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A COMPARATIVE STUDY ON SELECTIVE LASER MELTING AND ELECTRON BEAM MELTING PROCESS FOR ORTHOPEDIC IMPLANTS

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ABSTRACT: Selective laser melting and electron beam melting are two metal additive manufacturing processes which gained considerable attention in the recent years. These techniques allow large degree of freedom in design such that complicated geometries can be manufactured with higher accuracy and efficient use of materials. Due to its versatility, these processes find wide applications in aerospace and biomedical industry. In this review article, a comparison of both the processes with emphasis on orthopedic implants has been carried out. The comparison is made based on titanium alloy, Ti6Al4V, which is the work horse material for manufacturing orthopedic implants. Focus is given on the microstructural and mechanical property differences and how these differences influence the oseointegration and long term performance of the implants.

INTRODUCTION

Metal additive manufacturing (AM) process such as Selective Laser Melting (SLM) and Electron Beam Melting (EBM) are developed from the basic concepts of the well-established polymer based additive manufacturing processes. The advent of high energy intensity sources based on high power lasers and electron beam enabled the melting of metallic materials having high melting points. This paved way for the development of metal based additive manufacturing processes. Both SLM and EBM are powder bed fusion based process capable of producing metallic parts directly from metal alloy powders. SLM utilizes a fiber laser heat source with power ranging from 100 to 500W. The laser beam selectively melts the powder layer based on the computer aided design (CAD) model of the desired part. Details of SLM process can be found in Ref.1. In EBM, an electron beam is generated by applying a higher voltage of 60kV to a tungsten filament in an electron gun. The electron beam thus generated is focused to the powder bed and selectively melts the powder to obtain the desired part. The vacuum system in EBM provides an excellent environment for processing reactive materials like Ti and its alloys. Details of the EBM process can be found in Ref.2.

Metal additive manufacturing processes are more suitable for applications which demand the fabrication of complicated, low volume and customized components. All the orthopedic bio-implants fall in this category. Customized implants are conventionally manufactured using CNC machining from a bar stock followed by machining and grinding. This causes significant material wastage [3]. Other than material wastage one of the key issues is the property mismatch between the implant and the actual bone. The modulus or the stiffness of the bone is less than that of the modulus of an implant machined out from a bar stock. This mechanical mismatch between the machined out prosthetic stem and the patient's bone results in a phenomenon called stress shielding. Stress shielding causes bone resorption and may result in premature loosening of the implant due to insufficient loading of the bone [4]. A bone implant with porous structure can overcome this modulus mismatch. The porous structure also enhances the tissue/bone integration. So far many techniques have been implemented to generate metallic porous structures, including

powder sintering approach, space holder method, combustion synthesis, plasma spraying, and polymeric sponge replication [5,6]. All these processes have inherent limitations such as lack of control on pore size, pore shape and distribution of the pores. These limitations can be very well taken cared by the AM processes.

Many researchers have produced porous structured bioimplants with controlled porosity, different range of pore sizes and homogenous distribution of the pore using SLM [7,8] and EBM [9,10].

In this paper, SLM and EBM additive manufacturing process are compared with focus on the performance aspects of Ti6Al4V implants manufactured for orthopedic applications.

COMPARISON OF SLM AND EBM PROCESSES

Though SLM and EBM follow the fundamental principles of layer-by-layer manufacturing, the processing conditions are entirely different. The laser beam and electron beam interactions with metallic powder is quite different. Large amount of energy is lost due to reflection when the laser beam hits metallic powder. Laser radiation penetrates into powder through pores to a depth of several particle diameters because of multiple reflections. Laser radiation is scattered inside the powder layer and its absorption gives raise a volumetric heat source. When the laser power become high enough the powder starts melting [11]. The penetration depth of an electron beam into the irradiated material is multiple times greater than it is with a laser beam [12], which means the energy supplied is almost fully utilized. When the high speed electron beam hits the powder layer, kinetic energy is converted into thermal energy, causing the powder to melt. In SLM, the build chamber attains a temperature of only around 100 $^{\circ}$C during processing where as in EBM the build chamber temperature is approximately at 650 $^{\circ}$C. These differences in processing conditions essentially mean that there would be a considerable difference in the properties of the parts made out of these two processes. The differences are very obvious while processing Ti6Al4V alloy, which is the work horse material for orthopedic implant manufacturing.

Surface characteristics

Surface topography is one of the factors which determine the performance of orthopedic implants. The tissue/bone integration or oseointegration is greatly influenced by the surface condition of the implant. By having certain degree of surface roughness the oseointegration can be improved. Two conventional methods used to improve oseointegration process are by applying coatings on the surface of the implants and by chemical treatment of the implant surface. Plasma spraying of hydroxyapetite is the most commonly used coating technique. For SLM and EBM fabricated parts there is an inherent surface roughness as a result of the layer-by-layer addition of material. Also, surfaces with different roughness can be generated on the implants such that a post-processing could be avoided. By changing the process parameter settings the surface roughness can be varied between 1-20 micrometers [13]. Implants produced using SLM have a smoother surface finish when compared to EBM. This is because of the difference in powder particle size and the layer thickness. In SLM the powder particle size range is 25 to 50 microns and the layer thickness is 50 microns. For EBM the particle size ranges from 50 to 100 microns and the layer thickness is 100 microns [14]. Fig.1 shows a comparison of surfaces generated by SLM and EBM of Ti6Al4V using standard parameters provided by the machine manufacturers. A comparative study on bone in-growth potential of both electron beam and laser beam fabricated trabecular bone implant with porous structure has been studied by Biemond et al. [15]. The original design of the trabecular implant surface was similar for both production techniques. However, the SLM and EBM specimens that resulted after production were different in gross morphology and surface texture because of the differences in processing conditions. Based on all histological and push-out test

results they found that there are no large differences in in-growth potential between the trabecular bone implants made by the two processes.
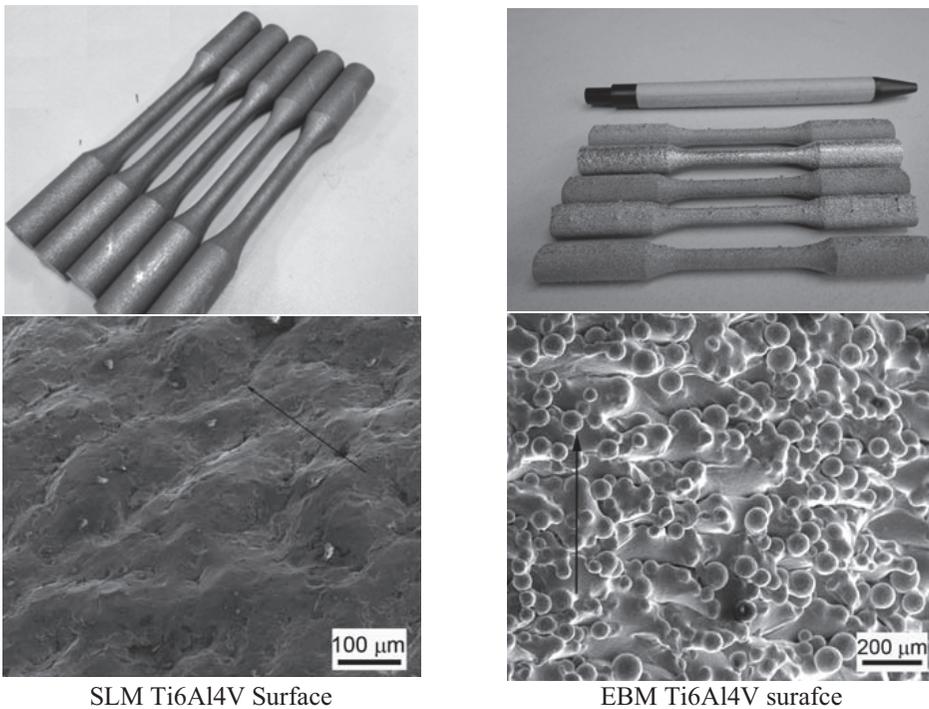

SLM Ti6Al4V Surface   EBM Ti6Al4V surafce

Fig.1 Comparison of surface characteristics between SLM and EBM produced Ti6Al4V samples. (arrow indicates build direction) [14]

Microstructural characteristics

The microstructure of Ti6Al4V obtained from SLM and EBM are quite different from the wrought or cast microstructures. Between SLM and EBM the Ti6Al4V microstructure differs in the phases present due to the difference in processing conditions. All the reported studies on SLM of Ti6Al4V show the final microstructure as completely martensitic (α'). Thijis et al. studied the evolution of Ti6Al4V microstructure in SLM. The short interaction time, high temperature gradients and high localization of the SLM process were attributed to the formation of complete martensitic phase. The grains were elongated on the direction of heat transfer and due to the partial remelting of the previous layer, the grains grown to several hundred micrometers [16]. Similar studies were carried out by Murr et al. [17] and Song et al.[18]. The microstructure observed in EBM processed Ti6Al4V was predominantly α with a small amount of retained β phase [19]. Martensitic phase was not observed. This is because during the entire processing the build chamber was at a temperature of around 650 °C. This temperature is above the martensitic start temperature (Ms) for Ti6Al4V. Once the processing is completed the entire build is cooled slowly to the room temperature which results in the formation of α rather than α' phase. Typical microstructures of SLM and EBM processed Ti6Al4V is shown in Fig.2. This difference in microstructures can have significant influence on the performance of the implants in terms of longevity.
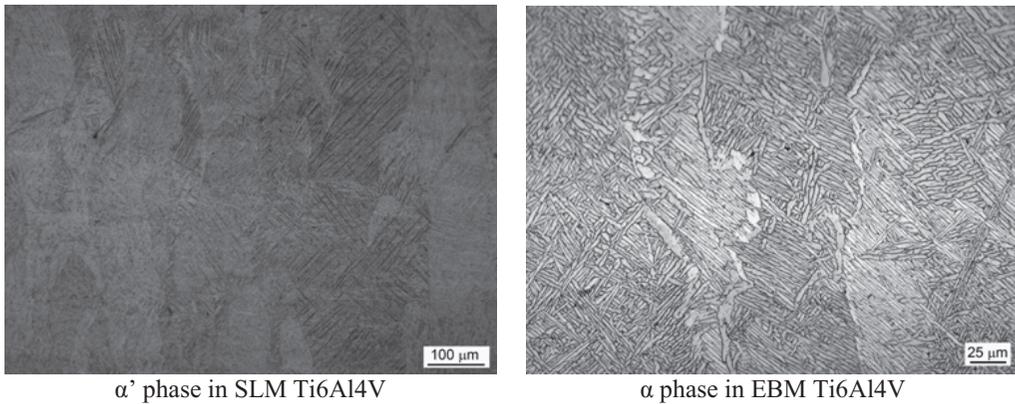
α' phase in SLM Ti6Al4V   α phase in EBM Ti6Al4V

Fig. 2. Optical micrograph of SLM and EBM produced Ti6Al4V samples [14]

Mechanical properties

One of the important reasons for considering Ti6Al4V alloy for implant manufacturing is that the elastic modulus (110 GPa) is almost half when compared to stainless steel and CoCr [20]. Other considerations in mechanical properties are ductility, tensile strength, compressive, fatigue resistance and fracture toughness. Many researchers have studied the mechanical properties of Ti6Al4V fabricated both in SLM [18, 21] and EBM [17,19] processes. Analysis of results from literatures shows a range of tensile properties which is comparable with the wrought or cast material in annealed condition. Typical range of tensile properties is shown in Table 1. The tensile properties are better for SLM produced parts as compared to EBM parts due to the presence of martensitic phase. The fatigue strength is also higher for SLM produced parts (550 MPa) due to the microstructural differences.

Material type	Yield strength YS (MPa)	Ultimate tensile strength, UTS (MPa)	% elongation
SLM Ti6Al4V	1050 -1150	1175 - 1250	5-8
EBM Ti6Al4V	890-950	970 - 1030	10-14
ASM handbook (cast and annealed)	885	930	

Table 1. Range of tensile properties for SLM and EBM produced Ti6Al4V parts

Most of the studies on Ti6Al4V orthopedic implants produced by SLM and EBM concentrated mainly on the oseointegration and the strength aspects. Noticeable differences were not found between SLM and EBM implants on oseointegration properties with experiments conducted for relatively short durations. Further investigations are needed to ascertain long term oseointegration performance. The suitability of metal additive manufacturing for bioimplants is justified by the comparable tensile properties with the conventionally fabricated parts. However, for orthopedic applications, fatigue and fracture toughness of the implants are equally important or even more than the tensile strength. But detailed studies on these areas are very limited in open literature. Considering the differences in microstructure of Ti6Al4V processed in SLM and EBM, studies on fatigue and fracture aspects got lot of significance. Microstructure of the alloy is one of the

important factors controlling its tensile properties, fatigue strength and fracture toughness. Depending upon the processing conditions, the alloy Ti6Al4V can have different forms of microstructures such as equiaxed, bimodal, and lamellar/columnar. Both SLM and EBM produce columnar microstructures. For SLM, the presence of martensitic phase enhances the strength of the alloy. But, in terms of fracture toughness, the hard martensitic phase may not be ideal. The fine acicular martensitic phase results in more brittle behavior and inferior fracture toughness [22]. In EBM, the resulting phase is α, which is softer compared to martensite. For a columnar microstructure with α phase, the fracture toughness of the alloy is controlled by the size of colonies of lamellar α-phase and thickness of α-lamellae .Colonies of α-phase lamellae with various orientations hinder the crack growth [23]. Microstructure of SLM produced Ti6AL4V can be further altered by post-heat treatment to enhance the fracture toughness.

SUMMARY

In this short review article the property requirements for biomedical implants manufactured by SLM and EBM process have been addressed. The properties of the parts made by these two processes differ, particularly for Ti6Al4V alloy, because of the difference in processing conditions. This is a matter of concern when considering the long term performance of orthopedic implants. This can be addressed by proper post-heat treatment, though it adds another processing step. However, the most significant advantage of these processes in biomedical application is to generate porous structures with controlled architecture. This enables the manufacturing of orthopedic implants which can match exactly with the morphology and properties of actual bone.

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