SLM-built titanium materials: great potential of developing microstructure and properties for biomedical applications: a review

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Abstract

Selective Laser Melting (SLM) is one of the most important additive manufacturing (AM) techniques used for developing the structures and the properties of different biomaterials such as stainless steel, Co–Cr alloys as well as titanium (Ti) and its alloys. In recent years, manufacturing of SLM-Ti materials for various biomedical applications has received much research attention due to the high engineering value of this advanced process in medicine. By SLM technique, it can be produced highly Ti implants along with unique structures and characteristics; especially, mechanical, tribological and corrosion properties, throughout precise controlling of different effective parameters. This review article aims to provide a comprehensive summary of the experimental data performed recently via the researchers and specialist for developing the mechanical, wear and corrosion properties of SLM-Ti implants. The integration between the microstructural features of SLM-Ti and these vital characteristics is also analyzed and evaluated with a serious endeavor to increase the knowledge about the influence of SLM process and its technical parameters on the structure and the performance of diverse Ti materials used for medical purposes.

1. Introduction

Nowadays, the universal society is suffering from numerous serious troubles regarding terrorism, wars and conflicts, car accidents, new diseases and sport injuries. Furthermore, it is of most important to alter the lifestyle of human beings and enhance the quality of their life. Therefore, these challenges should be resolved throughout developing the field of biomaterials and its technologies. This moved forward the scientist, researchers and engineers to invent new techniques with an aim to improve the required structures and properties of materials used for different biomedical applications. The term ‘bio-manufacturing’ is found out as a novel scientific area to address several biomedical issues. This term has been defined as ‘the use of additive technologies, biodegradable and biocompatible materials, cells and growth factors to produce biological structures for tissue engineering applications’ [1]. Hence, biomanufacturing can be applied as a non-conventional material-process-dependent production method.

For biomedical industry, such as medical implants, dental restorations, etc; the manufacturing processes should be exploited to create materials with highly considerable structures and characteristics. For many decades, medical implants have been fabricated using conventional technologies, such as casting, rolling, forging, machining, and powder metallurgy, with highly consummation of time, energy and materials. Therefore, advanced techniques and methods are becoming an imperative concern in order to overcome all these drawbacks. Recently, laser forming processes have attained a major position in the manufacturing of intricate three dimensional (3D) medical parts without necessity to pre- or post-processing [2, 3].

Additive manufacturing (AM), also known as additive layer manufacturing (ALM), rapid prototyping (RP), solid freeform fabrication, laser forming or laser manufacturing, is a standard term given to cover all manufacturing techniques which build up 3D end-use parts with outstanding characteristics by adding material in layers. ASTM defined an AM as ‘a process of joining materials to make objects from 3D model data, usually
layer upon layer, as opposed to subtractive manufacturing methodologies [4]. The AM techniques are absolutely opposite to conventional processes which depend on a subtractive manufacturing manner [5]. These techniques are increasingly utilized in manufacturing highly geometrical components used for different industries such as aerospace, automobile, jewelry and fashion products as well as medical applications [6]. Hence, these sophisticated techniques have taken so much interest and concentration from the specialists in both industry and academy fields due to their feasibility of fabricating light-weight metallic parts with added functionality directly from design data [7–11].

Selective laser melting (SLM), also known laser beam melting (LBM), is a promising technique in the field of AM as it can process relatively a broad variety of materials powders ranging from metallic to ceramic powders compared to other AM techniques. SLM process has strong capability to produce 3D and near net shape parts with relative high surface finish quality and bulk density using focused and computer-controlled laser scanning of a metallic powders. Moreover, SLM is a unique technology for manufacturing of functionally graded multi-material parts. The process can be employed at all the steps of the product progress, i.e. from design model to low volume production [12].

Metals and their alloys are extensively utilized for orthopedic implants and dental applications. Metallic materials are more appropriate for load-bearing applications in comparison to other engineering materials like ceramics or polymers because of their unique mechanical characteristics such as high strength and fracture toughness. Among various metallic biomaterials, Ti and its alloys are remarkably crucial biomaterials for various medical applications due to their excellent and unique properties like light weight, mechanical performance, corrosion behavior and biocompatibility [13–17]. It is well known that Ti based biomaterials were widely processed by traditional techniques as well as thermal and thermo-mechanical treatments. On the other hand, manufacturing of Ti and its alloys using advanced technologies like SLM is an urgent demand in order to overcome the drawbacks of the conventional processes. Therefore, many serious attempts and researches were performed to develop and improve the performance of different kinds of materials especially Ti and its alloys by SLM. Most scientific works were mainly focused on design of processing parameters, microstructure evolution along with material characteristics like mechanical strength [18], wear [19] and corrosion [20]. Recently, a comprehensive review on the dissimilarities in microstructure and properties of metallic components produced by SLM and other traditional processes has been introduced [21].

Depending upon our knowledge, very limited articles reviewed the importance of SLM process in manufacturing of implants using an advanced approach of SLM. Recently, some review articles presented some issues related to SLM-manufacturing of Ti materials [22, 23]. This article is a comprehensive review on the manufacturing of different biomedical Ti materials. The article addresses several vital issues related to biomedical Ti materials fabricated by SLM technology such as microstructure, characteristics and also applications. Based on the chosen parameters of SLM, the integration between the micro-structural features and the required characteristics of Ti materials is discussed in details. Moreover, in this work, comprehensive literature review on the current status and development of SLM-built Ti materials employed mostly for biomedical applications is presented. It is a serious attempt to increase the scientific knowledge for specialists from academicians, researchers and practitioners by compiling significantly important studies and investigations pertaining to SLM.

2. SLM process: fundamentals and its important issues

Through many decades, AM was employed to fabricate physical models in layer-by-layer style using 3D solid models constructed by computer aided design (CAD). These laser-based layer processes are able to fabricate many metallic parts of high density and better mechanical properties in short time [9]. Furthermore, AM suggests the reliability of fabricating many components with various geometries and compositions in short production runs using the same machine [24]. Recently, it has been published a comprehensive reviews with focused details on AM of metallic components fabricated by different techniques [25, 26].

As is well known that selective laser sintering (SLS) and SLM are considered as very important AM techniques for constructing several functionally parts with diverse structures and properties using high-energy light sources [27–30]. SLM is a sophisticated technology of SLS process as the entire melting of powder occurs rather than sintering or partial melting [31]. On the contrary to SLS, SLM does not require an incorporation of manufacturing post processes, such as milling and turning, to realize fully dense parts [32]. The process is also free of binders and fluxing agents [33]. It is very important to mention here that SLM permits the rapid fabrication of complex-shaped 3D components in bulk/porous designs directly from powders, particularly metallic powders [34–36]. Hence, the geometric limitations associated with the traditional fabrication processes have been overcome by using SLM. Moreover, comparing with another processes of AM, the highly vacuum system and preheating treatment of the powder are not required throughout the entire SLM process which is in
Table 1. Major benefits and drawbacks of SLM.

| Results | Materials | Manufacturing | Component quality | Economical situation |
|---------|-----------|---------------|-------------------|---------------------|
| Ad.     | Processing ability of hard materials (high melting point materials), No distinct binder and melt phases | Reduction of production steps, a high level of flexibility, and a high material use efficiency | Suitable for fabricating: fully dense parts in a direct way, high geometrical complexity parts, and a near net shape production. | Reduction or elimination of costly and time-consuming post-heat treatments. |
| Disad.  | Unsuitable for well-controlled composite materials (e.g. WC-Co) | Higher thermal stresses owing to large thermal gradients | Segregation phenomena and formation of non-equilibrium phases due to rapid solidification, a higher level of porosity and an increased surface roughness due to melt pool instabilities as a result of non-optimal scan parameters may cause | high laser power and high-quality beam (high cost machine), extensive build times due to lower scanning velocity. |
contrast to electron beam melting (EBM) process [37, 38]. Table 1 reviews the leading benefits and drawbacks of SLM [39].

In SLM process, a scanning of higher energy laser beams controlled by computer is applied for extremely melting fine metallic powder layers (25–50 μm) with an aim to obtain rapid fabrication of 3D structures of bulk or porous parts depending initially upon CAD files [3, 39–41]. The logical sequence of this superior technology is based on the selectively melting and consolidating of the irradiated particles, under a protective atmosphere, which directly solidify after cooling to form a full solid layer [33, 42]. Consequently, after the consolidation of one cross section layer, a new scanned solid layer of powder is deposited on the surface of the previously built layer, i.e. successive layers of powder are produced, and the process continues until final achievement of fully dense part [43]. The freeze powders that is not melted during the process can be recycled and used again after removing them from the completed parts. Furthermore, supports are necessary to anchor down certain unsupported features owing to shrinkage and bending of solidifying material. This restricts the process geometric freedom and incurs further post-processing procedures to remove supports.

Several physical phenomena are associated with SLM such as absorption and scattering of laser energy, heat transfer, phase transformation, liquid flow within the molten pool caused by surface-tension gradient, evaporation and emission of material as well as chemical reactions [44]. Moreover, regardless of the type of the material used, a considerable refinement of the microstructures of SLM parts is usually achieved as a results of different particular metallurgical phenomenon occurred throughout SLM like overlapping of the melt pools, layer by layer stacking, directional heat transfer, fast laser moving to obtain rapid solidification and large temperature gradient. However, during the layer-by-layer SLM method, non-equilibrium microstructures with lack-of-fusion defects formed at the interfaces between each laser pass and layer could be attained [30, 45]. As a result, it is very hard to produce fully dense or defect-free parts and many micro-structural defects could be created such as micro pores, cracks, and residual stress as well [21]. In addition, it was found that the cooling rate after solution treatment plays a major role in determining the microstructure of Ti alloys as martensite α’ phase will be the main phase when the heat treatment achieved at cooling rate higher than 525 K s\(^{-1}\) [46]. Therefore, in SLM process, β phase is entirely converted into acicular α’ martensite rather than α phase since the processed Ti materials experience an extremely high cooling rate in the order of 106 K s\(^{-1}\) [47, 48]. As a result from above data, SLM can be considered as a complex physicochemical metallurgy technique. It is important to mention here that many researchers have employed some techniques, such as hot isostatic pressing and post-heat treatment, for closing pores and homogenizing micro-structural features, respectively [18, 32, 36, 49]. Moreover, it is pointed out that the appropriate combination of the process parameters throughout an intense adjustment and controlling them, such as laser power, laser scan speed, hatch distance, hatch style, layer thickness and laser spot size, is a significant way to fabricate defect-free parts [50].

It was revealed that SLM has an edge over EBM, the most important powder based fusion process along with SLM, as SLM has a great ability to produce a wide range of engineering materials in comparison to the diversity of materials fabricated by EBM [51]. The SLM process has been used to fabricate parts from various engineering materials such as Fe [52], Ni [53–55], Cr–Ni [56], stainless steel [57–61], Al [11, 62, 63], CoCrMo [64, 65], Ti [34, 66–69] and composites as well [70–73]. Furthermore, different industrial sectors and fields are based on SLM process especially aerospace and biomedical applications [74]. It can be found more valued details on the SLM process from different references [35, 67, 75, 76].

It is well known that there are a large number of processing parameters control SLM technology such as laser power, laser scan speed, scan line spacing (hatch distance), thickness of layer, scanning strategy, working atmosphere, temperature of powder used and material–based input parameters. The accurate optimization of all these parameters could be performed by taking many considerations such as the part’s properties required, i.e. microstructure, mechanical, tribological, surface finish, etc. as well as the production time and cost. Nevertheless, the most important goal of SLM process is the fabrication of fully dense parts with higher performance. It is very necessary to scientifically understand the adjustment of all above processing parameters affected [44]. Therefore, the design of SLM processing parameters is an essential function for controlling the structures and the properties of built parts. For example, the laser energy density (E) depends mainly on the major processing parameters, incident laser power (P) and laser scan speed (v); E = \( P / v \) [47]. Furthermore, the volume energy (EV) supplied to the powder layer can be calculated depending upon laser power (PL), scan speed (Vs), hatch distance (HS) and layer thickness (DS); \( EV = PL \times VS \times HS \times DS \) [26]. Therefore, increasing the laser energy density leads to increase the temperature of the powder which means larger melting area and higher final density. However, using very slow speeds guides to decrease the density as a result of the formation of balling and dross in the melt pool, leading to a reduced surface roughness and lower density [39]. Song et al [77] proved that the microstructure, roughness, densification and micro-hardness of SLM-built Ti6Al4V parts are mainly dependent on the processing parameters. The results indicated that exceptional parts, in great micro-hardness (450 HV), smooth surface (2.1 μm) and higher density (4.13 g cm\(^{-3}\)), can be fabricated via SLM using preferable laser power 110 W and scanning speed 0.4 m s\(^{-1}\) that provide continuous melting mechanism.
3. SLM for different biomedical Ti materials

Today, Ti and its alloys are extensively employed in various biomedical applications such as joint replacements, bone plates and screws as well as dental root implants owing to their outstanding characteristics such as higher strength-to-weight ratio, superior corrosion resistance and exceptional biocompatibility [16, 78, 79]. In general, there are three groups related to the alloying elements in Ti: (i) α-stabilizer elements which stabilize α-phase, (ii) β-stabilizer elements which stabilize β-phase or (iii) elements with no obvious effect on the phase transformation [80]. Therefore, depending upon the amount, type and distribution of the alloying elements and phases, different microstructures and properties of Ti materials can be achieved for multi medical applications.

AM is a nonconventional fabrication process for the field of the biomedical engineering which essentially depends on advanced manufacturing and information technology [81]. The constructed computer model of the process can be built depending on the data of magnetic resonance imaging (MRI) or computerized tomography (CT) of patient. Therefore, this manufacturing capability enabled AM techniques to produce biomedical parts with highly structure-property developments [82, 83]. It was reported that SLM can fabricate implant-supported frameworks for dental implants using Ti alloys. The required properties of produced frameworks such as biocompatibility, mechanical, chemical, etc were highly controlled in order to meet the criteria of high quality. In addition, clinical investigations were made on the fabricated frameworks to confirm their work [84]. The most common (α + β) Ti alloys used for biomedical applications, i.e. Ti–6Al–4V and Ti–6Al–7Nb, are well fabricated using SLM process. The literature reported serious attempts from the researchers to produce and investigate both Ti–6Al–4V alloy [18, 47, 67, 85–88] and Ti–6Al–7Nb alloy [34, 49, 89, 90] owing to their highly importance in biomedical field. However, it was proved that the main alloying elements in these alloys are biologically unfavorable since neurological troubles and cytotoxic effect may be induced from Al and V respectively [91–93]. Therefore, there is an urgent need to develop novel Ti alloys that are free from the alloying elements could cause toxic or even allergic effects. In addition, developed materials with lower elastic modulus is another very important issue for implantology in order to avoid the modulus mismatch between the implant and surrounding bones. The term ‘stress shielding’ describes this phenomenon of modulus mismatch which absolutely causes inadequate load transfer from the artificial implant to adjacent bone and in turn leads finally to bone resorption in case of orthopedic prostheses [94] or may absolutely delay bone regeneration in case of bone substitute biomaterials [95]. Moreover, the aseptic loosening and failure of implants may be attained as a result of bone resorption caused by stress shielding [93, 96]. The bulk Ti surface may develop interfactual fibrous tissue to create encapsulation that separates the implants from their surroundings which finally leads to poor bone–implant interfacial bonding [97].

In recent years, many researches and investigations have focused on β-Ti alloys to be imperative materials for biomedical applications [98–101]. The clinical investigations indicated that β-Ti alloys with stronger β-stabilizing elements, such as Nb, Zr, Sn, Ta, etc, are most appropriate materials for biomedical applications due to their safe alloying elements with superior characteristics. Several types of β-Ti alloys and more details about these alloys have been reviewed by [14, 16].

4. SLM-bulk Ti materials: microstructures and mechanical properties

4.1. General sight

It is well know that the biomaterials that replace bone must be highly biocompatible and analogous to bone in terms of mechanical properties. The excellent mechanical properties for engineering metals and alloys used for different biomedical applications are greatly required in order to meet diverse external loads which are normally subjected to the implants during their physical function. Hence, in order to obtain better imitation with the natural bone, the mechanical properties like strength, stiffness, toughness, and fatigue lifetime are necessary factors for the surgical implants. Emerging AM processes such as SLM is a significant approach to fabricate intricate geometrical parts with mechanical properties comparable [3] or even much better [102] in comparison to their wrought counterparts.

For SLM-Ti materials, the microstructure should be effectively structural defect-free and contain appropriate phases that can result excellent mechanical properties like high strength and ductility. It was reported that the microstructure of the Ti materials fabricated by SLM is different from that created by traditional methods, which in turn causes a major influence on the mechanical properties [19, 72, 102]. The solidification characteristics of the molten pool with completed liquid formation; such as liquid flow, solidification rate and thermal history, are most important parameters for establishing the phase transformation in the SLM produced materials [66], which significantly affect on the mechanical properties. Moreover, the proper design of the parameters used during SLM may lead to build final part with ideal structures and high characteristics in terms of morphology, porosity and mechanical properties. Therefore, to attain the best
building conditions, many studies have investigated the influence of various SLM parameters; like laser power, scanning speed, layer scan strategy, oxygen content and layer thickness, on the microstructure and mechanical properties of Ti materials [18, 34, 47, 85, 86, 103].

4.2. SLM-built Ti–6Al–4V alloy
The mechanical properties of the SLM-produced Ti–6Al–4V alloy are mainly based on the morphology, size and orientation of its micro-structural phases especially prior-\(\beta\) grains. The resultant micro-structural features from SLM are frequently columnar prior-\(\beta\) grains filled with acicular martensite (\(\alpha^\prime\)) phase as this process performed at temperatures less than 230 °C [33, 45, 86]. These specific micro-structural features could assist intergranular failure [85] and cause a considerable anisotropic mechanical performance along with great inconsistency in mechanical behavior once an external load is subjected in various orientations [45]. In other words, the fine acicular \(\alpha^\prime\) is the main phase of the microstructure of Ti6Al4V built by SLM due to rapid cooling during this process [47, 85], which offers outstanding mechanical strength but with relatively low ductility (less than 10%) [104]. The fracture strain values of as-fabricated Ti–6Al–4V alloy are significantly lower than that of the standard specifications Ti–6Al–4V alloy as wrought [105] or cast [106] for medical implants [88]. Hence, it should be a good correlation between the microstructure and the mechanical properties of Ti6Al4V produced by SLM in order to obtain better mechanical performance.

A suitable post heat treatment is an essential method to homogenize the microstructure, eliminate surface defects, alter the acicular martensite phase into equilibrium (\(\alpha + \beta\)) microstructures and also to decrease the thermal stresses at the same time [18, 45]. All these positive results would definitely increase the mechanical properties of Ti alloys. Accordingly, post-SLM heat treatment regimes are often used for increasing the mechanical properties especially ductility. There are many researchers focused on this highly required goal [18, 45, 49, 86, 107]. Table 2 presents some results of the mechanical properties of as-fabricated and treated Ti–6Al–4V alloy produced by SLM.

According to Yadroitsev et al [112], a significant phase transformations (\(\alpha\) to \(\beta\)) with various micro-structural features taken place in SLM–Ti–6Al–4V through post heat treatment regimes in various temperatures, which have a significant effect on the micro-hardness values. It was found that the micro-hardness values of heat treated samples are lower compared to the as-made state. The coarsening of lamellar microstructure at high treatment temperatures also caused a reducing in micro-hardness due to the effect of Hall–Petch relationship.

As is well known that the very high cooling rate in SLM makes the microstructure and the phase transformation kinetics of Ti–6Al–4V fabricated by SLM process significantly different from those produced by traditional processes. Therefore, attaining the mechanical requirements for surgical applications, by understanding the mechanisms of the phase transformation associated with martensite decomposition of as-manufactured Ti–6Al–4V compared to traditional alloy, is an essential point of research. In this line, the optimization of the heat treatment parameters, especially temperature, is a significant issue in order to reach to the suitable microstructure and then the required properties. It was reported that the annealing at low temperatures could cause an increase in age hardening and embrittlement, whereas annealing at temperature above the \(\beta\)-transus (\(\sim 1000^\circ\text{C}\)) could result in an extreme grain growth of the \(\beta\) phase [41]. Furthermore, the previous studies verified that the SLM-built Ti alloys can be treated with sub-\(\beta\)-\(\beta\)-transus annealing at 800–850 °C, aiming to relieve residual stress and create chosen lamellar (\(\alpha + \beta\)) equilibrium structure with advantageous mechanical properties [18, 113].

Huang et al [88] investigated the effect of three specific heat treatment regimes in different temperatures and cooling rates, i.e. subtransus treatment, supersolus treatment and mixed treatment, on the microstructure features and mechanical properties of SLM Ti–6Al–4V alloy. The authors suppose that the decomposition kinetics of \(\alpha^\prime\) phase presented in SLM Ti–6Al–4V alloy is slower than that of traditional (water-quenched) \(\alpha^\prime\) phase, which would be an assistant agent in refining the structure of the alloy after caring out an appropriate heat treatment. Moreover, the subtransus treated samples showed the best mechanical properties (superior combination of strength and ductility comparing with supersolus and mixed treatments. This is due to the modification in the microstructures (figure 1) as the subtransus treated samples showed a basket-weave structure while a typical lamellar structure with continuous grain boundary \(\alpha\) and quiaxed prior \(\beta\) grains were observed in supersolus treated samples. However, mixed treated samples exhibited both micro-structural features of subtransus and supersolus treated samples, i.e. equiaxed prior \(\beta\) grains and the basket-weave structure.

Xu et al [110] reported a novel fabrication design to solve some micro-structural challenges and their effect on the mechanical properties of SLM-built Ti–6Al–4V alloy. The designed experiments in this investigation included many affected factors of process like single-track deposition, multi-layer deposition and post heat treatment, in addition to deep tuning of several parameters of process including layer thickness, focal offset distance (POD) and energy density (E). This design derived from both phase transformation and processing
Table 2. Some results related to mechanical properties of SLM-built Ti–6Al–4V alloy.

| Processing | Structure | UTS (MPa) | YS (MPa) | EL (%) | References |
|------------|-----------|-----------|----------|--------|------------|
| AF         | acicular α’ | 1140 ± 10 | 1040 ± 10 | 8.2 ± 0.3 | [86] |
| AF         | fine acicular α’ | 1206 ± 8 | 1137 ± 20 | 7.6 ± 2 | [45] |
| AF + SR (730 °C/2h) | α + β with residual α’ | 1046 ± 6 | 965 ± 16 | 9.5 ± 1 | [45] |
| AF + post HT (1050 °C/1h/WQ)+(820 °C/2h/AC) | equiaxed α’ + β + α | 1019 ± 11 | 913 ± 7 | 8.9 ± 1 | [45] |
| AF + post HT (950 °C/1h W−Q)+(700°C/2h/AC) | Columnar α’ + βm + α | 1036 ± 30 | 944 ± 8 | 8.5 ± 1 | [45] |
| AF         | α + α’ | 930 ± 5 | 840 | 6.8 | [108] |
| AF         | acicular α’ | 1267 ± 5 | 1110 ± 9 | 7.28 ± 1.12 | [18] |
| AF + post HT (850 °C/2h/FC) | lamellar mixture of α and β | 1004 ± 6 | 955 ± 6 | 12.84 ± 1.36 | [18] |
| AF + HIP (920 °C/103 MPa/4h) | α + β | 1040 ± 30 | 980 ± 30 | 12.5 ± 0.5 | [109] |
| AF         | Fine α’ in columnar prior-β | 1269 ± 9 | 1195 ± 19 | 5 ± 0.5 | [104] |
| AF         | acicular α’ | 1150 ± 5 | 1000 | 8.5 | [110] |
| In situ HT | ultrafine lamellar (α + β) structures | 1240 ± 7 | 1106 ± 6 | 11.4 ± 0.4 | [110] |
| AF         | acicular α’ | 1191 ± 6 | 970 ± 6 | 5.37 ± 1.39 | [88] |
| AF + post HT (800 °C/2h/AC) | basket-weave structure of α and β | 1073 ± 9 | 1010 ± 11 | 17.05 ± 1.14 | [88] |
| AF + post HT (1050 °C/1h/AC) | α and equiaxed prior β | 988 ± 8 | 869 ± 4 | 13.34 ± 0.67 | [88] |
| AF         | acicular α’ | 1165.69 ± 107.25 | 1055.59 ± 63.63 | 6.10 ± 2.57 | [111] |

Note: UTS: ultimate tensile strength; YS: yielding strength, EL: total elongation to failure, AF: as fabricated, SR: stress relieved, HT: heat treatment, WQ: water quenched, AC: air cooled, FC: furnace cooled, HIP: hot isostatic pressing.
optimization to in situ decompose martensitic structure in SLM-produced Ti–6Al–4V into ultrafine lamellar \((\alpha + \beta)\) microstructures, i.e. \(\alpha\) laths and retained \(\beta\) phase. The results showed a typical microstructure of a columnar prior-\(\beta\) grain structure in SLM-fabricated Ti–6Al–4V using a layer thickness of 60 \(\mu\)m, a FOD of 2 mm, while keeping the E constant at 30.62 J mm\(^{-3}\). These SLM conditions provided the best combination of mechanical properties such as high ductility with maintaining great tensile strength comparing with other conditions, as presented in table 2. The mechanical properties in this study were higher than of both SLM-fabricated Ti–6Al–4V containing non-equilibrium acicular \(\alpha'\) martensite and conventional mill-annealed Ti–6Al–4V. On the other hand, the fatigue life of heat treated SLM Ti–6Al–4V samples with an ultrafine lamellar \((\alpha + \beta)\) structure is much better than SLM Ti–6Al–4V samples with \(\alpha'\) martensite. However, the ultrafine lamellar \((\alpha + \beta)\) structure attained in this investigation demonstrated strong texture which can induce an adversely effect on the fatigue characteristic. Therefore, the authors suggested to perform further microstructure optimization in order to obtain improved microstructure with a specific amount of ultrafine globular \(\alpha\) in the matrix of ultrafine lamellar \((\alpha + \beta)\) [114].

In same line, hot isostatic pressing (HIP) processing can also be used after fabricating the parts by SLM with the intention of relieving stresses and deceasing the amount of the residual porosity. Qiu et al [109] pointed out that deforming SLM–Ti–6Al–4V parts by HIP in specific conditions (920 °C/4h at a pressure of 103 MPa) could induce a desirable phase transformation from columnar grains and fine martensitic needles into \(\alpha\) and \(\beta\) phases, with decreasing the porosity levels at the same time. This led to improve the ductility considerably but with reduction in mechanical strength, see table 2. It is important to mention here that the mechanical properties of deformed SLM alloy were comparable to those of thermo-mechanically and annealed treated samples.

4.3. SLM-built CP-Ti

Commercially pure titanium (CP–Ti) is an important kind of Ti-based materials used for biomedical applications owing to its superior corrosion resistance and outstanding biocompatibility [30, 79]. However, the relatively poor mechanical properties of CP–Ti restrict its use in medical applications regardless of above excellent characteristics. Hence, there is an urgent need to develop the mechanical strength of this material using advanced techniques such as SLM. Table 3 shows the results of the mechanical properties of some as-built and treated CP–Ti materials fabricated by SLM.

According to Attar et al [102], to produce nearly full dense (>99.5%) CP–Ti parts with developed mechanical behavior, the better optimization of the SLM parameters, especially laser power and laser scanning speed, should be made. The change in the micro-structural phases was induced from \(\beta\) to relatively coarse plate-like \(\alpha\) grains when the values of laser power and scanning speed were less than 100 W and 100 mm s\(^{-1}\), respectively, due to incidence of energy thermalization. On the other hand, refined martensitic structure from \(\beta\) phase was presented in the microstructure when the laser scanning speed increases to above 100 mm s\(^{-1}\) owing to an increase in both thermal and kinetics under cooling. In this work, the most important issue was attained as
the mechanical properties of SLM-processed CP–Ti parts are markedly improved compared with same parts produced by traditional methods. This is because of the formation of refined martensitic structure during SLM along with grain refinement. Moreover, the fractography images of CP–Ti samples showed that the weakness of building direction is the major motivation for the early fracture which leads to pull out of Ti particles owing to some defects in structure such as incompletely melted powders and porosities. In contrast, the optimization of SLM parameters altered the mechanism of failure to nucleation, growth and coalescence of the microvoids. However, slight weakness of the building direction in almost full dense sample caused small quasicleavage facets.

4.4. SLM-built β-Ti alloys

Beta (β) and metastable β-Ti alloys are generally exceptional materials for different biomedical applications, particularly for orthopedic applications, because of their better properties such as high mechanical strength, very low Young’s modulus and superior biocompatibility [14, 16]. The close match of Young’s modulus values between the cortical bone and the implant is an imperative issue for any implant to be successful for long term implantation since it works on reducing the stress shielding phenomenon by decreasing the discontinuity in elastic characteristics at the interface of implant–bone [116]. From biological view, the ultimate biomaterials must have excellent biocompatibility by being free of toxic elements such as vanadium, nickel and cobalt. Therefore, the common biomaterials used; such as stainless steel, Co–Cr-based and Ti–6Al–4V alloys, are not appropriate for long term implantation because of the highly cytotoxicity effect of their alloy elements released in the human body [117]. Depending on our knowledge, very few numbers of studies and investigations were accomplished regarding the manufacturing of biomedical β-Ti alloys using SLM irrespective of their advantageous properties. Recently, a number of researchers has investigated the effect of SLM process on the structure and the properties of some β-Ti alloys used for biomedical applications [3, 68, 111, 118–121]. Table 4 demonstrates some results related to the mechanical properties of β-Ti alloys fabricated by SLM.

It is well identified that Ti–Nb alloy system is one of the most important alloys applied in medical field. However, this type of Ti alloys has high fusion temperature which makes its fabrication more complicated by conventional methods or even by SLM process. Therefore, Fischer et al [122] developed the micro-structural and the elastic modulus of in situ SLM-built Ti–26Nb parts, using a non-spherical powder bed of a mixture of Ti and Nb elements. The authors found that the laser energy is the most significant parameter of SLM as compact and homogeneous parts along with very low content of porosity and Nb particles left were obtained at high levels of energy (200 J mm$^{-3}$). The microstructure was composed of fully large β grains, typically oriented along the building direction, with non-columnar elongated grains perpendicular to the melting pool boundaries. Moreover, the grains showed an epitaxial growth with a texture (001) $\parallel$ 100 $>$ which is more positive to fabricate an alloy with low isotropic elastic modulus (77 ± 1.4 GPa). This result of modulus is greatly lower than that of CP–Ti and close to that of a solution treated Ti–26Nb ingot.

Ti–Ta alloys can be considered as a very essential kind of biomedical β-Ti alloys. However, their use in implantology is still limited. It is well known that the manufacturing of these alloys is so difficult owing to the large difference in melting point and density between Ti and Ta [91], which could cause inhomogeneity in the microstructure along with segregation of elements during the fabrication process. Therefore, it is very urgent to resolve these fabrication problems. Sing et al [111] have successfully produced Ti–50Ta parts using SLM process with an aim to investigate the microstructure and mechanical properties of this important kind of biomedical Ti alloys. In this research, the alloying element Ta was selected as it has ability to stabilize β-phase in Ti alloys, which could develop the mechanical properties especially Young’s modulus. The microstructure of SLM built Ti–50Ta alloy was found to be consisted from only β grains after rapid cooling with random orientation growth within

| Type of alloy | HV (MPa) | UTS (MPa) | YS (MPa) | E (GPa) | EL (%) | References |
|--------------|---------|---------|---------|---------|--------|------------|
| Ti–24Zr–4Nb–85n | ~240 | 665 ± 18 | 563 ± 38 | 53 ± 1 | 13.8 ± 4.1 | [3] |
| Ti–26Nb | 268 ± 5 | — | — | 77 ± 1.4 | — | [122] |
| Ti–50Ta | 284.5 ± 11.06 | 924.64 ± 9.06 | 882.77 ± 19.60 | 75.77 ± 4.04 | 11.72 ± 1.13 | [111] |

Table 3. Some results of the mechanical properties of as-fabricated (AF) SLM CP–Ti.

| Structure | HV | UTS (MPa) | YS (MPa) | E (GPa) | EL (%) | References |
|-----------|----|-----------|---------|---------|--------|------------|
| acicular $\alpha'$ | 261 ± 13 | 1136 ± 15 | 560 ± 5 | — | 51 ± 3.5 | [115] |
| acicular $\alpha'$ | 261 ± 13 | 757 ± 12.5 | 555 ± 3 | — | 19.5 ± 1.8 | [72] |
| acicular $\alpha'$ | — | 1136 ± 15 | 560 #x00B1; 5 | 113 ± 3 | 50 | [73] |

Table 4. Some results of the mechanical properties of as-fabricated SLM β-Ti alloys.
composites and SLM process exploited to develop the wear property of Ti materials, such as thin imperatives for enhancing the long-term service of Ti implants. A series of methods and techniques were owing to the release and gathering of metallic wear debris in surrounding tissues, which can result adverse particular regions. The poor wear behavior is one of the major reasons of unanticipated failure of implants known that the Ti materials used for load-bearing joint replacements are frequently subjected to wear in higher wear resistance is very necessary for any material used for biomedical applications.

5. Wear behaviour of SLM-built Ti biomaterials

Higher wear resistance is very necessary for any material used for biomedical applications [123, 124]. It is well known that the Ti materials used for load-bearing joint replacements are frequently subjected to wear in particular regions. The poor wear behavior is one of the major reasons of unanticipated failure of implants owing to the release and gathering of metallic wear debris in surrounding tissues, which can result adverse allergic tissue reactions and reduction in implant’s lifespan [125]. Consequently, better wear characteristics is an imperative factor for enhancing the long-term service of Ti implants. A series of methods and techniques were exploited to develop the wear property of Ti materials, such as thin film surface modification, Ti matrix composites and SLM process [98].

CP–Ti is a well-defined material used for biomedical applications, but it still has serious limitations regarding wear and hardness properties which restrict the extent of its use in medical field [123, 126, 127]. The lower wear resistance of Ti may be related to an unsatisfactory thickness of the passive surface protecting film formed [128] or due to the plastic deformation occurred at the surface/subsurface layers [129]. Therefore, SLM is recently used as an advanced technology to resolve all these problems associated with the fabrication of CP–Ti. According to Gu et al [66], the inappropriate controlling of the SLM parameters such as applied scan speed and attendant linear energy densities (LED) could lead to fabricate CP–Ti with lower densification and limited hardness and wear resistance. The thermal microcracks, interlayer micropores and creation of relatively coarse microstructure are the main reasons behind these negative outcomes. The resulted showed that an extremely high hardness (3.89 GPa) as well as lower coefficient of friction (0.98) and wear rate (8.43 × 10⁻⁴ mm³ N⁻¹ m⁻¹) can be obtained for SLM-built CP–Ti samples using optimized parameters (300 mm s⁻¹ and 300 J m⁻¹). Moreover, the better wear performance is attributed to the change in the mechanism of material removal during sliding from abrasion to adhesion owing to the development of an adherent, plastically smeared protective tribolayer on the surface.

The SLM is a vital process to fabricate complex-shaped CP–Ti parts with outstanding wear properties compared with traditional casting method, as proved by Attar et al [19]. These authors investigated the wear characteristics of SLM-produced CP–Ti parts and compared the results with those related to parts fabricated by casting. The microstructure and micro-hardness results showed that the SLM-produced CP–Ti samples have martensitic (α’) phase with finer grain size (figure 2(a)) with superior micro-hardness, while plate-like (α) is the predominant phase in the microstructure of as cast-produced CP–Ti samples (figure 2(b)). Regardless of the same wear mechanism in parts produced by both SLM and casting, this change in the microstructure led to induce better wear resistance in SLM-produced parts compared to their cast counterparts as shown in figure 3.
6. Corrosion behaviour of SLM-built Ti biomaterials

Corrosion resistance of metallic biomaterials in physiological fluid is an extremely crucial issue. The human body environment is a very intricate solution which includes several erosive species like chloride ions \([130]\). Furthermore, the biocompatibility characteristics of biomaterials are directly coupled with their corrosion behavior \([131]\). The corrosion can cause an adverse effect on the bone healing and surrounding tissues as a result of releasing extensive metallic ions which may finally guide to implant’s mechanical failure \([132]\).

As is well identified that Ti and its alloys have generally higher corrosion resistance owing to the spontaneous formation of a chemically stable, strong and compact oxide film formed on the surface \([15, 133, 134]\). Among Ti alloys, Ti–6Al–4V is the most commonly alloys used for biomedical applications because of its brilliant corrosion behavior \([135]\). The high corrosion resistance of this alloy is due to the reliability of the main constituent phases presented in the microstructure \([133]\). Moreover, the microstructure is a strong parameter for determining the various characteristics required for this Ti alloy including corrosion resistance. Therefore, studying the corrosion behavior of the biomedical Ti materials inside simulated human body fluid is an urgent issue. However, there still exists an open question about the corrosion behavior of SLM-manufactured Ti parts, whether these parts can establish better corrosion performance in comparison with parts prepared by conventional technologies. Dai et al \([136]\) investigated the electrochemical characteristics of Ti–6Al–4V alloy manufactured by SLM and compared the results with those related to commercial Grade 5 alloy in order to describe and evaluate the stability of the surface film and the corrosion resistance. The results showed that the SLM-produced Ti–6Al–4V alloy possesses less stable passive film on its surface and worse corrosion resistance in comparison with Grade 5 alloy over a large range of potential. This is due to the different phase transformations occurred in both investigated alloys. The micro-structural analysis revealed the formation of high amount of inferior corrosion resistance phase (acicular martensite, \(\alpha'\)) with small amount of prior \(\beta\) phase in SLM-produced Ti–6Al–4V alloy, unlike the typical \((\alpha + \beta)\) structure in Grade 5 alloy. In other words, the unfavorable corrosion behavior of the SLM-produced alloy is owing to the significantly high amount of acicular \(\alpha'\) and less \(\beta\)-Ti phase in the microstructure compared to the Grade 5 alloy.

According to Dai et al \([20]\), fully dense (>99%) Ti–6Al–4V alloy is fabricated by SLM at two different sample planes, i.e. at build plane (XY-plane) and build direction plane (XZ-plane). This is in order to study the effect of change in building planes during SLM on the corrosion resistance in two solutions (3.5 wt% NaCl and 1 M HCl).

The corrosion results revealed the anisotropy in corrosion resistance of different planes of the SLM-produced Ti–6Al–4V in 1 M HCl solution regardless of very minor variation in NaCl solution. Depending upon the XRD analysis, the volume fractions of \(\alpha'\)-Ti and \(\beta\) phase in the XY-plane sample are 88.1% and 11.9% respectively, while in the XZ-plane sample are 95.0% and 5.0%, respectively. It was reported that the acicular \(\alpha'\) martensite is a high energy state and metastable phase with regard to corrosion property \([136]\). Therefore, this variation in the amounts of the micro-structural constituents is the main reason behind the lower corrosion resistance of XZ-plane sample compared to XY-plane of the SLM-produced Ti–6Al–4V alloy. Moreover, the EIS measurements and the immersion tests support the above electrochemical results since they revealed the fact that the passive film formed on the XY-plane of SLM-produced Ti–6Al–4V has a superior protective property.

![Figure 3. Sliding wear rates of CP–Ti samples produced by SLM and casting as a function of load.](image-url)
than that on the XZ-plane especially in a harsher environment. The SEM images disclosed that some pores and pits present in the microstructure of both XY- and XZ- planes of the SLM-produced Ti–6Al–4V alloy. The authors pointed out that the microstructure of XZ- plane of the SLM-produced Ti–6Al–4V alloy possesses higher amount of pits than that in the XY- plane, indicating that the passive film formed on the XZ-plane shows a poorer stability and protective ability compared to that on the XY-plane.

7. Conclusion

The advanced SLM technique has been developed for eventual fabrication of bulk Ti materials with an aim to be used for different specific biomedical applications. This review offers a comprehensive overview on different biomedical Ti materials fabricated by SLM with a focus on the correlation between process parameters, micro-structural features and required properties. For this purpose, several recent studies and investigations derived from literature were also presented. The SLM process is an optimal choice to fabricate different highly intricate and tailor-made Ti implants with superior properties such as mechanical, tribological and corrosion.

In general, non-equilibrium martensitic microstructure develops in SLM-built Ti materials due to high cooling rate associated with the process. The high solidification and cooling rates involved in SLM typically yield fine grained microstructures with exceptional mechanical strength. The mechanical properties typically meet or even exceed the properties of conventional counterparts. However, the tensile fracture strain in SLM-built Ti parts is still limited. The well-defined optimization of different SLM processing parameters is required urgently in order to build full dense biomedical Ti implants with desirable microstructures and final properties. Moreover, it is proved that the selection of a suitable heat treatment of as-fabricated SLM-Ti materials is a significant method for decomposition martensitic structure and obtaining an appropriate microstructure with superior combination of strength and elongation. The SLM is an essential method to produce medical Ti parts with exceptional wear performance compared with conventional methods. However, the worse adjustment of the SLM parameters could lead to fabricate Ti materials with lower wear resistance as a result of creation several structural defects such as microcracks, micropores and relatively coarse microstructure. The surface passive film of SLM-built Ti materials has less stability and poor corrosion resistance in comparison with traditional counterparts due to the variation in the micro-structural constituents. Hence, more and more serious attempts and investigations are still required for understanding the corrosion mechanisms of SLM-Ti parts with an essential goal to increase the corrosion characteristics. Production and design of different medical Ti implants with outstanding biofunctional performance are still an imperative future work direction, regardless of considerable achievements being made in this direction. Manufacturing of common and new SLM-built Ti materials is continually being investigated, with significant developments made from previous researches.

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