However, many defects such as porosity, balling, oxide inclusions, and cracking remain as metallurgical challenges for PBF. The details of these defects and their forming mechanisms have been discussed in recent reviews[21,22]. To minimize these defects in PBF parts, the effect of process parameters on the parts quality has been studied extensively. The commonly investigated L-PBF process parameters include laser power (P), scanning speed (v), hatch spacing (h), and layer thickness (d) and they are often discussed using one equation:

\[ \varepsilon = \frac{P}{v \cdot h \cdot d} \]  (1)

where, \( \varepsilon \) is termed as the volumetric energy density[23,24].

For EB-PBF, the key process parameters include acceleration voltage (V), beam current (I), scanning speed (v), hatch spacing (h), and layer thickness (d). They are also often discussed using one equation:

\[ \varepsilon = \frac{VI}{v \cdot h \cdot d} \]  (2)

where, \( \varepsilon \) is also the volumetric energy density[25,26].

Equation (ii) can be expressed as Equation (i) in which \( P = VI \) where P is the beam power[27,28].

As an example to elucidate the process parameters effect on β-Ti alloys, the fabrication of Ti53Nb using
volumetric energy densities from 16 to 317 J/mm³ was investigated ⁴⁴. The porosity present in the samples is formed due to incomplete fusion and overmelting. As observed for laser power of 100 W and 200 W, increasing scanning speed led to increasing lack of fusion pores ⁴⁴. It is common to have pores formation as a result of trapped gas or vapor during overmelting. In another study done, also using Ti53Nb (wt%), volumetric energy density between 32 and 95 J/mm³ was used, relative density of the samples ranged from 87% to above 99% ⁴⁵. Due to the high niobium content, the samples maintained the β-phase with no phase transformation ⁴⁶.

2.2. Properties of β-titanium alloys

Research on AM of β-Ti alloys confirmed the appearance of complex martensitic phases, the transformation of martensitic phases, and the improvement of mechanical properties ⁴⁷. In a study on Ti45Nb (wt%), X-ray diffraction patterns on the L-PBF part show β-Ti peaks with broadening characteristics possibly due to residual stresses in the parts ⁴⁸. The samples showed a compressive strength of 723 MPa. Another study was also done on Ti42Nb (wt%) ⁴⁹. In the work done on Ti53Nb (wt%), metastable β-phase is also reported ⁵⁰. A study done using Ti15Mo3Nb3Al0.2Si (wt%) showed the as-built parts exhibited columnar grains of β-phase oriented along the build direction. The samples also have microhardness of 278 HV, yield strength of 917 MPa, ultimate tensile strength of 946 MPa with ductility of 25.3% ⁵¹. A study done on Ti52Nb (wt%) shows that the scanning speed and laser power, or potentially any process parameters, have effect on the elastic modulus of the material. One of the parameters set obtained Young’s modulus of 70.5 ± 1.5 GPa ⁵². Schwab et al. studied the fabrication of Ti5Al5V5Mo3Mo (wt%) using L-PBF and were able to achieve relative density of 99.5%. The samples also showed an ultimate tensile strength of 800 MPa and maximum elongation of 14% ⁵³. In another study using Ti25Nb3Zr3Mo2Sn, the samples have ultimate tensile strength of 716 ± 14 MPa and ductility of 37 ± 5% ⁵⁴. Using Ti24Nb4Zr8Sn, L-PBF samples have Young’s modulus of 53 ± 1 GPa, ultimate tensile strength of 665 ± 18 MPa, yield strength of 563 ± 38 MPa, and elongation of 13.8 ± 4.1% ⁵⁵. In a study using Ti35Nb7Zr5Ta (wt%), relative density of more than 99.8% was obtained. The alloy exhibits an ultimate tensile strength of 631 MPa while having a low modulus of approximately 81 GPa ⁵⁶. The same alloy was also fabricated using EB-PBF and has a modulus of 92 GPa ⁵⁷. Wang et al. studied the fabrication of Ti24Nb4Zr8Sn using EB-PBF, however, no mechanical characterizations were conducted ⁵⁸. Sun et al. studied the fabrication of Ti15Mo5Zr3Al using both L-PBF and EB-PBF. It is shown that the L-PBF samples have higher ductility but lower strength when compared to the EB-PBF samples ⁵⁹.

3. Potential research for β-titanium alloys by PBF

3.1. In situ alloying

In situ alloying using PBF, especially L-PBF, which uses powder mixtures instead of the typical pre-alloyed powders, has been popular to achieve β-Ti alloys. The past studies showed that this approach can be used to obtain β-Ti alloys with desirable properties ⁶⁰-⁶¹. For example, Wang et al. used in situ alloying with L-PBF and found that the niobium addition into titanium leads to better in vitro apatite forming capability as compared to pure titanium ⁶². In a similar study done using in situ alloying of Ti25Nb (wt%), better cell spread and proliferation are observed when the alloy is benchmarked with titanium. The β-Ti alloy also has superior in vitro apatite forming capability ⁶³. Using EB-PBF, in situ alloying of Ti10Nb (at%) was done. It was concluded that β-phase is dominant in the resulting alloy ⁶⁴. Comprehensive reviews on in situ alloying have been conducted previously ⁶⁵-⁶⁶. Comparison between typical PBF and in situ alloying is shown in Figure 2.

3.2. Designed porosity for β-titanium alloys

There is a limitation to the minimum elastic modulus that can be achieved just by developing new materials. As such, an alternative solution to achieve modulus matching between the implants and bone is by introducing designed and controlled porosity to the materials. Porous lattice structures can be fabricated by L-PBF due to its capability for freeform fabrication ⁶⁷-⁶⁸. Coupling these structures with suitable alloys such as the β-Ti alloys, AM can potentially produce biomedical implants that meet the stiffness and strength criteria while achieving excellent osseointegration. In their work, Hafeez et al. fabricated porous structures with different pore dimensions using Ti35Nb2Ta3Zr by L-PBF. They maintained the porosity for these structures at around 50% and are able to obtain modulus of approximately 3.1 GPa, which is quite close to the modulus of bones ⁶⁹. Porous lattice structures were...
also fabricated using Ti24Nb4Zr8Sn and the samples have Young’s modulus of 0.95 ± 0.05 GPa\cite{50}. Porous lattice structures are also fabricated using β-phase Ti24Nb4Zr8Sn\cite{51}. Ti35Zr28Nb was used to fabricate porous lattice structures with FCCZ and FBCCZ unit cells. While the bulk material elastic modulus is between 57.0 ± 1.8% GPa, the porous lattice structures are able to achieve elastic modulus of between 1.1 ± 0.1% GPa and 1.3 ± 0.1% GPa\cite{52}. It was also concluded that the structures have good corrosion behavior and biocompatibility. Using Ti25Nb3Zr3Mo2Sn to fabricate porous lattice structures by L-PBF, Liu et al. found out that martensitic phase transformation takes place during yielding under compression loading for the samples, which prevents the samples from layer-wise fracture\cite{53}. Such phase transformation is also found to improve the tensile strength of β-Ti alloys\cite{54}. Using EB-PBF, Liu et al. fabricated porous structures made of Ti24Nb4Zr8Sn and compared the effect of different designed porosities\cite{55}.

3.3. Machine learning and artificial intelligence

The fluctuations in L-PBF make process control and prediction challenging. While physical simulations have been developed to provide deeper understanding of the process, it is still challenging for them to show the full aspects of L-PBF. Machine learning and artificial intelligence can be used to enhance and complement simulations in achieving higher part quality\cite{56}. The current practice to obtain defect-free parts, or commonly known as the process parameters optimization, involves multiple rounds of experiments which are costly and time intensive. Machine learning techniques potentially can be applied to establish relationships between the input and output of L-PBF. Smaller data sets can be used to develop and train the machine learning algorithms by measuring the part quality from different process parameters. Predictive models using genetic programming and neural network modeling for process planning purposes have been developed for L-PBF\cite{57}. Deep neural networks can also be used to interpret and classify melt pool images that can be used to predict the defects during the process\cite{58,59}. Shin et al. developed an artificial neural network to identify the optimum L-PBF process parameters to obtain defect-free Ti5Al5V5Mo3Cr (wt\%) and showed the potential of using this approach to reduce processing time and making the L-PBF process more cost effective\cite{60}. A conceptual framework on using machine learning to predict manufacturability for PBF is shown in Figure 3.

Comprehensive reviews on machine learning for AM are available\cite{56,61-63}. Using data-driven artificial intelligence, it is possible to predict and optimize process parameters that can obtain desired part properties\cite{64,65}. Furthermore, a conceptual framework that combines statistical analysis, mathematical modeling, and machine learning techniques has also been proposed\cite{66}.

4. Summary

In this perspective article, the feasibility of processing β-Ti alloys using PBF has been shown using currently available literature. It has been shown that these alloys have potential to be better candidates for biomedical implants due to their lower elastic modulus and high mechanical strength. However, to realize their full potential, future research in the area of new processing approaches and new designs are still needed.
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Conflict of interest
The author declared no known conflict of interest.

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Additive Manufactured Beta-Titanium Alloys

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