BÁNKI KÖZLEMÉNYEK
3. ÉVFOLYAM 1. SZÁM

Development of Selective Laser Melting of Ti6Al4V Alloy for Tissue Engineering: Review

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***Abstract* — 3D printing, particularly Selective Laser Melting (SLM), is an important technology in biomedical engineering applications. The recently developed Ti alloys for biomedical applications are followed by the challenges of understanding the phase transformations and mechanical properties during the SLM cycles. The present paper reviews the fundamental understanding of SLM thermal technology influencing the microstructure evolution of Ti6Al4V alloy. The focus is on the effect of parameters of SLM on microstructure and mechanical performance of Ti6Al4V alloy. In addition, review the most important problems and solutions.**

***Keywords*:** Selective Laser Melting; Ti-6Al-4V; phase transformations; Ti6Al4V.

1 INTRODUCTION

According to ASTM [1], the 3D printing is described as the processes of building three-dimensional parts by addition and joining the powder of material layer by layer based on a CAD file. currently, there are six types of 3D printing on the basis of the technique used:

1. Selective Laser Melting (SLM),
2. Digital Light Processing (DLP),
3. Ultrasonic Consolidation (UC),
4. Deposition Modeling Technology (FDM),
5. Electron Beam Melting (EBM), and
6. Selective Laser Sintering (SLS)

Selective laser melting (SLM) is one of the most common 3D printing process which is currently used in biomedical applications especially implants [2,3]. The most fundamental characteristics of SLM technique are

1. extra degree of freedom in complex design structure,
2. high production speed,
3. high precision.

Biocompatibility of implants, which is defined as the material's capacity to carried out with a suitable host response. The most suitable material for most implants in the human body is metallic alloys including Ti-alloys, CoCrMo, and stainless steel. Due to superior biocompatibility, excellent corrosion resistance, and unlike Co–Cr or stainless steel, does not cause hypersensitivity, make titanium alloys, particularly Ti6Al4V useful for medical engineering applications [4].

One of the key problems to be to be solved for SLM of Ti alloys is the influence of parameters of SLM on the energy density of laser beam that effects on the quality of the final part. Whereas, a too low energy density will not be able to generate adequate heat energy and leading to lack of fusion porosity and balling [5]. A too high energy density also is not advisable as it results in keyholing defects from the melting materials [4]. Keyholing reasons powder vaporisation and produces circular porosity throughout the construction. For example, low energy density can be produced from large layer thickness, high scanning speed, and low laser power. In addition, high energy density can be produced from high laser and low scanning speed. Therefore, it is needed to adjust SLM parameters so that the balling and keyholing defects are not guaranteed.

2 SELECTIVE LASER MELTING (SLM)

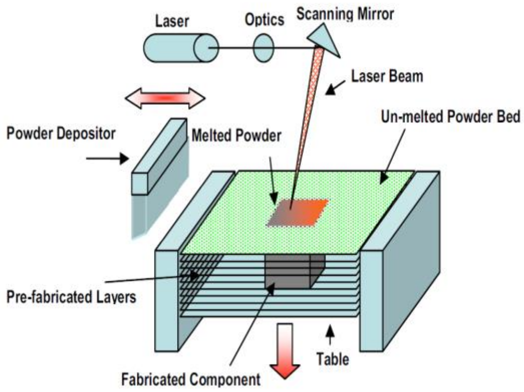
In SLM, the powder material is heated and melting in which heat is generated via laser beam, continually at a wavelength of 1075 nm by Yb: YAG or Nd: YAG crystal and rapid solidification of the material melted into the desired component. The SLM process is composed of a sequence of steps [6], starts from

i) 3D CAD file data,
ii) Stereo Lithography (STL) files,
iii) Slicing the layers,
iv) Construct the part in a layer by layer

The construction method for SLMs as shown in figure 1 begins with the use of a powder depositor to deposit a thin layer of metal powder as a layer thickness onto the top of the metal substrate plate. Using a high-energy laser after layer deposition, selected areas are melted and fused using the CAD data processed. After a single layer is scanned by laser, the substrate plate is lowered by a quantity equal to the thickness of layer and a new layer is deposited upon the previous layer using powder depositor. The SLM technology is replicated layer after layer till the needed product has been completed.

Titanium is highly reactive and sensitive to oxygen, carbon, hydrogen, and nitrogen in liquid state, and even when heated at temperatures above 650 ° C in the air [6].

The Ti alloys are likely to dissolve discreet quantities of oxygen and nitrogen into solution at high temperatures. For these purposes the titanium SLM needs protection against oxygen and nitrogen, such as argon inert gas to prevent embrittlement and contamination, even more than other metals like aluminum or iron.



1. Figure: Schematic diagram of Selective Laser Melting (SLM) [7].

The SLM is a complex process in which the metallurgical and mechanical performance of the final product are influenced by various parameters as shown in figure 2. As mentioned above, there is a minimum energy density of laser beam to ensure that no balling and maximum energy density of laser beam to ensure that no keyholling. Therefore, the most important SLM parameters, namely laser power, scan speed, hatch spacing and layer thickness, which determine energy density, should be adjusted so that the sufficient energy density is obtained. The physical principal for achieving this energy density (E) to a volumetric unit of powder material is expressed by the equation [8].

$$E_{density} = \frac{P_{laser}}{V_{scan}\, S_{hatch}\, t_{layer}} \qquad \mathrm{J/mm^3}$$

where

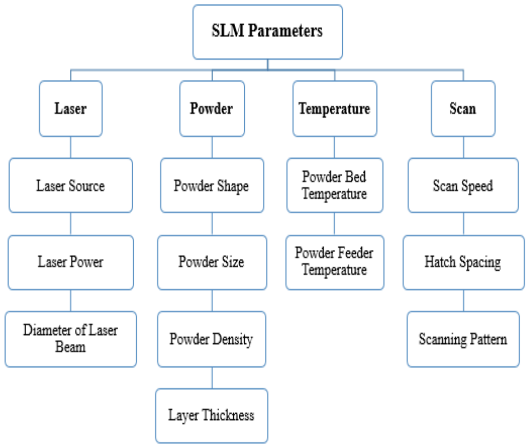
P = laser power (W),

v = scan speed (mm/s),

h = hatch spacing (mm) and

t = layer thickness (mm).

The above equations emphasize that the density of energy is strongly dependent on the layer thickness, incident laser power, hatch spacing, and scan speed. In various studies [2,6], effects on the microstructure and mechanical performance of 3D printed Ti6Al4V were studied to optimize the process, including layer thickness, laser power, hatch spacing, and scan speed.



2. Figure: Process parameters in SLM process [9].

3 TITANIUM ALLOYS

The most suitable material for implants in human budy are metallic alloys (stainless steel , CoCrMos or Ti-alloys) [3]. Because of continued increased requirements for biocompatibility in human hard tissue, such as dental and bone, Ti alloys are increasing rapidly in dental and biomedical applications. Furthermore, the relatively low conductivity of titanium (8.41 µm/m/°C at 20°C for Ti)(6.7 W/m.K for Ti6Al4V) and thermal expansion coefficient decreases the possibility of deformation caused by SLM technology opposed to other widespread implanted metals (21.4 W/m.K for stainless steel 316) [9]. In contrast with 316LSS (210 GPa) and Co-Cr (240 GPa), titanium alloys have much lower modulus of elasticity (55-110 GPa) [10]. Depending on its microstructure, the Ti alloys can be classified into four separate groups: commercially pure, Alpha and Alpha, Alpha-beta and Metastable-Beta alloys [11].

(i) Commercially pure: Pure titanium exhibits an allotropic phase change from the close-packed hexagonal (HCP) alpha (α) phase at low temperatures to the body centered cubic (BCC) beta (β) phase at approximately 882 °C. The commercially pure titanium alloys range in yield strength from approximately 240 MPa to 550 MPa. In this category are the five “unalloyed grades” of ASTM specification, including ASTM grade 1 (99.5%Ti), ASTM 2(99.3%Ti), ASTM 3 (99.2%Ti), ASTM 4 (99.0%Ti), and ASTM 7 (99.4%Ti).

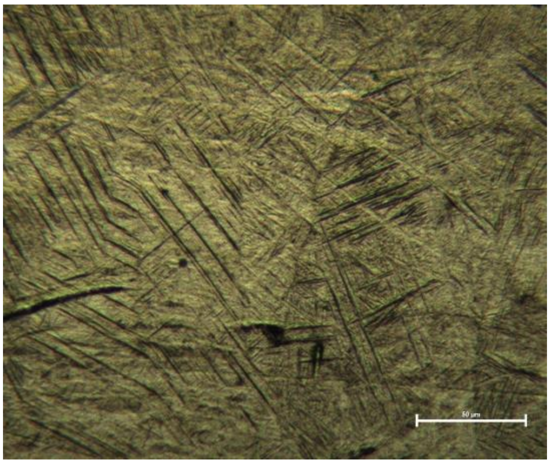
(ii) Alpha and near-alpha that possess primarily alpha phase based microstructures and less than 10% β phase, and in some cases, the microstructure is entirely alpha phase. Alpha and near-alpha grades such as Ti-6Al-2Sn-4Zr-2Mo, and Ti-1100.

(iii) Alpha-beta titanium alloys consist of a mixture of alpha and beta phase. Ti-6Al-4V is one of the most common alpha-beta titanium alloys which is currently used in biomedical applications.

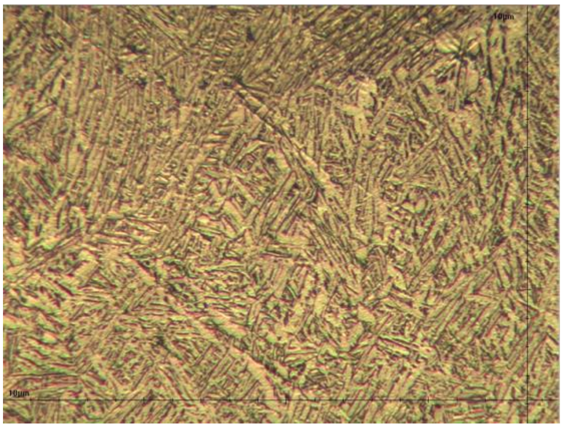
(iv) Metastable-beta titanium alloys, which include Ti-13Nb-13Zr and Ti-30Ta contain high percentages of beta phase stabilizing elements. The microstructure consists from beta phase which can be hardened by formation of alpha platelets during heat treatment. The Young's modulus of these alloys is less than 70 GPa.

Ti-6Al-4V alloy

The microstructure of Ti6Al4V consist of a mixture of close-packed hexagonal (HCP) alpha and body center cubic (BCC) beta phase at room temperature and is currently one of the most widely used titanium alloys [12,13]. The microstructure of SLM printed Ti6Al4V sample is fully α′ martensite (figure 3). After the heat treatment at 850 ℃ for 2 h followed by furnace cooling , the α′ martensite is transformed into α+β microstructure (figure 4). The martensite that occurs as a result of SLM of Ti6Al4V is comparable to that obtained by resistanse spot welding of steel [14,15]. Regarding the effect of adding 6% Al (α stabilizer) and 4% V (β stabilizer), it is interesting to note that beta transus increased to around 995°C [16] compared to the beta transus of pure titanium is 882°C [17]. The modulus of elasticity of Ti6Al4V is 110 GPa. The titanium alloy with a phase isomorphic beta stabilizer graph is shown in Figure 5, which could be used in additive manufacturing as a guide for tracking phase transformation in Ti6Al4V.



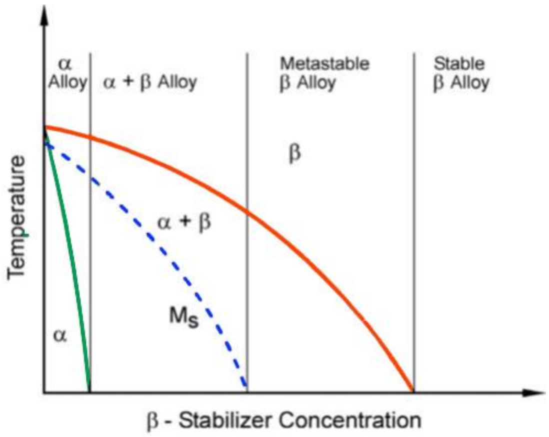
3. Figure: Fully α′ martensite microstructure of Ti6Al4V.



4. Figure: Fully lamellar microstructure of Ti-6Al-4V.

4 SELECTIVE LASER MELTING OF TI6AL4V

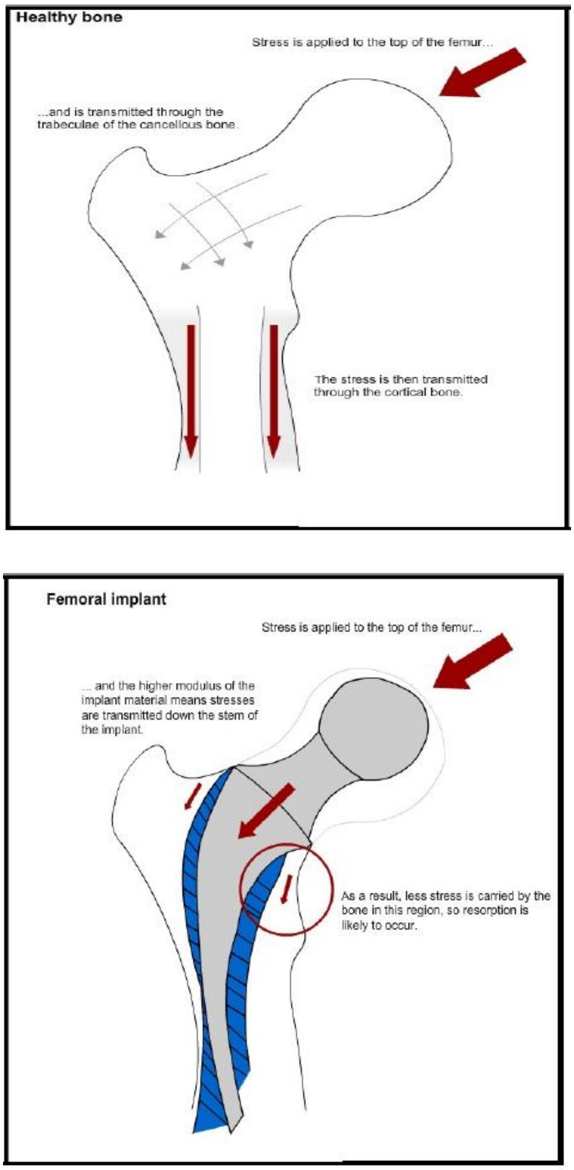
As mentioned above, one of the major problems regarding SLM of materials is their high probability to fail in balling, keyholing, and delamination depending on the parameters of technology. Parameters of the SLM technology effects on the mechanical performance and microstructure of 3D printed parts. Thijs et al.[18] investigated the SLM of of Ti6Al4V. They concluded that at power = 42W, velocity = 200mm/s, hatching spacing = 75 µm, and layer thickness = 30 µm the martensitic phase presented with hardness 410 ± 35 HV.



5. Figure: Pseudo-binary β isomorphous phase diagram [19].

Murr et al.[20], Qiu et al. [21], and Vrancken et al. [22] reported the formation of α' (hcp) martensite phase in the layers of SLM of Ti6Al4V.

One of the most important problems facing the full solid of Ti-6Al-4V is the difference between the Young’s moduli, where Young’s moduli of the metal is 110 GPa and the Young’s moduli of bone is 10 GPa. And this difference leads to non-gradual or irregular transfer of stresses across the human bone/tissue to the metallic implant, which caused stress shielding effect and decrease the life of an implant in vivo as shown in figure 6.



6. Figure: healthy bone and femoral implant after applying stress [23]

The effect of porosity on the elastic Ti6Al4V modulus was investigated by Bandyopadhyay et al.[24]. They demonstrated that the elastic modulus of the porosity of implants between 23-32 vol. percent is near that of the human cortical bone. The modulus of elasticity of implant parts with porosity in the 10-18% range has been observed to be declined in the range of 33-43 GPa in contrast with 248 GPa of solid implants, as discovered by España et al. [25]. And for the purpose of increase load-bearing and bioactive implants, Hao et al. [26] investigated SLM of hydroxyapatite (HAp) and 316L stainless steel (SS) powders mixture. Biemond et al. [27] investigated or implanted the implants of SLM of metal/ hydroxyapatite in the bone of goats and after 15 weeks, a good interfacial bond between tissue of bone and the implant was concluded. The crystallographic and chemical composition of hydroxyapatite (HAp) is the same to minerals of bone, which can promote good bone osseointegration [28–30].

5 Conclusions

In conclusion, SLM become the most important process in biomedical application, appears a relatively difficult process. In brief, this paper reviews the biological, mechanical, and structural mismatches between the Ti6Al4V implants fabricated by SLM and the bone. This work focus the defects manufactured by SLM using multiple combinations of parameters. Also proposes the ideal parameters in the viewpoint of superior mechanical properties and microstructure. This review has highlighted the development and challenges of SLM technology to connection of Ti alloy and bone tissue.

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