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RECENT MANUFACTURING TECHNIQUES OF METALLIC BIOMATERIALS

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ABSTRACT

Additive Manufacturing (AM) which is universally known by the acronym of 3D printing has transformed the process of producing metallic biomaterials, enabling the production of complex, customized, and high-performance implants. The range of applications of AM is increasing across all sectors due to its ability of producing defects free products and tailoring the properties as per the requirement. In order to gather most of the metallic applications of AM and its technicality at one place, this review is dedicated to AM techniques that are applied for the creation of metallic biomaterials, particularly for biomedical implants. These laser-based AM techniques allow precise control over microstructure, porosity, and mechanical properties, facilitating the development of patient-specific implants with enhanced biocompatibility and durability. The review delves into the suitability of various materials, let say alloy that contains titanium, cobalt, and chromium, for AM processes, highlighting their advantages in respect to mechanical strength, high corrosion resistance, and compatibility with biological tissues. Additionally, it addresses the challenges associated with AM, including thermal stress management, surface finish optimization, and quality control. The findings underscore the significant potential of AM in advancing medical technology, tailored to individual patient needs that will certainly help in developing understating for the creation of future biomaterial with ease.

Keywords: Metallic Biomaterials, Additive Manufacturing, Properties of Metallic Biomaterials

1. INTRODUCTION

Metallic biomaterials are engineered to support biological tissues within the body. They are utilized in various applications, such as hip replacements, dentistry implants, orthopedic interventions, and stents1). Because of ease in fabrication and the wide range of available fabrication methods and techniques, metals are preferred choice in orthopedics, dentistry, and neurovascular implants2). A wide range of metals are utilized in the manufacturing of bio implants for instance, bioimplants made of stainless steel (SS)3), Magnesium (Mg) alloy, Titanium (Ti) alloy, and Alloys of Chromium and Cobalt (Cr-Co) have high demand in biomedical implant sector4,5). Although, materials made of metals and alloys satisfy many requirements essential to biomedical implants, but because of their weak bonding with the surrounding tissue or bone, many challenges are encountered during the production of bio implants to fix the interface issue. The beginning of the failure of medical implants usually occurs at the interface of the implant and tissue which results in the development of a non adherent layer and movement at the tissue-implant junction6). There are several significant challenges related to the application of metallic biomaterial that can be used for implants. For example, controlling the biodegradation rate is crucial for resorbable metals like Mg or iron (Fe) because fast degradation can result in a loss of mechanical strength, failure of biomaterial before the healing of surrounding tissue or regained its function7). Conversely, if a metal is implanted inside human body for a long period, then chances of risk of cutaneous and hypersensitivity reactions8) will be high. Usually, in implants, high modulus (stiffness) of metals such as titanium, stainless steel, or cobalt-chromium alloys often exceeds that of natural bone tissue. This mismatch in stiffness can result in stress shielding, leading to osteopenia9). These issues can lead to increased healthcare expenses due to complications, follow-up surgeries, and extended recovery periods. Consequently, substantial efforts have been made to address these issues, leading to the development of numerous surface modification and bulk modification techniques for a variety of materials10) including plasma and acid etching, laser ablation, surface fictionalization, ion implantation, coating, and grain refinement, among others. Moreover, significant advancements are being made not only in the chemical composition of materials but also in the methods of fabricating implants11). The advent of three-dimensional (3D) printing in biomaterials field has revolutionized the creation of complex, patient-specific structures, enabling the precise fabrication of biomaterials that meet the unique anatomical and functional needs of individual patients12). Throughout time, biomaterials having porous structure are developed by using traditional fabrication techniques called powder metallurgy that includes metal injection moulding (MIM) and the technique based on space-holder13,14). AM has revolutionized the process of making implants made of metals, particularly for applications of bone tissue regeneration15). AM allows precise regulations on the design and structure of implants that enables the fabrication of complex geometries and porous structures that closely mimic natural bone16). Since the early 2000s, significant research and development have been dedicated to the application of MIM to produce biomedical implants and devices. Consequently, numerous articles have been reported on the advancement in AM for implants material where the primary focus was on using MIM to manufacture various materials with enhanced biocompatibility, particularly SS, Ti, CoCrMo alloys, as well as Mg and Fe alloys17). An investigation was performed for the entire processing chain for MIM of Mg–Ca alloys, aiming to attain improved tensile strength (ultimate) of up to 141 MPa, an enhanced tensile strength (yield) of 73 MPa, Young's modulus of 38 GPa and an elongation at fracture (Af) of 7%. A study was carried out for thermal deboning to identify the optimality of environment condition and furnace type, appropriate sintering time, sintering temperature, pressure, and heating rates18). Metallic implants are difficult to fix that leads to complications like implant loosening, instability, and reduced effectiveness19). The lack of osteo conductive and issues of corrosion resistance and wear can lead to formation of debris and release of corrosive ions20), particularly applicable for metals21). Consequently, the surface of a bioimplant is crucial in the biological environment, as reactions take place directly on the implant's surface22). Surface modifications methods of metallic bio-implants are being extensively researched with various bioactive materials to prevent adverse effects like poor compatibility with the bio fluids, infections after surgery, and increased risks of corrosion of implant surface23,24). Right surface modifications of the material allow for tailoring and enhancing cell interactions, biocompatibility, and adhesion25). This review incorporates the thorough discussion on the criticality of modern manufacturing methods and techniques that are applied on the surface are typically used in the manufacturing of metallic biomaterial implants.

1.1. REVIEW OF EXISTING LITERATURE AND JUSTIFICATION FOR THE CURRENT REVIEW

Many reviews explore metallic biomaterials and their manufacturing methods for biomedical implants. For instance, characteristics, compatibility, and applications are reported with recommendations in recent studies2,23). The advancement in AM was brought further by focusing on powder bed fusion and electron beam melting but limiting the work on existing technologies and sidelining the material behavior26,27).

Most existing reviews separate metallurgical details from biomedical performance. Few connect manufacturing methods with structure, properties, and final performance. AM technologies are evolving quickly, and demand for custom, high-performance implants is growing. There is a clear need to combine the recent advancements in laser-based, binder based, extrusion based and directed energy deposition methods in one place.

This review compares modern manufacturing techniques for metallic biomaterials, explains how they work, and links them to material choice and biological results. The aim is to give researchers, engineers, and clinicians clear guidance for selecting and improving processes for biomedical applications.

2. METALLIC BIOMATERIALS USED FOR BIOMEDICAL IMPLANTS

Metallic biomaterials are categorized on the basis of unique physical, mechanical, and biocompatible properties possess by the material, which makes them appropriate for wide range of biomedical applications28). SS, various Co-Cr alloys, Ti alloys, alloys of noble metals, shape memory alloys (SMA), and metals that are degradable in human environment like Mg, all offer distinct advantages29). For example, Ti is valued for its biocompatibility and strength, while SMA are known for their ability to return to a predefined shape after deformation, and biodegradable metals degrade safely in the body. Each material is selected based on the specific requirements of the medical application.

2.1. STAINLESS STEEL

In biomedical applications, SS are valued for their strength, corrosion resistance, and affordability. Traditional SS contains nickel, which provides corrosion resistance but can cause allergic reactions in some patients. The most commonly used SS in these applications is SS of conventional type30). Nickel-free SS is increasingly being applied in biomedical implants credit to its enhanced biocompatibility and resistance against corrosion compared to traditional SS that contain nickel. Nickel-free SS reduces the risk of allergic reactions or sensitization to nickel. Stainless steels are suitable for various applications, including biomedical uses like implants and surgical tools31). 316L SS is highly preferred implant material in the medical field because it has ease of fabrication, excellent mechanical performance, resistance to corrosion, biocompatibility, and machinability. Its medical effectiveness is also statistically proven32). And therefore suitable choice for making of valve of heart, cardiovascular stents, implants for orthopedic replacement, hip joints and artificial knees and the most important one is artificial bone materials33).

2.2. TITANIUM AND TITANIUM ALLOYS

Alloys made of Ti are highly suitable for use as permanent implants because of their excellent biocompatibility, superior resistance to corrosion, mechanical integrity, and stable properties34). Titanium can spontaneously form a highly stable and non-toxic passivation layer of TiO_2, which isolates the Ti implant from surrounding body fluids35). Ti6Al4V, the most commonly utilized Ti-based biomedical alloy consists of aluminum (6.5 % by weight) and vanadium (3.5–4.5% by weight). It exemplifies a structure containing two phases in which one phase is distributed within another phase. Annealing is performed to the material leading to precipitation followed by slow cooling. At last thermal aging process is performed on the material36). Ti15Nb4Ta4Zr alloys are well-suited for orthopedic implants because of its outstanding strength, high resistance against corrosion, cyto-compatibility, and biocompatibility37). Titanium's classification as a biologically inert material stems from its relatively lower bioactivity compared to ceramics38). However, its inert nature discourages bone or tissues bonding as some ceramic materials do. But, surface modifications can enhance its bioactivity; improving osseointegration in medical applications39). Recently Ti-based alloy with Mo and Si as chief alloying was developed through vacuum arc remelting (VAR) that showed enhanced chemical stability and biocompatibility40).

2.3. MAGNESIUM AND ITS ALLOYS

Alloys made of Mg as base material are regarded as ideal degradable materials for implants used in medical purposes41) and medical implants due to its bio-resorbable properties. Their ability to interact positively with macrophages is significant for promoting tissue healing and controlling inflammatory response. The compatibility with macrophages not only aids in faster inflammation regression but also supports overall implant integration and performance42).

2.4. ZINC AND ZINC ALLOYS

Zinc and zinc-based alloys are emerging as promising materials for biomedical implants, particularly for biodegradable implants43). Zinc-based biodegradable stents are being investigated for use in coronary artery disease treatments44). These stents provide temporary support to blood vessels, maintaining patency before gradually dissolving as the vessel heals. Unlike other degradable metals such as Mg and Fe, Zn corrosion process cannot generate hydrogen and occurs at a relatively moderate rate45). Zinc alloys are being studied for bone fixation devices, such as screws, plates, and pins, which provide temporary structural45) support to fractures before being absorbed by the body46). Zn based (Zn-0.8Cu, Zn-0.8Mn, Zn-0.8Li) stents have better hemo compatibility than stents of pure Zn and 316L SS47).

2.5. COBALT (CO) AND CHROMIUM (CR) ALLOYS

Cobalt and Cr alloys exhibit remarkable antimicrobial characteristics and are extensively implemented in wide range of biomedical applications, including dentistry, cardiovascular treatments, and orthopedic devices48). The alloy made of Co-Cr possess good resistance against corrosion due to the formation of a passive protective film made of Cr2O3. A 99 % success rate was achieved during the Co-Cr Alloy stent implantation process. A noble nickel-free Co alloy was developed for vascular stent preparation, which demonstrated excellent mechanical, corrosion resistance, and biological properties49).

2.6. HIGH ENTROPY ALLOYS

High Entropy Alloys (HEAs) are a unique class of materials consisting of multiple principal elements mixed in roughly equal proportions50). HEAs typically consist of five or more elements, each present in similar concentrations (usually between 5% to 35% atomic percentage)51). HEAs have garnered noteworthy attention of research society because of their wide range of functional properties, including excellent biocompatibility and mechanical properties52). The potential of using an Al0.1CoCrFeNi HEA as a peripheral vascular stent was confirmed by evaluating its mechanical properties and comparing them with those of commercial 316L SS stent materials53). The TiHfZrNbx HEA indeed has some impressive properties for medical applications. Its high radial strength and superior corrosion resistance make it a promising material for small diameter vascular stents54).

Table 1

Table 1 Advantages, Disadvantages and Applications of Metallic Biomaterials

Metallic Biomaterial	Advantages	Disadvantages	Applications	references
316L SS	Economical, High Mechanical strength	High stress-shielding effect, high level of Cr and Ni causes sansitization	Orthopediac implants,stents,hip joints	[33]
Ti and its alloys	Good corrosion resistance, excellent biocompatibility, mechanical integrity , stable properties	Low wear resistance, long term toxic effect	Knee, hip replacement, bone plates	[34,35]
Mg and its alloys	Biodegradable, biocompatible, non-toxic	Rapid degradation, poor ductility	Segmental defect in bone	[41,42]
Zn and Zn alloys	Biodegradable, biocompatible, moderate corrosion rate	Low mechanical strength	Cardiovascular stents, orthopaedic implants	[45]
Co and Co alloys	Good mechanical strength and corrosion resistance	Toxic and can trigger inflammation	Dentistry, cardiovascular treatment	[48]
High Entropy Alloys	Balanced mechanical properties, good biocompatibility	High cost, immature preparation process	Vascular stents	[52,53]

This makes it a compelling alternative to more traditional biomaterials like CP-Ti (Commercially Pure Titanium) and Ti6Al4V (Titanium alloy with aluminum and vanadium). HEA could be worth considering for exploring options for high performance implants55). Various advantages, disadvantages, and application of widely used metallic biomaterials are given in Table 1.

3. REQUIRED PROPERTIES OF METALLIC BIOMATERIALS

Biocompatibility Biomaterials need to be biocompatible, meaning they should not provoke an adverse response from the body. They must be non-toxic, non-carcinogenic, and possess appropriate physical and mechanical properties to effectively serve as substitutes for biological tissues, along with meeting other necessary criteria.

High corrosion resistance Corrosion occurs when metals degrade due to chemical reactions with the body's fluids, such as blood or tissue fluid, potentially leading to implant failure, inflammation, or toxicity56). An implant made from a biomaterial with low corrosion resistance may release metal ions into the body, leading to toxic reactions. To prevent this, implants are made from metals with high corrosion resistance ensuring intactness and perform intended functions over extended periods53).

Mechanical Strength The material needs to have a low modulus paired with high strength to extend the implant's service life and prevent loosening; thereby avoiding the need for revision surgery57). Fatigue resistance is crucial for biomaterials used in implants. Implants often undergo repeated loading and unloading cycles, which can lead to fatigue failure if the material isn’t adequately resistant58). High fatigue strength protects against stress-induced fractures. Fatigue has been reported as a cause of failure in hip prostheses59).

Osseointegration It is essential for providing long-term stability and functionality to various medical implants, improving quality of life for patients60). Osseointegration is the process through which a biomaterial, typically an implant, integrates with the surrounding bone tissue. This is a key factor in the success of many orthopedic and dental implants61). Surface roughness and topography are crucial factors in achieving effective osseointegration62). Implant relaxation occurs when the implant surface integrates with the surrounding bone. Some researchers argue that Osseo-integration is problematic due to the risk of difficulty to remove the implant later63). However, others have demonstrated that the implant can be removed safely. As a result, Osseo-integration is a valuable characteristic for certain applications, like implants, where it is essential for ensuring the suitable integration of the bone and other tissues with the implant64).

4. ADDITIVE MANUFACTURING METHODS OF METALLIC BIO-MATERIAL

The features of AM have made it an important method for making medical devices used in areas like dentistry implants, heart treatments (cardiovascular implants), orthopedic implants and bone repairs65). Additive manufacturing is grounded in rapid prototyping technology. In this method, material is fused layer by layer to create a laminate with a specified structure66). AM-based methods have made it possible to manufacture complex-shaped, multifunctional implants with engineered osteoconductive, interconnected porous structures that traditional methods could not produce67).

4.1. POWDER BED FUSION (PBF)

PBF (Powder Bed Fusion) is one among the advanced additive manufacturing techniques widely used in aerospace, medical implants, and complex industrial components. This method has ability to produce highly detailed and strong components. It uses either a Selective Laser Sintering (SLS) or Melting68) or a high energy Electron Beam Melting (EBM)69) to selectively fuse powder particles70). Classification of PBF techniques is shown in Figure 1. A thin and even layer of material in the form of powder (polymer, ceramic, or metal) is made to spread over the build platform using a roller71). Heat is supplied from a laser or electron beam that selectively fuses the powder in areas defined by the 3D CAD model of the part being printed72). The heat source follows a predetermined pattern to melt or sinter the powder in the exact shape required for that layer. After the completion of fusion of one layer, the lowering of build platform is lowered equal to the thickness of the next layer, and spreading of another layer is carried out. The heat source then selectively fuses the powder in this new layer, bonding it to the previous one. This process repeats until the entire object is built. There are three different types of PBF methods73).

Figure 1

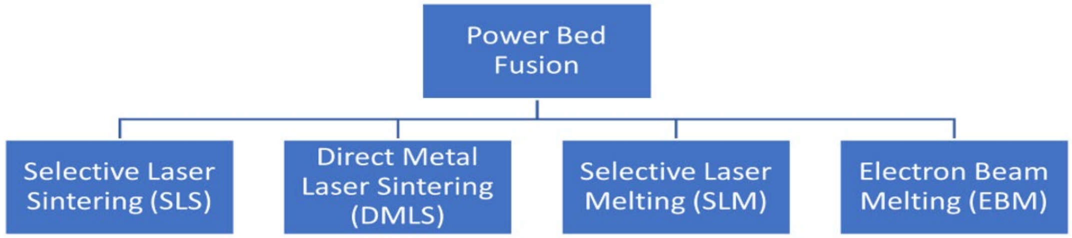

Figure 1 Classification of PBF Techniques

4.1.1. SELECTIVE LASER SINTERING (SLS) TECHNIQUE & DIRECT METAL LASER SINTERING (DMLS) TECHNIQUE

Selective Laser Sintering is a versatile AM technology that can work with both metallic materials (powder form) and non-metallic materials74). SLS works with a wide range of materials, including engineering-grade consisting metals and polymers75). The other method- Direct Metal Laser Sintering (DMLS) is specifically used for metals76). In SLS, material in the powder form is partially melted; this melting process is followed by rearrangement of solid-phase particle and after that liquid phase solidification to achieve the desired powder densification. SLS setup has a powder supply and recoating mechanism. Powder supply mechanism consists of a powder feed bin that holds the powder material used for building the product. Recoater or spreader has a blade or roller that after each layer is processed, the platform lowers incrementally to allow for the next layer to be spread. This sequence continues until all layers are built77). A high-powered laser is used to generate the heat required to fuse or sinter the powder. A laser scanning system directs the laser beam onto the specific areas of the powder bed as per the design. The type and power of the laser is depending on the material

to be processed. A control unit manages the machine operations, such as path of laser, platform operations, such as path of laser, platform movement, temperature, and recoating system. SLS is a relatively straightforward process that supports a wide range of moulding materials. However, due to the presence of solid-phase particles, SLS can result in lower tensile strength and poorer surface roughness78,79).

Figure 2

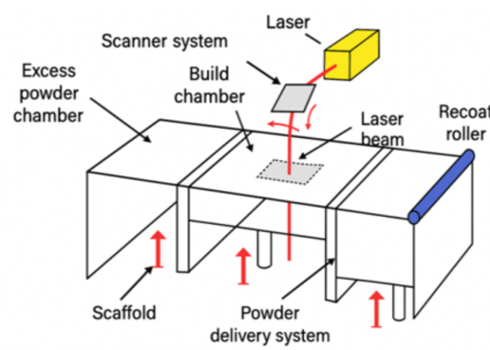

Figure 2 Selective laser Sintering [74]

4.1.2. SELECTIVE LASER MELTING (SLM)

In SLM (Selective Laser Melting) technique high-energy lasers are used to melt powders and manufacture metallic surfaces and parts with precise dimensions and desirable surface properties80). In SLM, a new layer of powder is layered above the previous layers in the build platform after the slicing of each layer is completed. This process takes place in a vacuum chamber. This vacuum chamber, also called protected chamber to prevent the metallic materials from reacting with other gases81). Although SLM is capable to produce finer and precise structures with excellent mechanical properties, it is fairly expensive. The selection of process parameters plays a critical role in the properties and dimensional accuracy of product generated82). Process parameters of SLM are shown in Figure 2. The fine focus spot of high-energy laser contributes to precise dimensional accuracy along with improved surface roughness83). SLM produces fully dense, crack-free Ti_2AlNb samples with substrate preheating ≥700 °C84). SLM-fabricated Ti_2AlNb/Ti 6Al 4V bimetallics required substrate preheating to avoid cracking, while annealing and HIP improved interface homogeneity and tensile strength85)

Figure 3

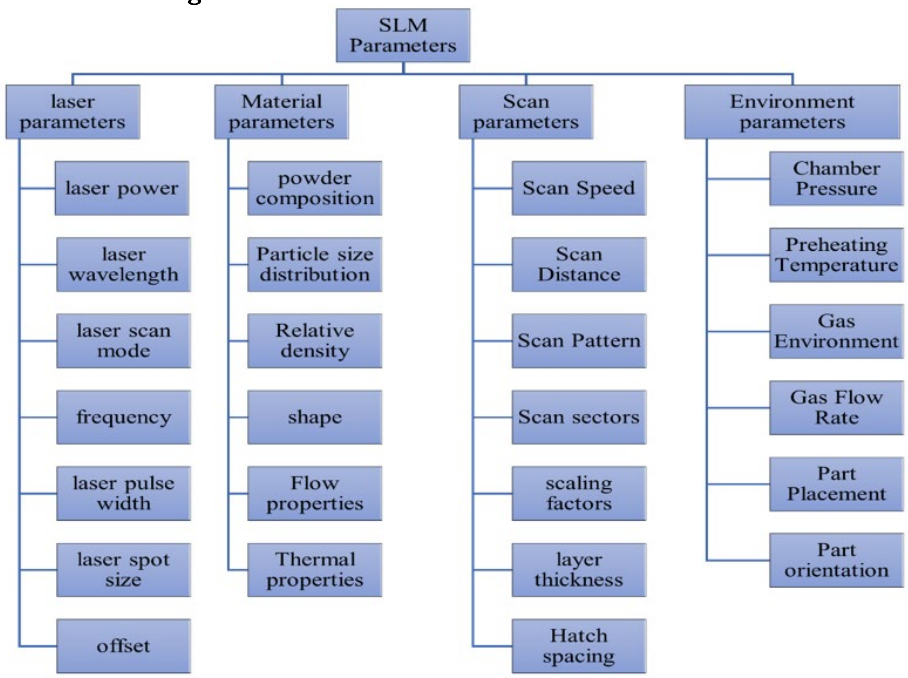

Figure 3 Selective Laser Melting Process Parameters [82]

SLM enables fabrication of porous titanium implants with controlled pore size and porosity; however, optimal pore structure for bone ingrowth remains unclear. A rabbit model study compared implants with 65% porosity and pore sizes of 300 µm, 600 µm, and 900 µm (P300, P600, and P900)86). Titanium clasps made by SLM, milling, and casting were compared for retention and durability. Milling provided stable retention, while SLM showed early fracture, indicating milling as a better alternative to casting and SLM require further reinfement87)

Surface modification of SLM Ti implants using sand blasting/alkali-heating or sandblasting/acid-etching reduced roughness and inhibited Osteo-clast differentiation88). A bionic porous Ti spinal implant was designed using CT-based CAD and fabricated via SLM demonstrated high strength, adequate porosity, and improved biocompatibility after heat treatment, showing SLM potential for clinical spine reconstruction89). High-density (>99%) SS316L parts were fabricated via SLM using fast scanning speeds enabled by a 380 W laser, increasing build rate by ~72%. The parts showed superior microhardness (213–220 HV) compared to annealed SS316L with no loss in density or mechanical properties90). SLM was used to fabricate 316L SS porous structures with designed pore sizes (0.4–1.0 mm) and interconnected architectures exhibiting minimal corrosion (3.0 mpy) and supported cell viability, with 0.78 mm pores (61.2% porosity) promoting highest biofilm formation and cell ingrowth91). SLM is also a popular AM technique for Mg alloys92). Building on the understanding of SLM process–property relationships, SS 316L was initially used to optimize printing parameters, achieving >99.8% dense parts with biocompatible surfaces suitable for mesenchymal stem cell growth. Using these insights, Mg alloys were processed by SLM to study the effects of alloy composition and laser power on porosity, corrosion behavior, and degradation rate, revealing critical process–property–performance relationships for biodegradable implants93). Also, Mg/Ti bimetallic joints were successfully fabricated by embedding an SLM-produced Ti lattice into molten Mg via compound casting, achieving a high bonding strength of 95.4 MPa and minimal porosity (0.048%) due to the optimized lattice design ($l/ds = 5.8$, $dn/ds = 2.4$) and rough SLM surface promoting a serrated interface94). Similarly, Mg/Ti bimetal composites produced using an ultrasonic-assisted compound casting method with an SLM-fabricated Ti lattice interface and optimized geometric parameters ($d = 1$ mm, $l/d = 3$, $d_1/d = d_2/d = 2.5$) demonstrated a joint strength of 77.3 MPa, confirming the importance of lattice architecture in enhancing mechanical performance95). Furthermore, SLM-fabricated CoCr alloys showed improved hardness (>36%) and tensile strength (1395 MPa vs. 1283 MPa) after porcelain sintering, attributed to a γ (FCC) to ε (HCP) phase transition, though at the expense of reduced corrosion resistance97).

4.1.3. ELECTRON BEAM MELTING (EBM)

The Electron Beam Melting (EBM) technique utilizes high-energy beams of electrons as a heat source to melt metal powders. Initially, metal powder is spread precisely on a flat horizontal plane surface. An electron beam then aligns the powder according to a CAD file design followed by layer-by-layer melting of metallic powder on the flat plane in vacuum chamber until the entire part is built98). The important factors in EBM are action time, current and acceleration voltage. Parts produces by EBM technique have lower residual stresses and reduced gas contamination. However, obtaining precise dimensional accuracy and fine structures is challenging due to poor focus of electron beam on a very fine spot80). EBM has wideapplication in manufacturing of bio metallic implants. Lattice structures of Ti–6Al–4V produced by EBM showed that geometry, dimensions, and relative density significantly affect compressive strength and energy absorption99). Corrosion and biocompatibility studies on Ti–6Al–4V samples from EBM and SLM revealed superior corrosion resistance and bioactivity in EBM samples due to a higher β phase fraction and greater surface roughness, which enhanced apatite nucleation and growth101). Furthermore, patient-specific EBM-fabricated dynamic thoracic implants exhibited flexibility comparable to native thoracic structures, with tress values below 30% of Ti–6Al–4V yield strength, confirming adequate mechanical performance for clinical applications102).EBM-fabricated porous Ti-6Al-2Zr-2V-1Mo alloys with controlled porosity (22–32%) achieved tailored mechanical performance (250–350 MPa) and effective metallurgical bonding, offering promise for biomedical implants103). EBM-fabricated Ti-6Al-4V implants with lattice pores (880–1400 μm, 57.5% porosity) and ACaHW surface treatment achieved bone-matched stiffness (3.57 GPa), induced apatite formation, and showed superior osseointegration and fixation strength104). High-nitrogen steel billets produced by electron-beam additive manufacturing exhibited a dual-phase microstructure (~40% ferrite) due to Mn depletion and altered solidification105). Preferential formation of nitrogen-rich austenitic dendrites caused compositional inhomogeneity, resulting in reduced strain hardening and ductility but retained high yield strength106). EBAM-fabricated AZ31 Mg alloy showed grain refinement, Al_8Mn_5 and $Mg_{17}Al_{12}$ precipitation, and achieved superior tensile strength (230 MPa) and elongation (13.5%) compared to die-cast AZ31107). ZKX50 Mg alloy processed by EBP and FSP with heat treatment showed refined microstructure and spheroidized $Ca_2Mg_6Zn_3$ particles, reducing microgalvanic corrosion. This improved corrosion resistance while maintaining strength, making it promising for biodegradable implants108). EBM-fabricated Ti-6Al-4V porous implants with Voronoi and randomized structures were evaluated mechanically and in vivo109). Pure iron fabricated by electron beam melting (EBM) showed distinct micro structural features compared to hot-rolled iron, resulting in significantly lower fatigue strength under both dry and corrosive conditions110). Table 2 Summarizes the several research works on the fabrication of biomedical implants by EBM.

Table 4

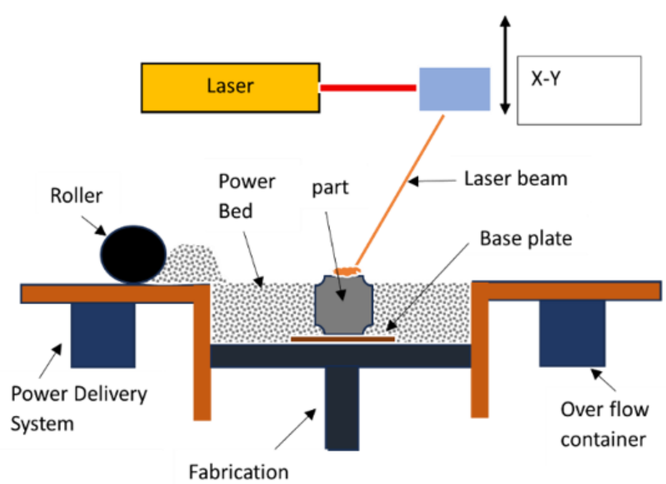

Figure 4 Selective Layer Melting [96]

Figure 5

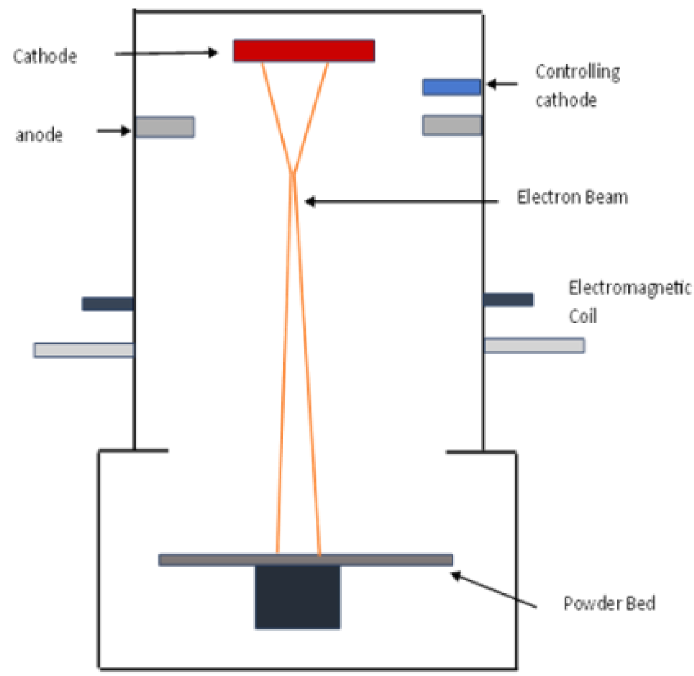

Figure 5 Electron Beam Melting [100]

Table 2

Table 2 Summary of Research Work on Fabrication of Biomedical Implants by EBM

Material / Alloy System	Additive Manufacturing Process	Study Focus / Application	Key Findings	Mechanical Properties / Performance Highlights	Clinical / Functional Relevance	Reference
High-nitrogen steel (Fe–20.7Cr–22.2Mn–0.3Ni–0.6Si–0.15C–0.53N)	Electron Beam Additive Manufacturing (EBAM)	Microstructure, mechanical properties, fracture behavior	Dual-phase (~40% ferrite) due to Mn depletion; high yield strength, lower ductility; transgranular fracture.	High yield strength, lower strain hardening and ductility	Implant potential with high strength	[106]
Ti–6Al–4V lattice structures	Electron Beam Melting (EBM)	Effect of geometry & density on compressive strength	Geometry-dependent mechanical behavior; same density gave different Young's modulus.	Tailored compressive strength & energy absorption	Lightweight implants with customized stiffness	[99]
Ti–6Al–4V	EBM vs. SLM	Corrosion properties & biocompatibility	EBM superior to SLM due to higher β phase & no α′ martensite; enhanced bioactivity.	Better corrosion resistance & bioactivity	Improved osseointegration	[101]
Dynamic thoracic implant (Ti–6Al–4V)	EBM	Patient-specific thoracic reconstruction	Native-like flexibility; stress <30% of Ti–6Al–4V yield strength; good mechanical performance.	Adequate strength, flexibility	Patient-specific chest wall reconstruction	[102]
Porous Ti–6Al–2Zr–2V–1Mo	Electron Beam Selective Melting (EBSM)	Porous structures for biomedical applications	Controlled porosity (22–32%) tailored strength (250–350 MPa); strong metallurgical bonding.	Tensile strength 250–350 MPa	Lightweight, bone-integrated implants	[103]

Ti–6Al–4V lattice pores (880–1400 µm)	EBM + ACaHW treatment	Implant osseointegration &fixation	Bone-matched stiffness (3.57 GPa), apatite formation, superior fixation strength.	Compressive load 78.9 MPa, Young's modulus 3.57 GPa	Enhanced osseointegration &fixation	[104]
High-temp Ti–Al–Zr–Si–Mo–Nb–Sn	EBM	High heat-resistant alloy production	Defect-free, homogeneous microstructure; complex silicides; improved heat resistance.	Refined microstructure; heat resistance	Aerospace & medical potential	[105]
AZ31 magnesium alloy	EBM	Microstructure & mechanical properties under varying energy densities	Grain refinement (20–40 µm), Al_8Mn_5 and $Mg_{17}Al_{12}$ precipitation, layered deposition structure; optimum manufacturability at 1.432×10^{10} $J\cdot m^{-3}$	Tensile strength 230 MPa, elongation 13.5%, superior to die-cast AZ31	Lightweight structural and potential biomedical applications	[107]
ZKX50 magnesium alloy	Electron Beam Processing (EBP) + Friction Stir Processing (FSP) + Solution Heat Treatment (HT)	Effect of microstructure refinement on corrosion rate for biodegradable implant use	Uniform refined microstructure, spheroidized $Ca_2Mg_6Zn_3$ particles reduced microgalvanic corrosion	Enhanced corrosion resistance without grain coarsening; mechanical property improvements retained	Suitable for biodegradable implants with improved degradation control	[108]
Ti–6Al–4V porous structures (Voronoi & randomized)	EBM	Mechanical and in vivo performance for orthopedic implants	Voronoi structure showed higher mechanical strength and greater bone ingrowth	Improved tensile, shear, and abrasion performance	Enhanced osseointegration and implant fixation	[109]
Pure iron	EBM vs Hot Rolled	Microstructure, corrosion, and fatigue behavior for biodegradable implants	EBM iron showed different microstructure with lower fatigue strength than HR iron, both in dry and corrosive conditions	Reduced fatigue strength due to AM microstructure	Potential for biodegradable implant design with tailored corrosion but limited fatigue life	[110]

4.2. INKJET 3D PRINTING OR BINDER JETTING

Inkjet 3D printing, also known as binder jetting (BJ), uses liquid droplets to selectively bind particles together on a powder bed. In this process, droplets of liquid or solid suspension materials are deposited through a small nozzle at low temperatures and pressures111). In this process, a roller spreads the metal powder over the build plate, and a liquid binder is selectively deposited by a print head. Layers are added sequentially until the part is fully 3D printed. The printed part is then removed from the build plate and de-powdered. Afterward, it is placed in a furnace to burn out the binder, resulting in the formation of voids and micro-porosity within the structure. Due to fragility, a secondary process like infiltration or sintering is necessary to strengthen it. In this step, a low-melting-point powder, such as bronze, is infiltrated into the voids using capillary action to reduce micro-porosity112). During the sintering or heat treatment process, it is placed in a high-temperature furnace. In this step, not only is the binder burned out, but the metal powder is also partially melted and fused together to form what is known as the "brown part"113). A key advantage of this process is that support structures are not needed for complex geometries. Additionally, the low cooling rate significantly reduces residual stress in the final product. Binder jet 3D printing of Ti–6Al–4V, optimized by powder size selection and high-temperature sintering (≥1250 °C), achieved ≥96% density, fine α+β microstructure, and mechanical properties (880 MPa, 6% elongation) comparable to MIM parts. This demonstrates its potential for ASTM F2885-compliant biomedical implants114). Binder-jetted Ti–6Al–4V parts exhibit anisotropic shrinkage during sintering, requiring predictive modeling. Interrupted sintering and dilatometry enabled calibration of phenomenological models to map densification and shrinkage, supporting distortion compensation for high-accuracy biomedical parts115). Binder jet 3D printing with fine powder mixing doubled the capillary force and green body strength, enabling Ti–6Al–4V parts with 95.2% density, 316 HV hardness, and 589 MPa yield stress after sintering. This demonstrates BJ3DP's potential for producing high-performance lightweight biomedical parts116). Furthermore, Hot isostatic pressing (HIP) of binder-jetted Ti–6Al–4V significantly improved mechanical and fatigue properties, with the 850 °C/200 MPa process outperforming conventional HIP. However, oxygen from binder decomposition reduced ductility and fatigue life, highlighting the need for oxygen control for biomedical applications117). Binder jet printed Ti–6Al–4V parts showed nonuniform density evolution during sintering, with lower edge density linked to binder and powder process effects. Understanding these variations is critical for improving densification and consistency in biomedical components118). Drying spherical Ti–6Al–4V powder (6 h at 200 °C) improved flowability and reduced impurities, resulting in dimensional accuracy improvement (±1.5% → 0.3%) in binder jet printed parts. This enhances binder jetting's capability for high-precision biomedical components119). Metal BJ offers opportunities for biomedical Ti implants but achieving high density and purity is challenging. Optimized powder size (~20 µm) and high-temperature sintering (≥1250 °C) enabled Ti–6Al–4V parts with ≥96 % density, fine α+β microstructure, and mechanical properties (880 MPa, 6 % elongation) comparable to MIM parts, meeting ASTM F2885 standards for custom biomaterials120).

Figure 6

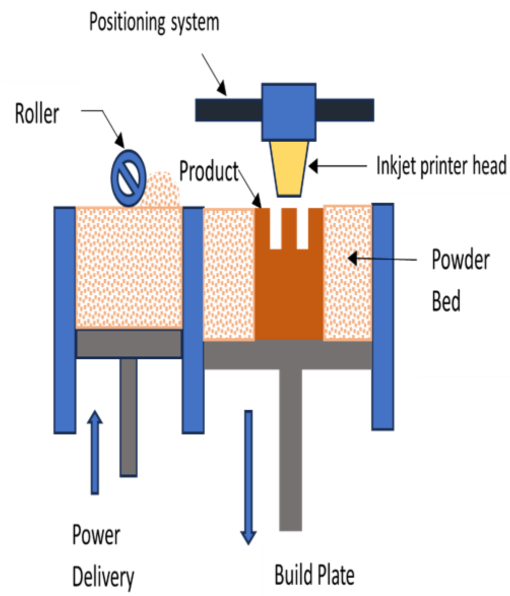

Figure 6 Binder Jetting [121]

Metal BJ enables complex, biocompatible CoCrMo parts for biomedical and industrial applications. Shell printing and sintering at 1325 °C achieved a density of 7.88 g/cc, hardness 18.5 HRC, and UTS 520 MPa with Cr-rich and Mo-carbide phases, making it suitable for high-performance medical implants122). BJ enables near fully dense Co-Cr-Mo parts with mechanical properties comparable to cast alloys. Sintering and aging improved hardness and strength (522 HV, UTS 854 MPa) due to carbide and ε-phase formation while maintaining high biocompatibility (95% cell viability) and supporting biomedical implant applications123). BJ enables Ni–Ti SMA implants with tunable porosity and functional phases. Higher sintering temperature (1185 °C) reduced pores (5 %), increased hardness (763 Hv), and introduced Ni_4Ti_3, while maintaining good cell adhesion, supporting medical implant applications124). BJ of NiTi enables complex shapes, but hydrogen sintering causes elemental segregation and void formation degrading mechanical integrity and sintering to maintain performance for medical applications125). Hybrid BJ with automated machining enables fully digital production of Mg implants. Sintered parts achieved 87 % density, ±0.2 mm accuracy, and < 1.3 µm surface roughness, supporting scalable personalized biodegradable implant manufacturing126). Binder jet printing (BJP) was used to fabricate porous Mg–Zn–Zr structures for potential biodegradable implant applications. These porous BJP samples exhibited higher corrosion rates than solid Mg controls, primarily due to their microstructure, with nonuniform corrosion observed inside the pores127). Binder jet additive manufacturing produced highly interconnected porous Mg–Zn–Zr structures with ~13.3% porosity, >95% interconnectivity, and mechanical properties comparable to human cortical bone exhibiting excellent cytocompatibility and improved cell proliferation, highlighting their potential for bone fixation and bone graft applications128). The processing of WE43 alloy using binder jet additive manufacturing followed by full liquid-phase sintering was explored to achieve near-dense WE43 components (~2.5% porosity) while significantly reducing sintering time and improving both mechanical performance and corrosion resistance129). Stepwise sintering has also been proposed to overcome challenges of oxide-layer-limited diffusion in BJ of Mg alloys. Applied to AZ91D, this refined strategy achieved optimal densification (91.77%), compressive strength (202.33 MPa), and tensile strength (89.45 MPa) under the TH4 profile, showing significant improvement over conventional two-step sintering130). Binder jetting AM has also been explored for functional surface modification. A novel additive micro-texturing approach was used to fabricate SS 316L parts with circular micro-textures (<250 μm feature size, 7.62% dimensional error). These textured surfaces exhibited improved wettability, with up to 72% higher surface free energy and enhanced scratch resistance, indicating potential for biomedical applications requiring hydrophilic and wear-resistant surfaces131). Biocompatibility assessment of BJP 316L SS porous structure has also been explored, focusing on enhancing Osse-ointegration through surface modification132). To address environmental and safety concerns associated with phenolic-based binders, an eco-friendly PVA-based binder was developed for BJ of SS316L. This water-rich binder improved powder layer bonding and enabled the fabrication of fully dense parts with low carbon and oxygen residuals133). Table 3 summarizes the research work covered on the fabrication of biomedical implants by BJ.

Table 3

Table 3 Summary of Work on Binder Jetting

Material / Alloy System	Additive Manufacturing Process	Study Focus / Application	Key Findings	Mechanical Properties / Performance Highlights	Clinical / Functional Relevance	Reference
Ti-6Al-4V	BJ 3D Printing + Sintering	Effect of powder size and processing on microstructure & mechanical properties for biomedical implants	Optimized particle size (~20 μm, wide distribution) and sintering (≥1250 °C) achieved ≥96% pore-free density, fine α+β microstructure, low impurity content, and mechanical properties comparable to MIM Ti	Tensile strength 880 ± 50 MPa, elongation 6 ± 2%	Enables ASTM F2885-compliant high-density Ti alloy for dental & orthopedic implants	[114]
Ti-6Al-4V	BJ 3D Printing + Sintering	Densification modeling and anisotropic shrinkage	Interrupted sintering and dilatometry enabled mapping of densification behavior and anisotropic shrinkage	Provides predictive capability for density evolution and shrinkage compensation	Enables accurate design for biomedical implants with reduced distortion	[115]
Ti-6Al-4V	BJ 3D Printing + Sintering	Improved powder bed properties and part performance	Fine powder mixing increased capillary force (8.35 → 16.27 kPa), boosting green body strength (1.5 → 3.21 MPa); sintered at 1420 °C yielded 95.2% density, 316 HV hardness, and 589 MPa yield stress	95.2% relative density, microhardness 316 HV, yield stress 589 MPa	Enables lightweight porous Ti–6Al–4V parts with improved performance for biomedical use	[116]
Ti-6Al-4V	(BJ3DP) + Sintering + HIP	Effect of HIP on mechanical and fatigue properties	HIP at 850 °C/200 MPa (HPLT) improved mechanical and fatigue properties more than conventional HIP; oxygen from binder decomposition	Higher fatigue strength and density vs. as-sintered, but limited by oxygen pickup	----------------	[117]

			reduced ductility and fatigue life			
Ti–6Al–4V	BJP + Sintering	Effect of sintering on density evolution and process-related variations	Density was lower at edges and higher in curved regions due to binder impact, spreading, particle disruption, and powder bed effects (printhead speed, satellite particles, pressure variation)	Identified process-driven nonuniform densification	Highlights need for process control to achieve uniform, high-density biomedical parts	[118]
Ti–6Al–4V	BJP	Powder drying strategy to improve flowability and accuracy	Optimized drying (6 h at 200 °C) reduced water content and improved flowability, enabling dimensional accuracy improvement from ±1.5 % to 0.3 % and better powder bed quality	Improved dimensional accuracy and powder spreading	Supports high-quality, precise biomedical components via binder jetting	[119]
Ti–6Al–4V	BJ3DP + Sintering	Effect of particle size and sintering temperature on density and microstructure	Binder jetting produced high-density Ti–6Al–4V parts (~96–97 % PFD). Optimized particle distribution (~20 µm) and sintering (> 1250 °C) minimized pores, yielding properties comparable to metal injection molding.	Tensile strength 880 ± 50 MPa, elongation 6 ± 2 %	Custom biomedical implants meeting ASTM F2885	[120]
Co–Cr–Mo	(BJT/M) + Sintering	Properties of printed biomedical-grade Co–Cr–Mo alloy	Binder jetting produced near fully dense Co–Cr–Mo parts (7.88 g/cc). Hardness and UTS were comparable to cast parts, with intergranular Cr-rich and Mo-rich carbides detected.	Hardness 18.5 ± 1.8 HRC, UTS 520.5 ± 44.6 MPa	Biomedical applications requiring high strength and corrosion resistance	[122]
Co–Cr–Mo	BJ + Sintering + Aging	Effect of sintering temperature and aging on microstructure & properties	Near-fully dense parts (99.1 %) with enhanced hardness and UTS after aging. Formation of Cr23C6, CrMo, and ε phase increased strength but decreased ductility; good fibroblast cell viability (95 ± 2 %).	Hardness 522 HV0.1, UTS 854 MPa, YS 641 MPa	Suitable for orthopedic/dental implants	[123]
NiTi (Shape Memory Alloy)	BJ+ Solid-State Sintering	Effect of sintering temperature on porosity and phase formation	Higher sintering temperature (1185 °C) reduced porosity (5 %), formed Ni4Ti3, and improved hardness. Functional martensitic transformation	Nanohardness 8.1 GPa (763 Hv), martensite start 34 °C	Shape memory medical implants and tissue engineering	[124]

			detected in designed thin-strut structures; good biocompatibility confirmed.			
NiTi (Shape Memory Alloy)	BJ + Hydrogen Sintering	Effect of hydrogen on microstructure and elemental segregation	Hydrogen promoted grain boundary segregation and void formation. Reduced mechanical integrity highlights need for optimized sintering atmospheres.	Mechanical strength reduction due to voids	Reliable NiTi implants with improved processing control	[125]
Magnesium (Mg)	BJ + Automated Dry Machining	Digital manufacturing of biodegradable Mg implants	Hybrid binder jetting with automated machining enabled fully digital Mg implant production. Achieved ±0.2 mm accuracy, 87 % density, and < 1.3 µm surface roughness.	15.2 % shrinkage, defect-free machined surfaces	Personalized biodegradable Mg implants and scaffolds	[126]
Mg–Zn–Zr	BJP	Corrosion performance of porous Mg implants in body fluids	Porous BJP samples corroded faster than solid Mg, mainly due to their microstructure. Hydroxyapatite or polymer impregnation improved corrosion resistance.	Porous parts showed nonuniform corrosion behavior	Enhancing corrosion resistance is crucial for biodegradable Mg implants durability	[127]
Mg–Zn–Zr	BJP + Sintering	Porous Mg alloy for bone tissue engineering applications	Produced ~13.3% porosity with >95% pore interconnectivity and excellent cytocompatibility, outperforming cast Mg alloys.	Tensile strength ~130 MPa, yield strength ~100 MPa, elastic modulus ~21.5 GPa, compressive strength ~349 MPa (matching cortical bone range)	Suitable for bone fixation, craniomaxillofacial implants, and bone graft substitutes	[128]
WE43 (Mg–4Y–3RE–0.7Zr)	BJP + Full Liquid Phase Sintering	Reduce sintering time and improve density for biomedical/structural use	Achieved ~2.5% porosity with reduced sintering duration; oxides (Y_2O_3, Nd_2O_3) enabled shape retention and strong sintering neck formation	Improved corrosion resistance and mechanical properties via near-dense microstructure	Suitable for load-bearing biomedical and structural components	[129]
AZ91D (Mg–Al–Zn)	BJP + Stepwise Sintering Process	Overcome diffusion barrier for improved densification	Optimal TH4 sintering gave 91.77% density, 202.33 MPa compressive strength, and 89.45 MPa tensile strength; improved shape fidelity	+14.7% density, +41.9% compressive strength, +106% tensile strength vs. two-step sintering	-----------	[130]
Stainless Steel 316L	BJP + Micro-Texturing	Fabrication of micro-textured hydrophilic parts	Achieved <250 µm periodic features (7.62% error) with improved surface free energy (+72%) and scratch resistance	Contact angles 85.7°–82.5° (water), 50.2°–36.7° (HBSS)	Enhanced hydrophilicity and durability for biomedical surfaces	[131]

Stainless Steel 316L	BJP + DLC Coating	Corrosion and cytotoxicity for porous implant design	Low cytotoxicity for both coated and uncoated porous structures; minimal effect from DLC coating due to thickness variation and mechanical damage	Maintained porosity and structural integrity	Highlights importance of coating optimization for complex implant surfaces	[132]
Stainless Steel 316L	BJP + PVA-based Eco Binder	Sustainable binder for dense structural parts	Fully dense parts with low carbon/oxygen residuals and fine equiaxed grain structure achieved	Isotropic UTS = 535 MPa, elongation = 53%	Demonstrates environmentally friendly, high-performance binder jetting	[133]

4.3. EXTRUSION BASED 3D PRINTING

Extrusion-based (EB) 3D printing technique has become a prominent and promising fabrication method. This technique can precisely develop complex and intricate geometries through CAD and due to this it has widespread acceptance, user-friendly nature, and use of various solidification techniques134). Though, it is significant to note that this technique requires materials with specific printability characteristics135,136). Compared to other available alternative 3D printing methods the use of this 3D printing method is potentially less challenging and generally more cost-effective. Additionally, in this technique the fabrication as well as the geometrical parameters can be easily corrected to fulfill the specific scaffold requirements such as attaining a high modulus and assuring structural integrity.

Figure 7

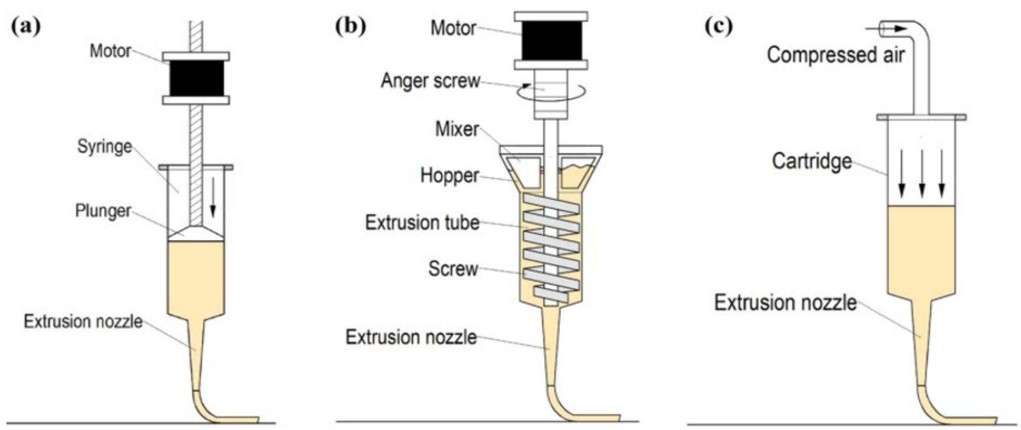

Figure 7 Extrusion Based 3d Printing [137]

Structural integrity is enhanced by using cylindrical fibers in layer-by-layer fabrication as compared to other 3D printing techniques, like droplet or inkjet-based fabrication. The complexity involved in replicating tissues or providing structural support often requires the use of multiple materials, which leads to the needfulness for multi-material extrusion138,139). Multi component systems have the capability to generate interfacial tissues in various biological structures, including organogenesis, vasculature, muscle, and bone. Combining EB 3D printing with subsequent debinding and sintering processes is indeed a robust approach for producing porous scaffolds, especially for applications in tissue engineering and regenerative medicine. Which is beneficial when materials are rigorous to process using other available AM methods are dealt. For instance, porous iron scaffolds with a specific lay-down pattern were successfully fabricated using extrusion-based technique of 3D printing.

These scaffolds exhibit enhanced mechanical properties like tensile strength and fatigue strength as well as biodegradability that closely resemble natural bone, making them highly promising as bone substitutes139). Extrusion-based AM of Mg–Zn alloys has been explored to overcome the rapid biodegradation seen in pure Mg scaffolds fabricated via similar techniques140).

4.4. NANOPARTICLE BASED JETTING

In 2016, a latest technique was developed in 3D printing called nanoparticle jetting (NPJ). Unlike metals that were used extensively in 3D printing methods, NPJ uses liquid materials in the form of ink to encapsulate particles in the form of powder rather than relying solely on metal powder particles. Metal pieces are broken into nanoparticles and embedded into binder to form ink of uniform mixture141). Metal particles are suspended automatically in the ink in dispersed form and then ejected through the nozzle to build the print layer. This process brings smoothness and leads to improved surface finish of the final product. After printing, excess binder evaporates due to heating, leaving only the metal component. NPJ boasts a precision of around 1 micrometer and operates at approximately 300 °C. It can eject 221 million ink droplets per second, making it five times faster than traditional laser printing methods. NPJ offers cost savings, reduces material waste, and enables the creation of complex shapes with high accuracy and surface quality and eliminates the need for designing and removing composite support structures142). However, its main drawback is lower temperature tolerance compared to conventional metal 3D printing methods.

4.5. DIRECTED ENERGY DIFFUSION

In Direct energy deposition (DED) technique various heat sources are used for creating implants including lasers, electron beams and plasma. Other names of this techniques are: Laser Metal Deposition LMD, Laser energy Net Shaping (LENS), or Laser Cladding (LC), and all these names come under the category of laser-based AM processes143). In this technique, Material, usually in the form of metal powder or wire, is fed into the deposition area where the energy source is focused and is fed through a nozzle and melted by a focused laser beam to create a melt pool on the work piece144). In many DED systems, the motion of the nozzle occurs layer by layer in upward direction, while in other metal AM systems, the work piece is moved downward direction with a stationary nozzle145). This process is more economic compared to methods like LPBF and EPBF due to low generation of waste material for recycling. DED employs a high-power laser, ranging from 4 to 10 kW. To avoid material oxidation during 3D printing, the oxygen level in the inert chamber can be reduced to less than 10 ppm146). DED has been utilized to fabricate Ti_6Al_4V/diamond metal matrix composites to overcome Ti alloys low thermal conductivity. A high diamond retention (up to 29 wt%) was achieved during laser processing, resulting in a 200% thermal conductivity increase at 400 °C, along with improved compressive strength and modulus. These 3D printed composites present significant potential for advanced aerospace and biomedical applications requiring high thermal and mechanical performance147). To improve the mechanical and biological performance of dental-grade Ti-6Al-4V (TC4), DED was used to fabricate TC4/ZrO_2 composites with varying ceramic content148). To enhance the poor impact characteristics of Ti alloys, a multilayer Ti-6Al-4V/Cp-Ti composite was fabricated using laser DED149). Study was conducted by using Laser (DED) to fabricate Ti–xCr alloys (6–15 wt% Cr) to examine phase evolution and mechanical performance near the eutectoid composition150). The tribological characteristics were improved by laser DED by applying Hydroxyapatite (HA) on Ti, Al, and V biomaterial alloy151). Likewise, several AM techniques such as laser beam DED, wire and powder based DED, and hybrid DED are frequently utilized to improve the corrosion, interface adhesion and surface quality respectively152-154). In order to join dissimilar metals, better machine ability and thermal performance of material, laser DED techniques is very frequently applied due to its ability of tailoring metal characteristics155-158).

Table 4

Table 4 Summary of Work Covered on DED

Material / Alloy System	Additive Manufacturing Process	Study Focus / Application	Key Findings	Mechanical Properties / Performance Highlights	Clinical / Functional Relevance	Reference
Ti_6Al_4V/Diamond Metal Matrix Composite	DED	Improve thermal conductivity and strength	Achieved 29 wt% diamond retention; thermal conductivity increased by 200% at 400 °C	Enhanced compressive strength and modulus at up to 20–30 wt% diamond loading	Potential for aerospace and biomedical applications	[147]

Ti-6Al-4V/ZrO_2 (TC4 Composite)	DED	Dental implant composite with nano-ZrO_2	Hardness increased with ZrO_2 content (308 → 485 HV); corrosion resistance slightly decreased in acidic/F^- media	Maintained corrosion resistance; improved hardness via grain refinement	Biocompatible and suitable for dental implants	[148]
Ti-6Al-4V/Cp-Ti Multilayer	Laser DED	Improve impact toughness using graded layers	Improved energy dissipation and crack deviation at layer interfaces; enhanced impact strength	Intermediate strength and ductility compared to homogeneous Ti alloys	Potential for impact-resistant biomedical and aerospace components	[149]
Ti–xCr (6–15 wt%)	Laser DED	Effect of Cr content on Ti alloy properties	Suppressed eutectoid reaction; α+β microstructure refinement; ω phase-induced brittleness at eutectoid composition	Ti–10Cr: UTS = 1042 MPa, elongation = 7.1%; Ti–15Cr: UTS = 941 MPa, elongation = 20.3%	Tunable strength–ductility balance for biomedical and aerospace applications	[150]
Ti_6Al_4V + LaB_6 + Hydroxyapatite (HA)	Laser DED	Improve bio-tribological properties	La_2O_3 and TiB	Ti_6Al_4V + LaB_6 + Hydroxyapatite (HA)	Laser Directed Energy Deposition	[151]
NiTi vs. Ti-6Al-4V	Laser Beam DED (LENS)	Synergistic wear–corrosion behavior for implants	NiTi showed lower corrosion tendency and ion release risk than Ti-6Al-4V	Improved wear resistance in both alloys under tribocorrosion	Suggests NiTi as a promising implant material alternative	[152]
NiTi + CuSn10 + SS316L (Dissimilar)	Wire- and Powder-based DED	Dissimilar metal joining using CuSn10 interlayer	Prevented Fe–Ti intermetallic formation; fishbone and dendritic interface morphologies formed	NiTi–CuSn10 interface microhardness = 529 $HV_{0.2}$	Enables reliable joining for biomedical and industrial applications	[153]
316L Stainless Steel	Additive/Subtractive Hybrid Manufacturing (DED + Milling)	Improve machinability & surface quality of AM parts	Grain refinement increased hardness & tensile strength; deeper work hardening observed during milling	Good surface quality and dimensional accuracy	Suitable for high-precision functional components	[154]
AlSi10Mg + Inconel 625 (Ni/Al)	Laser DED	Dissimilar metal joining & interface	Low VED reduced IMC thickness (Al_3Ni_5, NiAl) and interfacial cracking	Tensile interfacial strength = 11–34 MPa	Beneficial for multi-material Ni–Al joint structures	[155]
Ti-Zr-Nb β-Ti alloy (with/without Mo)	Laser DED	Fabrication of β-Ti alloys for biomedical	Formation of ω phase due to Nb clustering; cellular	High strength (~1 GPa), elongation (~15%), low elastic modulus (~65 GPa)	Suitable for orthopedic implants requiring high	[156]

		load-bearing implants	microstructure observed without dendritic segregation		strength and low stiffness	
Inconel 718 – W7Ni3Fe bimetallic structure	DED	Enhancement of thermal and mechanical properties for high-temperature applications	100% improvement in thermal diffusivity; extensive alloy mixing due to laser remelting	100% increase in yield strength; 50% reduction in elastic modulus vs. Inconel 718	------------	[157]

5. OTHER TECHNIQUES

5.1. LAMINATED OBJECT MANUFACTURING (LOM)

In addition to the commonly used techniques for fabricating metal implants, other additive manufacturing methods have emerged to support their biomedical applications. One such technique is metal sheet lamination, also known as laminated object manufacturing (LOM), which is a cost-effective 3D printing method for metallic and ceramic materials158). In this technique, rolled metallic sheets coated with adhesive substances are employed to create 3D structures159). The LOM technique offers several advantages over other AM methods. It eliminates the need for supporting materials and allows for the straightforward 3D printing of multi-metal parts by alternating different sheets in each layer160). Despite its advantages; LOM has some limitations that affect its scalability. It generates a significant amount of waste material and has a relatively low printing speed, particularly for complex objects. Additionally, there is a high risk of layer delamination in printed components that are exposed to harsh environments161).

5.2. METAL INJECTION MOLDING (MIM)

Metal Injection Molding (MIM) holds promise for producing dental implants because it can handle the complex geometries and quality requirements of these components. Various studies have demonstrated the effectiveness of the MIM process for manufacturing implants made from SS, Ti, and Mg17). In a recent work a porous pure iron implants manufactured via MIM using a novel eco-friendly natural rubber-based binder. The implants demonstrated good mechanical properties and were found to be cyto-compatible, hemo-compatible, and biocompatible in both in vitro and in vivo studies, making them suitable for biomedical use162). A novel intravascular stent made from 316L stainless steel was evaluated using MIM, focusing on carbon impurity effects. The stents exhibited no cytotoxicity, good mechanical stability, and rapid endothelialization in animal models, indicating strong potential for clinical applications163). Powder injection molding (PIM) of pure iron for manufacturing endovascular orthoses was explored. The results showed that mixtures above the critical powder volume fraction improved porosity and mechanical properties, making the process suitable for producing biodegradable stents164). A comparison was made for Mg hip stents fabricated via modified injection molding and via conventional machining. Machined stents showed slower degradation, better calcium phosphate deposition, and improved biocompatibility, especially with paraffin canal-filling, indicating their potential for orthopedic applications requiring controlled biodegradation165). A chitosan–zinc composite was introduced fabricated using injection molding and via galvanization, for use in biodegradable flexible implant circuits. The composite exhibited strong biocompatibility, structural stability, and effective electrical conductivity addressing common issues with unstable connections in implantable electronics166). An optimization of MIM parameters was carried out for producing facial orthopedic implants using Catamold 316L feedstock. Simulation using Sigmasoft identified optimal settings—180°C melt temperature, 1s fill time, and 700 bar pressure—to prevent injection defects and ensure complete mold filling167). A short-term cytotoxicity and genotoxicity was carried out for porous NiTi (pNiTi) dental implants produced by MIM. The implants were found to be cytocompatible and genocompatible across multiple assay systems, meeting ISO 10993 standards, and supporting their safe use in biomedical applications168). MIDC developed a jawbone connector implant prototype using the MIM process to reduce reliance on imported medical devices in Indonesia. The precision mold and injection process successfully produced implants meeting geometric specifications, confirming the suitability of MIM for small-scale biomedical components169). Examination of porous NiTi implants was carried out fabricated by

MIM for corrosion resistance, toxicity, and in situ tissue response. The implants showed excellent pitting corrosion resistance, no signs of systemic toxicity or skin reactivity, and demonstrated Osseo-integration comparable to Ti6Al4V, confirming their biocompatibility for long-term implantation170). A comparison was made for low-modulus β-Ti alloys processed via SLM and MIM for biomedical use. SLM yielded dense structures with high strength, ductility, and low modulus, while MIM showed similar strength-to-stiffness ratios but reduced ductility due to Ti_2C formation from carbon pickup and slower cooling171). Development of personalized, drug-eluting inner ear implants using micro injection molding (μIM) was performed with medical-grade silicone and Dexamethasone demonstrating successful manufacturing, sustained drug release (~10% over 6 weeks), high biocompatibility, and anti-inflammatory effects, indicating strong potential for long-term inner ear therapy172). Development of hybrid porous Ti6Al4V dental implants was carried out using the MIM process. Sintering time affected porosity and hardness, while increased surface roughness enhanced shear bond strength, promoting better osseointegration and reducing stress shielding in jawbone applications173). An investigation was carried out to assess the mechanical, corrosion, and tribological performance of a low-carbonCo–Cr–Mo alloy fabricated using optimized MIM. The alloy showed excellent wear resistance, fatigue behavior, and corrosion resistance suitable for biomedical use174).

Table 5

Table 5 Summary of Literature Covered for MIM

Material / Alloy System	Additive Manufacturing Process	Study Focus / Application	Key Findings	Mechanical Properties / Performance Highlights	Clinical / Functional Relevance	Reference
Pure iron (with natural rubber binder)	MIM	Fabrication and evaluation of biodegradable implants	Eco-friendly binder used; good cell and tissue compatibility confirmed	Adequate density, microhardness, strength, and stretchability	Promising for biodegradable implants due to mechanical integrity and biocompatibility	[162]
316L stainless steel	MIM	Development of vascular stents for clinical use	Carbon content affects corrosion resistance and strength; stents showed excellent biocompatibility and stability	Stable form, structure, and rapid endothelialization	Suitable for clinical use as intravascular stents due to good safety and performance	[163]
Pure iron	PIM	Development of injectable iron for endovascular orthoses	Mixtures above critical powder volume enhanced structure and performance	Improved porosity, mechanical and surface properties vs. fusion-processed iron	Promising for biodegradable stent applications	[164]
Magnesium (Mg)	Modified Injection Molding & Machining	Fabrication of biodegradable hip stents	Machined stents with paraffin-filled canals showed improved corrosion resistance and biocompatibility	Slower degradation, better calcium phosphate deposition, enhanced cell compatibility	Promising for orthopedic implants requiring controlled biodegradation	[165]
Chitosan–zinc composite	Injection Molding with Galvanization	Biodegradable flexible implants and electronic circuits	Stable chitosan-zinc bonding solved connection instability; high signal transfer quality	Shear-thinning, structurally stable, low resistance; suitable for signal conduction	Promising for next-gen bioresorbable implantable electronics	[166]
316L stainless steel (Catamold)	MIM with simulation	Facial orthopedic implants and injection defect prevention	Simulation optimized injection parameters to prevent incomplete fill defects	Optimal settings: 180°C melt temp, 1s fill time, 700 bar pressure	Improved quality and reproducibility of facial implants via defect-free MIM process	[167]
Porous NiTi (pNiTi)	MIM	Dental implants and short-term	No cytotoxicity or genotoxicity; met ISO	High cell viability (71–94%), no DNA	Safe for short-term biomedical implant	[168]

		biosafety evaluation	10993 Parts 3, 5, 33 standards	damage, no mutagenicity	applications; promising for dental use	
Ti alloy	MIM	Development of precision jawbone connector plate	Precision mold successfully created; geometry matched target dimensions (2 × 20 × 0.5 mm)	Prototype demonstrated dimensional accuracy through molding and testing	Supports local production of jawbone implants to reduce import reliance	[169]
Porous NiTi	MIM	In vivo corrosion, toxicity, and tissue response	No systemic toxicity, erythema, or inflammation; strong corrosion resistance and osseointegration	High pitting corrosion resistance; low biological reactivity; stable in tissue	Suitable for long-term implantation; comparable to Ti6Al4V in osseointegration	[170]
Low-modulus β-Ti alloy	SLM vs. MIM	Comparison of mechanical behavior for biomedical applications	SLM produced ductile, low-modulus structures; MIM had high strength but reduced ductility due to Ti_2C formation	SLM: high ductility, low modulus; MIM: high strength, reduced ductility	Guides process selection for implants balancing flexibility, strength, and fracture resistance	[171]
Medical-grade silicone with 10 wt% Dexamethasone	Micro Injection Molding (µIM)	Drug-eluting implant for inner ear therapy	Personalized implants manufactured with precision; sustained drug release and bioefficacy	Released ~8.2 µg (10%) over 6 weeks; ~80% cell viability; TNF-α reduction observed	Promising for long-term drug delivery in human inner ear treatment	[172]
Ti6Al4V (hybrid porous)	MIM	Dental implant design to enhance osseointegration	Sintering time influenced porosity and hardness; surface roughness improved shear bond strength	Max shear strength: 1.54 MPa at Ra = 2.37 µm; successful fabrication at 180 °C mold temp	Enhances bone ingrowth and reduces stress shielding for dental implants	[173]
Co–Cr–Mo (low carbon)	MIM	Evaluation of mechanical, corrosion, and tribological behavior	Optimized MIM parameters led to dense parts with good mechanical/corrosion balance	High fatigue resistance, good tensile and bending strength, excellent wear and corrosion resistance	Suitable for biomedical implants due to stable in-body performance	[174]
Mg-0.5Ca alloy	MIM	Processing, sintering, and property evaluation	Achieved metallic form with Mg_2Ca phase at 600 °C after 5 h sintering	Avg. hardness 49.9 $HV_{0.1}$; grain boundaries well-formed	Promising for biodegradable implants with tailored degradation	[175]
Mg WE43 alloy	MIM	Introduction of WE43 to binder-based PM & 3D-printing	High strength and ductility observed post heat treatment; FGF-compatible	UTS up to 226 MPa, 7.6% elongation; fine microstructure	Suitable for future 3D-printed, patient-specific biodegradable implants	[176]
Magnesium powder	MIM	Biodegradable metal implant development	Rheological behavior of Mg/binder feedstocks evaluated; good flow at optimized loading	Feedstocks with 0.65–0.67 powder loading showed ideal flow for MIM	Essential for biocompatible, degradable implant processing	[177]
AZ31 Magnesium alloy	MIM	Feedstock optimization for biodegradable implant production	Optimal powder loading of 65 vol% led to best green part properties	Improved density, dimensional accuracy, and strength at 65 vol%	Supports accurate fabrication of biodegradable Mg-based implants	[178]

Mg-0.5Ca alloy was successfully fabricated using metal injection molding followed by sintering at 600 °C for 5 hours. The resulting microstructure revealed Mg_2Ca phase formation, distinct grain boundaries, and a hardness of 49.9 HV, indicating suitability for biodegradable implant applications175). For the first time, WE43 Mg alloy was processed using MIM to enable its adoption in binder-based powder metallurgy and additive manufacturing. Heat-treated samples showed excellent mechanical properties, with UTS up to 226 MPa and 7.6% elongation, making WE43 promising for patient-specific biomedical implants176). Formulation and rheological analysis of Mg powder feedstocks for biodegradable implants was carried out using MIM177). Effect of powder loading on the properties of AZ31 Mg alloy feed stocks used in MIM was examined. Among the tested loadings (63%, 65%, 67% vol), 65% provided the best balance of density, dimensional accuracy, and strength in the green part178). Table 5 shows the summary of significant of MIM.

5.3. WIRE ARC ADDITIVE MANUFACTURING (WAAM)

WAAM is a type of DED AM technique and WAAM is the most popular AM technique among the other metal AM techniques due to low material cost and good deposition efficacy179). In WAAM, a metal wire is used as a raw material (feed stock) and an electric arc or plasma act as heat source is used to melt down the feed stock wire to fabricate a part using layer-by-layer approach180,181). WAAM is capable of manufacturing several types of metallic biomaterials like Ti, SS, Aluminum with low complexity182).WAAM offers benefits such as high deposition rates; reduced material wastage, and cost-effectiveness183). Three metal transfer modes in Ti-WAAM using GMAW for complex Ti-alloy part fabrication was examined. Controlled short-circuiting metal transfer showed superior geometry preservation stability184).The feasibility of fabricating degradable Zn components via WAAM was demonstrated and showed comparable microstructure, hardness, corrosion behavior, and biocompatibility to wrought Zn, supporting its potential for bioresorbable implant applications185). The effect of welding current and travel speed on the bead geometry and microstructure in WAAM of 308L SS was examined186).The impact of Laser Shock Peening (LSP) on WAAM-fabricated SS316L bone staples was investigated. LSP enhanced tensile strength, corrosion resistance, and significantly reduced bacterial colonization, indicating improved mechanical and antibacterial performance for medical implant applications187).It was demonstrated that WAAM-processed nitinol, after optimized heat treatment, achieves super elastic recovery (~98%) comparable to wrought alloys. Enhanced microstructure and biocompatibility make it a promising candidate for biomedical implants188). A comparison was made between WAAM-fabricated SS316L and NiTi alloys for bio-implant applications. NiTi exhibited superior tensile and yield strength due to its super elasticity and shape memory effect, showing favorable biocompatibility and antibacterial performance189). Table 6 shows the summary of literature on WAAM.

Figure 8

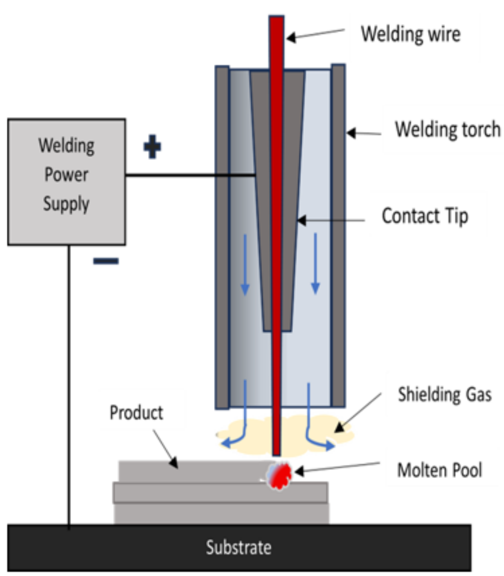

Figure 8 Schematic diagram of WAAM

5.4. FUSED DEPOSITION MODELLING (FDM)

In the Fused Deposition Modeling (FDM) technique a continuous filament made of thermoplastic polymer, like PLA or ABS190), is fed into the heated nozzle where it is melted and after that extruded onto a build platform. The nozzle deposits the molten material layer by layer to form the desired shape191,192). As each layer is laid down, it cools down and solidifies quickly and make bonding with the previous layer. This process continues until the entire shape is complete. FDM technique is widely used for the rapid prototyping due to its simplicity, low cost and its ability to produce complex shapes193). It has limitations in surface finish and mechanical strength compared to other methods194). The feasibility of Metal Fused Filament Fabrication (MF3) for creating patient-specific Ti-6Al-4V maxillofacial implants was examined. Sintered parts exhibited 81% relative density, open porosity for osteo integration, and surface properties suitable for biomedical applications195). The use of FDM for fabricating various biomedical devices using biocompatible PLA/PHA blends was examined. Patient-specific prototypes from CT scans were successfully printed, demonstrating FDM's versatility for hard and soft tissue applications, including stents and nerve conduits196). The importance of monitoring environmental and machine parameters during FDM of biocompatible materials was highlighted. Deviations in nozzle temperature, filament diameter, ambient humidity, and chamber temperature were observed, which can critically impact the quality and usability of biomedical products197). Optimization was performed FDM processing parameters for a novel PLA–stainless steel composite to produce biocompatible implants198). Medical-grade PMMA in FDM for fabricating porous and patient-specific implants was investigated. Findings showed that tip wipe frequency, layer orientation, and porosity significantly influenced compressive strength and stiffness, validating PMMA's potential for craniofacial and orthopedic applications199). Wang, Lei, et al. demonstrated the clinical application of 3D-printed PEEK implants for chest wall reconstruction. A multidisciplinary approach enabled the design, simulation, and fabrication of 114 personalized implants using a newly developed FDM technique200). A cost-effective method combining FDM with debinding-sintering for fabricating 316L SS parts was explored. Using Design of Experiments, optimal FDM parameters were identified to significantly enhance tensile strength and elongation of sintered components201). An SS 316L biomedical implants fabricated using FDM, vapor smoothing, and investment casting was investigated. In-vitro tests confirmed good corrosion resistance and biocompatibility, validating the effectiveness of the combined manufacturing approach202). PLA/SS316L composites was investigated to improve mechanical performance while ensuring compatibility with FDM printing. Findings suggest that adjusting metal filler concentration and extrusion temperature can enhance strength and durability in printed parts203). Hybrid manufacturing techniques—combining FDM, chemical vapor smoothing, silicon molding, and investment casting were investigated to produce SS-316L biomedical implants with enhanced surface quality, dimensional accuracy, and hardness204). Developing customized knee implants through additive manufacturing to address joint pain and mobility issues. The implant was fabricated using Fused Deposition Modeling (FDM), with particular attention given to printing time optimization and material consumption analysi205). Recently highlighted that how residual stresses in FDM-printed medical implants critically affect mechanical performance and dimensional accuracy. Taguchi methods were used to optimize process parameters like temperature and layer thickness, particularly for fiber-reinforced composites206). Table 7 represents the summary of literature covered on FDM.

Table 6

Table 6 Summary of Literature Covered on WAAM

Study Focus	Material	Process	Key Findings	Post-processing Effects	Reference
WAAM of CP Ti	Commercially Pure Ti	WAAM with heat treatment and LSP	α + β phases formed; β-phase increased after heat treatment	LSP increased hardness (~225 HV) and enhanced antibacterial properties	[183]
Metal transfer modes in Ti-WAAM	Ti Alloys	GMAW (CMT, short circuiting, self-regulated)	Similar microstructures with α-basketweave and α-colony structures; controlled short-circuiting transfer gave best geometry stability	Geometry preservation not significantly influenced by metal transfer mode	[184]

WAAM of degradable Zn for implants	Commercially Pure Zinc	WAAM vs. Wrought (WR) comparison	Similar hardness (~41 HV), texture variation, and biocompatibility; slight increase in corrosion rate for WAAM	Bioresorbable Metallic Implants	[185]
Process parameter effects in WAAM	308L SS	WAAM with varying current and travel speed	Higher current increases bead width; optimal deposition at 120 A & 25 mm/min; ferrite content varies with speed and current	Customized Orthopedic Implants (e.g., knees)	[186]
LSP post-processing of WAAM bone staples	SS316L	WAAM + LSP	LSP improved strength, refined microstructure, enhanced antibacterial resistance, and reduced biofilm formation	Medical Implants (Bone Staples)	[187]
Superelasticity improvement in WAAM nitinol	Biomedical-grade Nitinol	WAAM with post heat treatment	Achieved 98% superelastic recovery; presence of B2 austenite; improved cell attachment post-treatment	Biomedical Devices & Implants	[188]
Comparative study of WAAM SS316L and NiTi	SS316L & NiTi Alloys	WAAM (Fixed process variables)	NiTi showed superior strength and shape memory; both materials exhibited good biocompatibility and antibacterial properties	Surgical/Bio-Implants	[189]
WAAM of thin-walled 308L SS	308L SS	WAAM (parameter optimization)	Achieved desired weld geometry; austenitic dendrites with ferrite; UTS (532–553 MPa) and elongation (~40–54%) comparable to wrought steel	Industrial Structural Components	[179]
Layer heterogeneity in WAAM-deposited NiTi on Ti	NiTi (Ni50.9Ti49.1) on Ti substrate	GMAW-based WAAM	Ti diffusion altered Ni concentration in lower layers; varying transformation temps; confirmed shape memory effect	Smart Biomedical Implants, Actuators	[181]

Table 7

Table 7 Summary of Literature Covered on FDM

Study Focus	Material	Process	Key Findings	Application Areas	Reference
MF3 for patient-specific maxillofacial implants	Ti-6Al-4V	Metal Fused Filament Fabrication (MF3)	Achieved 81% relative density, 11% open porosity, Ra ~18 µm, suitable for osteointegration and custom fitting	Dentistry, Maxillofacial Implants	[195]
FDM for diverse biomedical devices	PLA/PHA blend	Fused Deposition Modeling (FDM)	Enabled patient-specific printing from CT scans; PLA/PHA suitable for hard tissues; elastomeric PHAs needed for soft tissues	Bone Implants, NGCs, Stents, Tissue Repair Patches	[196]
Process monitoring in FDM of biocompatible materials	Biocompatible thermoplastics	FDM with sensor monitoring	Detected up to 3% nozzle temp deviation, 13.7% filament diameter variation, 6.5 °C chamber temp increase; ensures product validity	Biomedical Device Manufacturing & Quality Control	[197]
Optimization of FDM for PLA–stainless steel composite	PLA + Stainless Steel	FDM	UTS ~69 MPa (45°), toughness 18 kJ/m^2; biocompatibility retained; raster angle affects strength	Biomedical Implants, Bone Scaffolds	[198]
FDM processing of PMMA for porous biomedical implants	PMMA (medical-grade)	FDM	Higher tip wipe frequency and transverse layer orientation improved strength (16 MPa) and modulus (370 MPa); increased porosity reduced mechanical performance	Craniofacial Implants, Orthopedic Spacers	[199]
Clinical translation of 3DP PEEK implants	PEEK	Modified FDM	Designed and implanted 114 personalized PEEK implants; established clinical standards and	Chest Wall Reconstruction	[200]

for chest wall reconstruction			addressed complications through surface modification		
Optimization of low-cost metal 3D printing via FDM and sintering	316L Stainless Steel	FDM + Debinding & Sintering	Optimized flow rate (110%), layer thickness (140 µm), and nozzle temp (240 °C) yielded 513 MPa tensile strength and ~60% elongation	Biomedical, Prosthetics, Surgery	[201]
Fabrication and testing of biomedical implants	SS 316L	FDM + Vapor Smoothing + Investment Casting	Demonstrated good corrosion resistance and biocompatibility in in-vitro tests	Biomedical implants	[202]
Development of PEEK-cHAp composite scaffolds for bone regeneration	Poly-ether-ether-ketone with calcium hydroxyapatite (PEEK-cHAp)	FDM, SBF immersion, cytotoxicity test	Spherical nanoparticle cells promoted microstructural growth; scaffolds showed minimal toxicity and high bioactivity; optimal for bone fracture repair	Bone implants, biomimetic scaffolds for bone healing	[203]
Mechanical properties of PLA/SS316L composites	PLA reinforced with SS316L (2–12% wt.)	FDM 3D printing (0.4 mm nozzle, 200–220°C)	Optimal results at 12% SS316L (52.83 MPa tensile strength, 0.1275 kJ/m^2 impact strength); higher temperatures improved performance	Patient-specific implants, orthopedic devices	[204]
Hybrid manufacturing of SS-316L implants	SS-316L	FDM + CVS + SM + IC	Achieved ISO-compliant dimensional accuracy (UNI EN 20286-I), Cp/Cpk >1.33 for hardness/radial dimensions; NIH-3T3 cell viability confirmed	Biomedical implants, Tissue engineering	[205]
Residual stress minimization in FDM	Carbon fiber-nylon 12 composite	Taguchi-optimized FDM (255°C, 0.3mm, 50mm/s)	Achieved residual stresses of 40.7±7.7 MPa (experimental) matching cortical bone properties	Orthopedic hip implants	[206]

Figure 9

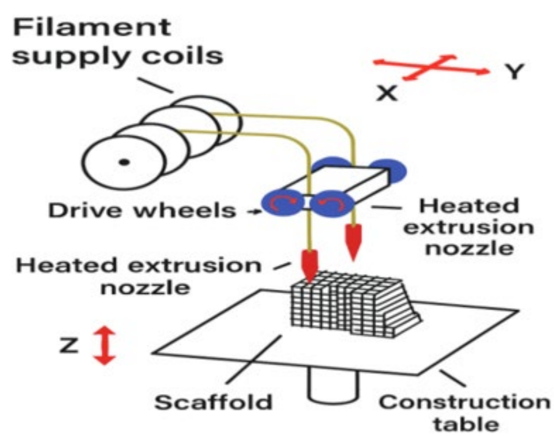

Figure 9 Schematic Diagram Of FDM [**206**]

6. CONCLUSION

Additive manufacturing (AM) methods have revolutionized the fabrication of metallic biomaterials, offering unprecedented opportunities for customization, precision, and performance enhancement in biomedical applications. Techniques such as Directed Energy Deposition (DED), including Laser Metal Deposition (LMD), Laser Energy Net Shaping (LENS), and Laser Cladding (LC), have demonstrated their capability to produce complex geometries, optimize material properties, and enable patient-specific implant designs. The ability to control microstructures and tailor the mechanical and biological properties of metallic implants through these AM processes has been pivotal in advancing the field of bioengineering. The integration of materials like titanium alloys and cobalt-chromium alloys in AM processes has further improved the biocompatibility and durability of implants, catering to the stringent requirements of medical applications. However, challenges remain, particularly in ensuring consistent quality, managing thermal stresses, and achieving the desired surface finish and porosity. Continued research and development in post-processing techniques, as well as in the refinement of AM parameters, are essential to fully harness the potential of these technologies. In conclusion, the adoption of additive manufacturing methods in the production of metallic biomaterials represents a significant leap forward in medical technology. As these methods continue to evolve, they are likely to play an increasingly critical role in the creation of advanced, patient-specific medical solutions, ultimately improving patient outcomes and expanding the possibilities in healthcare.

7. OUTLOOK AND RECOMMENDATIONS

Additive manufacturing (AM) of the metallic biomaterials is advancing rapidly with the increasing demand. It offers new ways to make patient-specific and high-performance implants. Yet, several challenges limit its full clinical use. Researchers must create standard protocols for process control, post-processing, and biological testing. These steps will improve consistency and meet regulatory needs.

Future work should explore new materials beyond standard alloys. Bioresorbable metals, functionally graded materials, and custom alloy designs can provide the right balance of strength, biocompatibility, and controlled degradation. Advanced surface treatments can also boost bone integration and fight infection.

Hybrid manufacturing can combine the strengths of different AM techniques or merge AM with subtractive methods. This approach can produce complex shapes with tailored properties. Using computer modeling, AI, and machine learning can speed up design optimization and predict implant performance before production.

Collaboration is essential. Materials scientists, engineers, clinicians, and regulators must work together to turn lab innovations into safe, affordable, and approved medical implants. With the more research on the fabrication techniques and modification techniques and focus on patient-oriented implants fabrication, AM can transform metallic biomaterial use in healthcare.

CONFLICT OF INTERESTS

None.

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None.

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