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Review paper

Conventional to advanced techniques: Progress in titanium alloys for biomedical applications

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Abstract

The biocompatibility, high strength-to-weight ratio, mechanical properties, and resistance to corrosion of titanium and its alloys, making them suitable for implants. The major factors influencing titanium implant performance depends on fabrication technique, microstructure, resistance to corrosion, tribocorrosion, osseointegration and its stability. This review focuses the impact of conventional manufacturing technique such as welding, casting, powder metallurgy, forging, rolling, and extrusion and their effect on metal implants. To overcome the limitations of conventional techniques, further development of advanced manufacturing strategies capable of achieving complex geometries and tailored properties for next-generation biomedical implants. Specific focus is placed on powder bed fusion, directed energy deposition, material extrusion, binder jetting, and material jetting. The impact of processing technique on mechanical properties, microstructure, corrosion resistance and overall implant longevity is discussed. Furthermore, challenges such as the high elastic modulus of conventional Ti alloys, limited control over the strength, ductility, biocompatibility, surface degradation and tribocorrosion, and the need for adaptive and long-term implant monitoring continue to drive research toward integrated material, computational, manufacturing, and digital strategies. Emerging approaches such as high-entropy titanium alloys, artificial intelligence, machine learning, 4D printing, smart implant technologies, and digital twinning for predicting and controlling implant performance, are emphasised in this review. Their incorporation with additive manufacturing (AM) opens new possibilities for enhancing Ti implants, which can offer improved mechanical compatibility, corrosion resistance, wear resistance, personalized functionality, and consistent long-term performance.

Keywords

Metallic biomaterials; implants; fabrication techniques, tribo/bio-corrosion, additive manufacturing, smart technologies

Introduction

An increase in the prevalence of age-related disorders and the growing incidence of accidental trauma have increased the use of biomaterials to boost the quality and lifetime of implants. Biomaterials are natural or synthetic materials designed for direct contact with living tissue, maintaining compatibility without releasing toxic metal ions or causing adverse effects. One of their most significant applications is in the development of implants, which are used to replace, support, or restore the functionality of damaged tissues and organs. Biomaterial requirements for implants include 1) biocompatibility, 2) biostability, 3) corrosion resistance, 4) good mechanical properties and 6) sterilizability [1] (Figure 1). The classification of biomaterials can be discussed under two main aspects: 1) functional classification, which includes bioinert, bioactive, and bioresorbable materials, and 2) chemical composition, such as metals, polymers, ceramics, and composites. Functional classification depends on the biological interaction of the biomaterial with the tissue surrounding following implantation. The nature of each interaction differs, and this variability influences how tissues respond, integrate, and perform over time. Bioinert materials are defined as those that do not undergo chemical changes, toxic effects, or immune responses when interacting with living tissues. Conversely, bioactive materials interact directly with tissue and bone, facilitating integration and healing. Bioresorbable materials are substances that progressively degrade, get absorbed, or are displaced by tissues that are subsequently generated over time [2]. The choice of a particular type of biomaterial depends on the intended clinical application and the desired biological response.

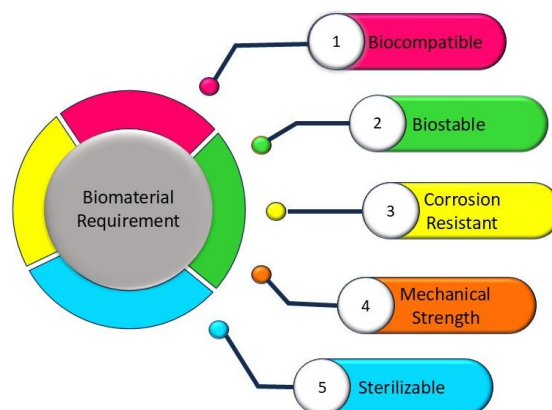


Figure 1. Schematic illustration of biomaterial requirements for titanium implants

Ti and its alloys are essential in biomedical applications due to their distinctive features, such as biocompatibility, corrosion resistance, high strength-to-density ratio, and mechanical properties [2]. Ti is a low-density element (60 % less than that of stainless steel) that can be strengthened by alloying or surface modification; it has high yield strength and high specific strength [3]. It is an allotropic element, having a crystallographic structure that can change depending on temperature [3]. The mechanical properties of various metal alloys are compared with cortical bone, as listed in Table 1.

Table 1. A comparative mechanical property of cortical bone with various metal alloys [2].

Metal alloys	Elastic modulus, GPa	Tensile strength, MPa	Fracture toughness, MPa m ^{1/2}
Cortical bone	10 to 30	35 to 283	3 to 6
Mg-based alloys	37.5 to 65	190 to 250	15 to 35
Stainless steel	193	485 to 860	50 to 200
Ti-based alloys	79 to 110	830 to 1025	55 to 115
Co-based alloys	220 to 230	655 to 1900	100

Conventional implant materials like Co-based alloys and stainless-steel alloys typically exhibit much greater stiffness compared to cortical bone, which can result in stress shielding and potentially lead to bone resorption. While Ti and its alloys provide superior mechanical compatibility owing to their lower modulus of elasticity, along with remarkable corrosion resistance and biocompatibility. Yet, further decreases in elastic modulus via alloy design, porous structures, and advanced manufacturing techniques are still required to enhance load transfer and long-term implant effectiveness. Despite their biocompatibility and corrosion resistance, the bioinert nature of Ti and its alloys can limit osseointegration and slow direct bonding to bone. A naturally forming Ti passive oxide layer on the surface, although typically uniform, can still lead to corrosion and wear; chemical reactions and tribocorrosion may occur simultaneously, potentially releasing metal ions and reducing implant stability. The released ions and surface degradation products influence the local cellular response, including osteosarcoma cells, making it important to understand how Ti alloy fabrication, surface condition, and oxide-layer stability affect both corrosion behaviour and biological interactions.

Scope of review

This review highlights the study of conventional and fabrication techniques for Ti alloys along with their properties. AM is the effective method for producing Ti alloys, leveraging its ability to generate intricate shapes, components made exactly as needed, personalized parts, and minimizing material wastage. Despite these advantages, several obstacles persist to limit its widespread adoption, which includes defects and porosity, residual stresses, surface roughness, limited material compatibility, and the need for careful control of processing parameters. In recent years, machine learning (ML) and artificial intelligence (AI) have become notable tools for addressing some of the challenges through process optimization, defect prediction, quality monitoring, and material property prediction. However, the integration of AI in additive manufacturing is still at a developing stage, and several issues remain, such as the limited availability of reliable and standardized datasets, high computational requirements, model interpretability, and challenges in transferring models between different materials, machines, and processing conditions. Therefore, extensive research is required to develop robust AI-assisted AM frameworks that can effectively link processing parameters with microstructure, defects, and final component performance. Such developments could improve process reliability, reduce experimental and manufacturing costs, accelerate optimization, and contribute to the broader industrial adoption of additive manufacturing.

Review methodology

Earlier pioneering studies included Ti alloys as implants for biomedical applications. Published Scopus country-wise research contributions and their international collaborations literature were identified from 2015-2026, represented in the bibliometric network visualization in Figure 2a, whereas Figure 2b represents various fabrication techniques used for Ti-based alloys for biomedical applications. Advanced fabrication techniques such as AI-assisted, 4D printing, High entropy alloy (HEA) and digital twin technologies are the areas that need to be focused on for better implant performance.

Keywords used for the literature search included combinations of terms related to various fields of study, such as titanium alloys, biomedical implants, fabrication technologies, surface modification, and emerging materials technologies. Some of the keywords were: Ti alloys, biomedical application of Ti alloys, Ti implants, conventional fabrication, additive manufacturing, 3D printing, 4D printing, powder bed fusion, electron beam melting, selective laser melting, directed energy

deposition, material extrusion, binder jetting, corrosion, tribocorrosion, surface modification, osseointegration, high entropy alloys, artificial intelligence, smart implants, and digital twin. To find other relevant published studies that were not discovered through keyword searches, publications were used.

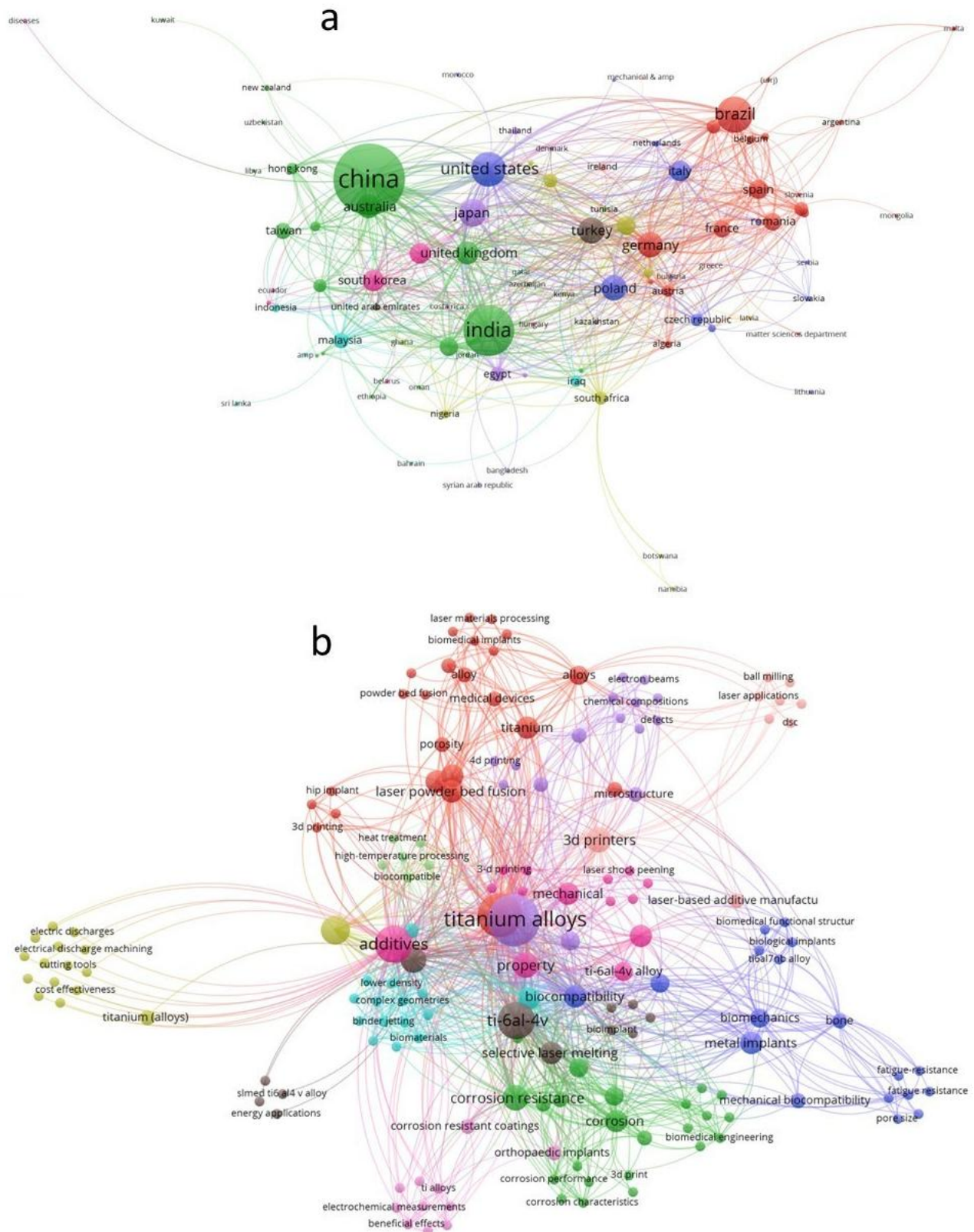


Figure 2. Bibliometric network visualisation (a) country-wise research contributions and international collaborations in the field of Ti alloys for biomedical applications and (b) various fabrication techniques used for Ti-based alloys for biomedical applications

Overview of Ti-based alloys

Conventional Ti alloys

Commercially pure (CP) Ti is unalloyed Ti containing 99.0 to 99.6 % pure Ti by mass. The remaining balance consists of controlled trace quantities of interstitial elements and impurities (oxygen, iron, carbon, hydrogen and nitrogen). As the oxygen content is increased, the strength of CP-Ti improves, while ductility decreases. Ti is the most widely used metallic biomaterial for implants due to its biocompatibility, anti-corrosion and favourable mechanical properties. However, untreated Ti has a bio-inert surface, which limits its ability to form chemical bonds with adjacent bone tissue and, as a result, reduces its osseointegration potential, potentially affecting long-term implant stability. Pure Ti (Grade 1-4) has low mechanical strength when compared to Ti alloy. A key feature of CP-Ti is the spontaneous development of a stable titanium dioxide (TiO_2) passive layer on its surface, which provides excellent corrosion resistance and contributes to its biocompatibility. Because of these properties, CP-Ti is widely used in dental implants rather than load-bearing hip or joint implants. Its comparatively low strength limits its use in many biomedical implants, which has spurred the development of advanced Ti-based alloys. To mitigate this limitation, several Ti alloys have been developed by incorporating alloying elements. The microstructure and properties of these alloys are largely governed by the allotropic transformation behaviour of Ti.

Conventional Ti alloys are classified according to their microstructure and the relative stability of α and β phases. The three principal categories are alpha and near alpha, alpha-beta, and beta Ti alloys [4]. Table 2 compares various Ti alloys based on microstructure and application.

Table 2. Comparative study of various Ti alloys, based on microstructure and their application [5]

Ti-based alloys	α /near- α Ti-alloys	$\alpha+\beta$ Ti-alloys	β -type alloys
Stabilizer	C, N, O, Al, Sn, and Zr	Combination of α and β stabilizer	Mo, V, Nb, Cr and Ta
Crystal structure	HCP (at room temperature)	HCP and BCC	BCC (above 883 °C)
Elastic moduli, GPa	102.7-104.1	114-112	80-84
Yield strength, MPa	170-480	825-1250	200-1600
Tensile strength, MPa	240-550	890-1150	850- 1500
Examples	Cp-Ti grade 1-4, Ti-5Al-2.5Sn, CP-Ti, Ti-8Al-1Mo-1V, Ti-6Al-2Sn-4Zr-2Mo	Ti-6Al-4V (annealed), Ti-3Al-2.5V, Ti-5Al-2.5Fe, Ti-6Al-7Nb	Ti-12Cr, Ti-13Nb-13Zr (aged), Ti-12Mo-6Zr-2Fe (annealed), Ti-15Mo (annealed), Ti-15Mo-5Zr-3Al (aged), Ti-16Nb-10Hf
Properties	Enhanced corrosion resistance, good weldability and high creep resistance	Exhibit good fabricability, high strength	Non-toxic beta stabilizer is used, heat treatable, lower elastic moduli, better Biocompatibility and corrosion resistance
Application	Medical application	Orthopedic joint replacement, screw	Biomedical implants, aerospace and defence
Limitation	Low mechanical strength at room temperature, low fatigue strength	Toxicity of metal ions	High cost

Alpha or near alpha Ti-based alloys

α and near α Ti-alloys represent the earliest generation of Ti biomaterials. These alloys predominantly consist of the α -phase, having hcp crystal structure. Near- α alloys have trace amounts of β -stabilizing elements, with the α -phase being the dominant phase. The α and near- α Ti alloys are characterised by excellent corrosion resistance [5]. In α -Ti alloys, Al serves as the primary α -stabilizing alloying element, enhancing tensile strength and improving resistance to creep deformation. These alloys provide a balanced combination of moderate mechanical strength and

good formability, making them appropriate for manufacturing complex and thin-walled components through cold-forming processes such as deep drawing. However, unlike other Ti alloy classes, α alloys cannot be strengthened through heat treatment; their deformation behaviour is characterised by the hcp crystal structure of the α -phase, where the limited number of available slip systems and grain orientations lead to pronounced anisotropy during plastic deformation [6]. Whereas near- α Ti alloys contain a trace fraction of β -stabilizing elements, enabling them to retain the excellent creep resistance and anti-corrosion properties of α alloys while offering improved strength, fracture toughness and elevated-temperature performance [6]. Overall, α and near- α Ti alloys offer excellent resistance to corrosion and biological compatibility; however, their relatively high stiffness and moderate mechanical strength limit their suitability for demanding orthopaedic applications. This limitation has driven the progress of $\alpha+\beta$ Ti alloys with improved mechanical properties while maintaining favourable biocompatibility.

Alpha+beta Ti-based alloys

Among the different classes of Ti alloys, $\alpha+\beta$ Ti alloys have gained considerable attention for applications in biomedical applications, owing to their favourable combination of mechanical properties and corrosion resistance. The most prominent member of this group is Ti-6Al-4V, which is used broadly in orthopaedic and dental implants for several decades. Around the 1970s and 1980s, the use of Ti and its alloys in orthopaedic implants increased drastically with the development of Grade 5 Ti alloy (Ti-6Al-4V) [7]. This alloy significantly increases strength and resistance to corrosion when compared to commercially pure Ti, making it suitable for load-bearing applications.

Ti-6Al-4V alloys have been extensively utilised as orthopedic implants, but continuous use leads to the elution of toxic metal ions such as Al and V, which is responsible for cytotoxicity and severe bone diseases such as osteomalacia and neurological diseases such as Alzheimer's. This has prompted concern about inflammation and allergic reactions. Consequently, considerable research has now focused on enhancing the manufacturing of Ti-6Al-4V alloys to improve their durability and biocompatibility [8].

To address concerns related to vanadium toxicity, Ti-6Al-7Nb alloys were developed by replacing vanadium with niobium, thereby enhancing Ti corrosion resistance by promoting stability of the oxide layer, enhancing biocompatibility and improving its passive behaviour [9]. Its comparatively low elastic modulus is similar to that of human bone, which gives it importance for orthopaedic implants.

β -Ti alloy

β -Ti alloys have been studied for biomedical applications given their lower Young's modulus and improved biocompatibility compared with Ti-6Al-4V [10]. One of their major advantages is that conventional Ti implants (Ti-6Al-4V), which have an elastic modulus of 110-114 GPa, are stiffer than actual human bone, which has an elastic modulus of 15-30 GPa. This mismatch in stiffness induces stress shielding, which subsequently promotes bone resorption and implant loosening over time [11,12]. To address this limitation, β -Ti alloys containing β -stabilizing elements such as Nb, Ta, and Zr is developed, which reduces the elastic modulus while maintaining adequate mechanical strength.

Nb plays a critical role in stabilizing the beta phase while maintaining adequate mechanical strength. Nb has also been widely studied for its ability to impart superelasticity at room temperature, as studied in Ti-Nb-Zr and Ti-Nb-Ta [13]. By adjusting the alloying components like Nb, Ta, and Zr, transformation temperature and transformation strain can be controlled. Zr was found to increase the beta phase lattice parameter with Nb, which increases its superelasticity, resulting in greater recoverable deformation. Whereas Ta reduced the transformation strain in superelastic

alloys, which may limit flexibility while improving its mechanical stability and strength [14]. The alloying elements Zr, Ta and Nb are non-toxic elements with high resistance to corrosion, enhanced biocompatibility, and high strength.

Properties of Ti alloys

The mechanical properties, corrosion resistance, bio-tribocorrosion, and osseointegration of implants are important factors governing their behaviour in the physiological environment and affect their functionality and durability. Better mechanical compatibility reduces stress shielding, and high corrosion and tribocorrosion resistance, along with good osseointegration, stability, biocompatibility, and clinical success, support long-term performance.

Corrosion behaviour

Corrosion is the chemical or electrochemical reaction of metal alloys with the environment, such as physiological fluids containing oxygen, chloride ions (Cl^-) and hydrogen ions (H^+), which leads to gradual degradation. This phenomenon results in material loss, deterioration of mechanical properties, and reduced functional performance of metal alloys [4]. Although corrosion is a common phenomenon affecting almost all metals, the severity and rate of corrosion vary depending on the metal's chemical nature and its inherent ability to form a passive layer. Some metals, such as Ti, have an inherent property of forming a passive oxide layer when the metals react with or deteriorate in the presence of oxygen [4]. A thin layer of TiO_2 spontaneously forms on the surface of Ti and provides excellent corrosion protection in the presence of oxygen [15]. Although highly resistant to corrosion, the passive film can be damaged under adverse corrosion conditions (acidic environments, high-chloride conditions, mechanical wear). Damage to the passive film may result in metal-ion dissolution, localized corrosion, and reduced implant longevity [15] is represented in Figure 3.

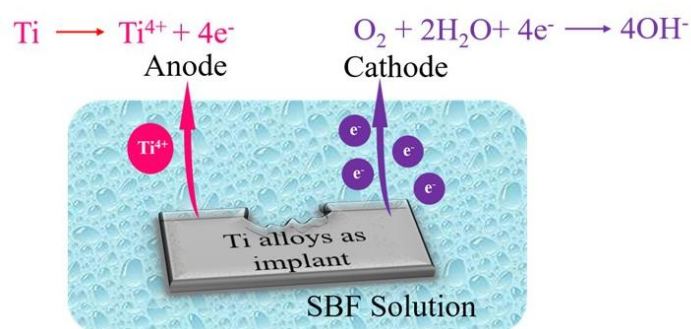


Figure 3. Illustration of corrosion of Ti alloy in simulated body fluid (SBF) solution

Various factors are said to influence the corrosion behaviour of Ti alloys. In addition to alloy composition, temperature, pressure, and halide-ion concentration are crucial for assessing passive-film stability and implant performance. Ti alloys are integrated with elements such as Al, Nb, Mo, Ta and Zr. Among them, Al stands out as the most favoured α -stabilizer, improving the mechanical properties of Ti alloys. It also aids in resistance to hydrogen embrittlement within a concentration range up to 7 %. In contrast, elements like Mo, Nb, Ta, and Zr are β -stabilizers that improve corrosion resistance by stabilizing the passive oxide film. The release of harmful metal ions in a physiological environment has been reported with vanadium, and thus research has focused on substituting vanadium with non-toxic elements [16].

Multiple modes of corrosion can occur in Ti implants, such as galvanic corrosion, a type of corrosion in which two different metals are coupled electrically in the presence of an electrolyte,

resulting in preferential corrosion of the less noble metal. Soares *et al.* [17] studied the galvanic corrosion behaviour of Cp-Ti implants coupled with CoCrMo prosthetic components. The study demonstrated that maintaining the stability of the Ti oxide layer is essential for ensuring long-term corrosion resistance of Ti-based biomedical devices.

Crevice corrosion is caused by stagnant solution in narrow gaps and joints, which destroys the inherent passive oxide film. Crevice corrosion behaviour of Ti-6Al-4V alloy under a simulated inflammatory environment is investigated to check its feasibility for biomedical applications. The inflammatory conditions can alter the electrochemical behaviour of Ti-6Al-4V by promoting localised corrosion processes within crevice regions. The importance of understanding the stability of the passivation layer under physiological and pathological conditions to ensure the long-term reliability of titanium implants. Pitting corrosion is localised corrosion that begins with the failure of the passive TiO₂ film, typically in chloride-rich environments. Golvano *et al.* [18] evaluated the effect of pH on the electrochemical behaviour of titanium. They reported that acidic environments (pH 3.5) reduce the stability of the passive film, thereby weakening its protective function and increasing Ti susceptibility to corrosion. Hydrogen-induced cracking is caused by the diffusion of hydrogen into the alloy, leading to embrittlement and cracking, while stress corrosion cracking results from tensile stress and a corrosive environment. Microbiologically influenced corrosion is linked to the formation of microbial biofilms, thereby altering the environment of the corroded area and accelerating the corrosion process. Furthermore, in load-bearing orthopaedic implants, cyclic loading and repeated articulation break down the passive oxide layer and lead to higher wear, metal ion release and implant failure; corrosion fatigue and tribocorrosion are of special interest here [16]. The elemental composition of a Ti alloy affects the alloy's microstructure, thickness and defect density of the TiO₂ passive film. A lower corrosion current density (j_{corr}), high charge transfer resistance (R_{ct}), positive corrosion potential (E_{corr}), polarisation resistance (R_p) and repassivation, and can produce a stable, uniform, low-defect, self-healing and passive layer on Ti alloys. Electrochemical corrosion parameters and electrochemical impedance spectroscopy (EIS) characteristics of Ti alloys based on fabrication techniques are listed in Table 3.

Table 3. Electrochemical corrosion parameters and EIS characteristics of titanium alloys based on fabrication techniques [16,18]

Ti alloy / fabrication technique	EIS	E_{corr} / V vs. SCE	R_p / kΩ cm ²	j_{corr} / μA cm ⁻²
CP-Ti (Grade 2) / wrought, polished	Increase in R_{ct} , stable TiO ₂ film	≈ -0.24 to -0.16	≈210-600	≈ 0.02 to 0.10
Ti-6Al-4V / cast	more stable passive film than LPBF; coarse α+β, lower porosity	≈ 0.20 to -0.10	≈175-450	≈ 0.04 to 0.12
Ti-6Al-4V / wrought, polished	R_{ct} dominated by compact TiO ₂ ; slight increase in C_{dl} with roughness	≈ -0.21 to 0.12	≈160-400	≈ 0.05 to 0.16
Ti-6Al-4V / LPBF (as-built)	Lower R_{ct} vs. wrought; martensitic α', residual stress, higher defect density	≈ -0.26 to -0.16	≈80-245	≈ 0.10 to 0.28
Ti-6Al-7Nb / cast-wrought	Increase in R_{ct} , Nb stabilizes passive film	≈ -0.20 to -0.10	≈180-480	≈ 0.03 to 0.12
β-Ti (Ti-Nb-Zr-Ta) / wrought-extruded	Nb/Ta/Zr enhance passivity and R_{ct} .	≈ -0.24 to -0.12	≈280-680	≈ 0.02 to 0.08
Ti-Nb-Zr-Ta / powder hot extrusion	Improved R_{ct}	≈ -0.22 to -0.10	≈250-750	≈ 0.02 to 0.07
Ti-based HEA / powder hot extrusion	Stable passive film	≈ -0.20 to -0.08	≈300-900	≈ 0.01 to 0.06
Ti-6Al-4V/ machined	Increase in R_{ct} with optimized depth of cut and speed	≈ -0.18 to -0.10	≈200-450	≈ 0.03 to 0.10
Ti-6Al-4V/ laser-assisted ECM	More dissolution rate; tailored porosity	≈ -0.22 to -0.12	≈150-350	≈ 0.04 to 0.12

Figure 4 compares current density and corrosion rate in Ti alloys, highlighting microstructures resulting from fabrication. Among them, Ti13Nb13Zr alloys display favourable current behaviour, suggesting improved passivity and resistance to corrosion and tribocorrosion, which could enhance osseointegration and prolong implant life. The inclusion of Zr and Nb enhances the chemical stability of the passive layer, leading to decreased electrochemical activity and improved corrosion resistance [19].

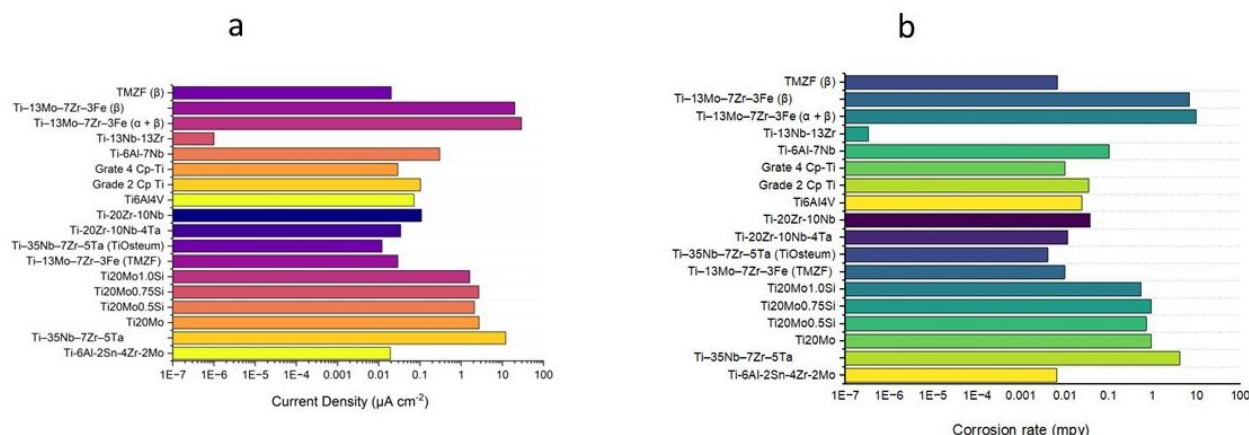


Figure 4. Comparison of (a) corrosion current density and (b) corrosion rate for various titanium alloys

Tribocorrosion

Tribocorrosion is characterized by the material's gradual wear and tear resulting from the concurrent effects of mechanical friction and electrochemical corrosion [20]. This phenomenon is particularly crucial in biomedical applications, where titanium implants are subjected to cyclic mechanical stress in corrosive bodily fluids that include chloride ions, proteins, and dissolved oxygen[21]. In daily activities, for example, hip and knee orthopedic implants, which undergo repeated sliding and loading, are particularly prone to degradation due to tribocorrosion. The disruption of the passive oxide layer during mechanical contact exposes fresh metal surfaces to the physiological environment, as a result accelerating corrosion reactions and increasing material loss [21]. The accumulation of wear particles from the Ti implant can cause periprosthetic inflammation and osteolysis, ultimately resulting in implant loosening.

Gabriel *et al.* [22], reported that localised electrochemical cells can develop between stressed and unstressed regions of Ti alloy, leading to an increase in corrosion. The mechanical stress generated during implantation and service creates electrochemical heterogeneity, resulting in differences in corrosion potential between adjacent regions. This stress-assisted corrosion mechanism promotes localised material degradation and significantly influences the long-term reliability of biochemical implants. Ti-6Al-4V alloy remains the widely utilised titanium alloy because of its good biocompatibility, high specific strength, and favourable corrosion resistance. However, long-term exposure to biological environments may result in the release of metallic ions from the alloy. The dissolution of alloying elements such as aluminium and vanadium has raised concerns about their potential cytotoxicity and other adverse biological effects. Released metal ions can interfere with cellular metabolism, induce inflammatory responses, and contribute to tissue damage surrounding the implant site [23]. Consequently, considerable efforts have focused on expanding vanadium-free and aluminium-free Ti alloys with enhanced biological performance.

Longhofer *et al.* [24] demonstrated that incorporation of niobium into Ti alloys improves passivation behaviour, enhances corrosion resistance, and reduces ion release in physiological

environments. Niobium acts as a strong β -phase stabilizer that promotes the generation of a stable and protective oxide film, thereby improving the performance of Ti-based implants. Terheyden *et al.* [25], studied implant failure, which is associated with a combination of mechanical, biological and electrochemical factors. Common causes include bacterial infection leading to peri-implant inflammation, stress shielding resulting from the disparity between the elastic modulus of natural bone and the prosthesis, inadequate osseointegration, and the generation of wear debris particles. The accumulation of wear debris can cause inflammatory responses and osteolysis, ultimately leading to implant loosening and premature failure. Therefore, improving both the corrosion and tribocorrosion resistance of Ti-based biomaterials continues to be a crucial hurdle in the progression of advanced next-generation biomedical implants. Dziaduszevska and Zieliński [26] provided a thorough review of the microstructures, mechanical properties and biomedical potential of Ti and its alloys. Nonetheless, few studies have primarily examined their tribocorrosion behaviour, despite its substantial relevance for the long-term functionality of implants. *In vivo*, Ti implants are subjected to blood plasma, protein-rich tissues, actively flowing fluids, and other biological media, wherein proteins and biological components can substantially affect both wear and corrosion processes. It is therefore essential to understand the tribocorrosion behaviour of Ti alloys under such conditions to predict implant longevity and ensure clinical success

Osseointegration

Osseointegration is the direct structural and functional bond between the implant surface and the bone tissue. The stability of this bond largely depends on the adhesion of cells to the surface of the implant. After attachment, cells spread, proliferate, and facilitate new bone formation around the implant, aiding in successful integration [27]. Even with a successful implantation, there remains a risk of infection at the site [28]. The osseointegration process consists of several crucial stages: 1) initial contact with the implant, 2) inflammatory and cellular reactions, 3) bone growth, 4) maturation and remodelling. Ti remains the preferred material for biomedical applications owing to its remarkable corrosion resistance, favourable mechanical properties, and superior long-term biocompatibility. A durable, self-passivating TiO_2 oxide layer enables Ti to coexist with surrounding tissues without triggering harsh inflammatory reactions. However, its biocompatibility is not necessarily equivalent to true bioactivity. Although it can passively promote bone apposition, Ti continues to be fundamentally bio-inert and fails to actively signal or stimulate host cellular responses [29]. According to Ogawa *et al.* [30], the limitations of current implant materials are important, particularly under suboptimal biological conditions. Histological examinations consistently show that bone contact of up to 45 to 65 % occurs on the implant surface, indicating that a substantial portion of the implant might not undergo true osseointegration. Consequently, some areas on the Ti surface may serve not just as inert zones but as bio-tolerant surfaces that the body can accommodate without achieving direct integration. Despite their broad clinical use, Ti implants still face challenges, including insufficient osteogenesis, osteoconduction and osseointegration. To tackle these limitations, considerable research has centred on surface modification techniques designed to improve implant performance through increased biocompatibility, enhanced osseointegration, and the provision of antibacterial properties.

Significant developments have been made in surface modification, all aimed at improving osseointegration, enhancing implant performance, and increasing the longevity of the implant. Surface roughness, although promoting osteoblastic differentiation, reduces cellular proliferation simultaneously [30]. This trade-off in cellular responses may contribute to the difficulty in achieving complete and stable osseointegration. The design of an implant is important for successful

osseointegration and long-term stability. Implant shape, geometry and porosity should be taken into careful consideration, as they influence both mechanical and biological performance. Chemical treatments like fluoride treatment, ion implantation and acid etching are used for metal implants. Bioactive coating like hydroxyapatite (HA) and tricalcium phosphate are being used to improve bone integration [31]. Cell culture has been studied to understand the behaviour of how osteoblast cells respond to different implants by examining morphology, growth and differentiation. Surface characteristics play an important role in successful implant integration, while surface topography is essential for cellular response and improved osseointegration. The increased roughness in Ti-6Al-4V implants made from grit blasting showed higher bone-to-implant contact and improved bonding strength, accelerating collagen production and calcification when compared to a polished surface [32].

Surface energy is another factor influencing osseointegration. It refers to the excess energy present at the implant surface. A higher surface energy increases the wettability of the implant surface, which allows biological fluids, proteins and cells to spread more effectively across the implant [33]. Nanostructured surfaces can be engineered on Ti implants by nanoscale HA or titanium dioxide.

Fabrication techniques

The fabrication technique used for Ti and its alloys determines the material's microstructure, porosity, and morphology, and therefore has a major influence on the performance of Ti-based implants. Tailoring the fabrication techniques of Ti-based alloys it can enhance their mechanical properties and biocompatibility, improving implant function and biological outcomes.

Conventional

Conventional manufacturing techniques, including machining, welding, casting, powder metallurgy, forging, rolling, and extrusion, have been widely used to fabricate Ti-based alloys because of their desirable mechanical properties and biocompatibility. The choice of an optimal fabrication method significantly affects the microstructure, mechanical properties, corrosion resistance, and compatibility of Ti alloys. Therefore, various fabrication routes are important for tailoring the structural and functional properties of Ti alloys to meet biomedical application requirements.

Machining

Machining of Ti alloys refers to subtractive manufacturing, which mainly affects the surface rather than the bulk microstructure [34]. The low thermal conductivity, low elastic modulus, high chemical reactivity, and high-temperature resistance make Ti and its alloys challenging to machine [35]. Ezugwu and Wang [35], reported that a significant amount of heat generated at the workpiece interface leads to degradation of the cutting edge. In addition, Revuru *et al.* [36], noted that the low elastic modulus of Ti alloys promotes significant elastic recovery (spring back) during machining, which induces vibrations and chatter; these dynamic instabilities degrade surface finish and introduce surface irregularities. Machining can expose existing pores, inclusions, or casting defects near the surface, especially in cast or porous Ti parts, and once exposed, these defects can become sites for corrosion and lower implant life [37]. Therefore, in biomedical applications, machining with sharp tools, controlled cutting conditions, and proper cooling is important because it helps maintain surface integrity, corrosion resistance, and long-term performance [37]. Hence, to fully utilize the beneficial characteristics of Ti alloys for highly efficient and affordable biomedical implants, it is necessary to develop and implement improved machining strategies.

To address the high manufacturing cost and machining problems of Ti-based alloys, Esteban *et al.* [38], adopted two distinct solution strategies: the first is minimizing or eliminating machining

through near-net-shape processing routes (powder metallurgy, additive manufacturing, and advanced solid-state processes, *e.g.*, FAST-forge, FAST-DB), and the second is optimizing conventional machining parameters to maximise productivity and minimize waste. Studies show that machining involves not just shaping but also surface engineering, allowing for tailoring to modify the microstructure and thereby improving corrosion protection in Ti alloys.

Welding

Welding involves joining metal parts by heat to create a solid bond, and it is employed in biomedical fields to produce complex implants and tailored structures. However, welding Ti is challenging because Ti reacts readily with air at high temperatures. The weld area must be carefully protected from contamination to avoid brittleness and defects [39]. Primary welding techniques for Ti include tungsten inert gas (TIG/GTAW), metal inert gas/gas metal arc welding (MIG/GMAW), electron-beam welding, resistance spot and seam welding, plasma welding, friction welding, and, in some cases, brazing and explosive welding. TIG is among the most common welding methods because it offers good control and produces clean welds. Electron-beam welding is chosen when deep, precise welds are required [40].

Xu *et al.* [41], studied the impact of welding thermal input on the corrosion properties of TA10 Ti alloy pipe joint welds. The results showed that increasing heat input altered the weld microstructure by promoting phase transformations, residual stresses, and defects, thereby affecting passive film stability and corrosion resistance. Excessive heat input increased localized corrosion susceptibility due to microstructural heterogeneity and micro-galvanic effects between different phases. The study highlighted that optimizing welding parameters is essential to maintain the corrosion resistance of Ti alloy welded components. However, the thermal cycles involved can greatly impact the microstructure and corrosion resistance of the welded joints. Changes in phase distribution, grain morphology, residual stresses, and defect formation in the fusion zone and heat-affected zone can impact passive film stability and corrosion resistance. Thus, optimising welding parameters, especially heat input, is crucial to achieve a balance between mechanical performance and improved corrosion resistance for the long-term biomedical use of Ti alloys.

Casting

Casting is one of the oldest manufacturing methods, where molten metal or alloy is injected into a mold cavity as needed, followed by allowing it to set. The solidified component is then removed from the mold, yielding a near-net-shape product. Watanabe *et al.* [42] investigated the casting process of Cp-Ti and discovered that ultra-high-speed centrifugal casting enhances mold filling and decreases porosity relative to pressure-difference casting. This results in a denser microstructure and higher-quality castings. Pani *et al.* [43], studied new β -Ti-rich compositionally complex alloys (CCAs) in their as-cast form, examining how their multi-element composition and casting-related microstructure influence their mechanical and corrosion properties for biomedical use. Their research demonstrates that these Ti-Zr-Nb-Fe-Mo CCAs undergo notable Nb/Zr segregation following casting, yet they still attain strength and ductility that are comparable to or surpass those of other materials, such as Ti-6Al-4V, while displaying excellent corrosion resistance in biological environments.

Ti casting poses unique challenges because Ti has a melting point of about 1668 °C and a strong affinity for oxygen, nitrogen, and hydrogen at high temperatures. Therefore, casting is typically performed under vacuum or in inert gas atmospheres to prevent contamination. Cast Ti implants exhibit good corrosion resistance and acceptable mechanical properties; however, defects such as porosity, shrinkage cavities, inclusions, and coarse-grained structures may form during solidification. These

defects can adversely affect fatigue strength and implant reliability. Consequently, cast components often require subsequent thermomechanical processing and surface finishing to improve performance.

Forging

Forging is a thermomechanical processing technique in which Ti alloys are plastically deformed under compressive forces at elevated temperatures, typically within the $\alpha+\beta$ phase region. This process refines the grain structure, eliminates internal defects, and improves mechanical properties.

Zare *et al.* [44] investigated Cp-Ti subjected to multi-directional forging (MDF) with intermediate annealing at 650°C between forging cycles. Their work shows that this thermomechanical route produces an ultrafine-grained α microstructure with a large fraction of high-angle grain boundaries, resulting from grain fragmentation assisted by twinning and boundary conversion during MDF. The main limitation of hot forging lies in the necessity of expensive tooling and increased production costs. Moreover, creating intricate shapes solely by forging can be challenging, frequently requiring supplementary machining procedures.

Powder metallurgy and rolling

Powder metallurgy (PM) is a manufacturing process that involves compacting metallic powders and then sintering them at temperatures below their melting point. The process includes powder preparation, blending, compaction, sintering, and secondary finishing steps. PM offers advantages for biomedical Ti alloys, including high material utilisation, lower processing temperatures, reduced manufacturing costs, and the ability to fabricate components with controlled porosity. Interconnected pores are beneficial for orthopaedic implants, as they promote bone ingrowth and enhance osseointegration. Furthermore, PM facilitates the production of near-net-shape components, minimising machining requirements [45].

Rolling is a metal-forming process in which Ti alloys are rotated between two cylinders, reducing their thickness and changing microstructures. The process varies with temperature and can be categorised as either hot rolling or cold rolling. Hot rolling is typically used to shape cast material into sheets, plates, rods, and bars. It encourages grain refinement and enhances mechanical properties *via* controlled deformation. Cold rolling increases strength further through strain hardening and also improves surface quality and dimensional precision [46,47].

In biomedical Ti alloys, rolling helps create fine-grained microstructures that improve strength, fatigue resistance, and corrosion resistance. Rolled Ti sheets and rods are frequently used as raw materials for the subsequent fabrication of surgical instruments, dental implants, fixation plates and orthopaedic devices.

Extrusion

Extrusion is a metalworking process where metal is pressed, whether heated or not, through a die, thereby shaping the metal as it moves through the die cavity. The substance that results from the mould is referred to as the extrudate. The metal withstands compressive and shear stress to attain the desired die shape. Extrusion is generally classified into four types: direct, indirect, impact, and hydrostatic. During direct extrusion, a solid ram forces the entire cast billet through a fixed die. In this process, an extra amount of force is needed to overcome the frictional resistance between the sliding belt and the inner wall of the container (extrusion chamber), leading to high energy consumption. In the indirect extrusion process, a hollow ram carries the die and forces it into a stationary billet enclosed within the container. Because there is no relative motion between the billet and the chamber wall, friction at the billet-container interface is eliminated, resulting in a lower extrusion force than

with direct extrusion. In the impact extrusion process, the extruded component tends to adhere to the punch after forming; therefore, a stripper plate is used to remove the part efficiently from the punch. In hydrostatic extrusion, a high-pressure fluid transmits the extrusion force to the billet, forcing it through the die. Since the billet is surrounded by the pressurised fluid, friction between the billet and the chamber is eliminated. Furthermore, the fluid functions as a lubricant at the interface between the billet and die, lowering the extrusion force and enhancing material flow. A key challenge in finite element (FE) simulation of extrusion is mesh distortion, which can compromise the accuracy and stability of the analysis. [48]. Comparative analysis of conventional techniques with microstructure, corrosion resistance, tribocorrosion and service life is shown in Table 4.

Table 4. Comparative analysis of conventional fabrication techniques with microstructure, corrosion resistance, tribocorrosion and service life

Fabrication technique	Microstructure	Corrosion resistance	Influence on tribocorrosion	Implant life
Machining	Changes the near-surface layer; may create residual stress, surface damage, or a thin altered layer	Poor machining finish can damage the surface and reduce corrosion resistance	Rough or damaged machined surfaces can increase wear and corrosion	Better machining and polishing can improve long-term performance [37]
Welding	Creates a fusion zone and heat-affected zone and texture changes	Poor shielding or contamination can cause oxidation and reduce corrosion resistance at the weld	Welded zones can also show higher tribocorrosion if the structure or shielding is poor.	Good welding practice is important to avoid early failure and maintain implant durability [41,49]
Casting	Coarse grains segregation	Segregation and casting defects can promote corrosion	Moderate to weak unless carefully refined; cast surfaces and segregation can accelerate tribocorrosion damage	Shorter unless followed by processing or heat treatment, because defects and inhomogeneity can reduce fatigue and corrosion performance [43]
Powder metallurgy	Fine, equiaxed grains; homogeneous phase distribution; can be nanocrystalline after intensive milling and consolidation	Improved because of better homogeneity and refined phases; can be tuned strongly by milling and sintering parameters	Good, especially when wear-resistant phases and dense consolidation are achieved; still depends on pore control and oxide contamination	Usually favourable when porosity is low, and phases are controlled, since uniform microstructure delays localized attack & fatigue damage [45]
Forging	Refined and recrystallized grains; controlled texture; more uniform phase distribution than casting	Usually better than cast material because forging reduces segregation and can improve passive film stability through microstructural refinement	Strong option for load-bearing implants, especially when combined with proper heat treatment to optimize wear and corrosion	Typically good, since forging improves strength-ductility balance and fatigue resistance, which are important for long implant service [50]
Rolling	Re refined grains, elongated or recrystallized structure; hot rolling can also reduce defect sensitivity	Grain refinement and microstructural uniformity reduce corrosion	Good, but performance depends on texture control and surface finish; deep rolling and post-rolling heat treatment can further improve wear-related behaviour	Improves strength and fatigue performance, which supports longer implant service [46]
Extrusion	Refined, elongated α or $\alpha+\beta$ structure	Enhanced corrosion resistance	Improved repassivation and lower wear, but in oral/body fluids the passive film can still break under sliding.	Expected to improve if extrusion is combined with proper heat treatment and surface modification [48]

The selection of a traditional method for Ti-based alloys depends on the specific implant use and the desired material properties. Forging and extrusion are frequently used for load-bearing parts, as they enhance strength, fatigue resistance, and microstructure refinement. Powder metallurgy is ideal for applications needing controlled porosity to facilitate osseointegration. Although casting enables the production of complex shapes, its application is limited by challenges related to defects, segregation, and microstructural control. Therefore, the suitability of a fabrication technique should

be evaluated based on the functional necessities of the implant rather than considering a single technique. The limitations of conventional approaches have further motivated the development of advanced manufacturing strategies capable of achieving complex geometries and tailored properties for next-generation biomedical implants.

Additive manufacturing technique

Additive manufacturing (AM) technique, commonly known as 3D printing, is a fabrication technique that creates 3-D objects by successive material deposition layer by layer [51]. AM technique has gained growing acceptance because of its ability to simplify the manufacturing process [51]. Figure 5 shows the medical application of the AM technique in metal implants. AM can produce parts with highly intricate shapes while requiring minimal post-processing, utilizing tailored materials with near-zero waste and accommodating a broad spectrum of biomaterials, which include metals [52]. Based on the material deposition mechanism and energy source, additive manufacturing techniques used for biomedical applications are generally classified into powder bed fusion (PBF), directed energy deposition (DED), binder jetting, and material extrusion, which allow fabrication of Ti implants with personalized shape, graded porosity, complex geometry, and suitable mechanical properties. Examples include the Delta TT porous Ti acetabular cup, produced by selective electron beam melting (SEBM), and the CONDUIT spinal fusion device, a 3D-printed implant with an elastic modulus comparable to trabecular bone, which received FDA approval in 2017. Similarly, the NEST spinal fusion device incorporates a porous core and solid outer structure to promote bone ingrowth and improve biological fixation [53].

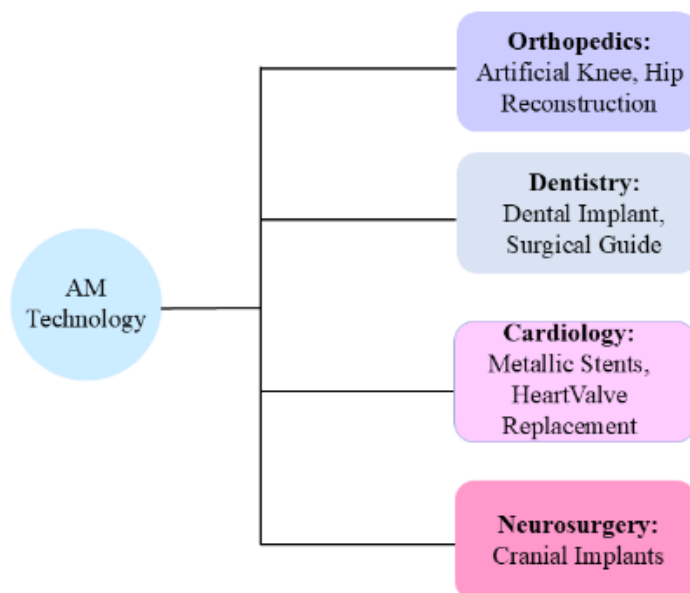


Figure 5. Schematic illustration of additive manufacturing's application as medical implants [52]

Powder bed fusion

Powder bed fusion (PBF) uses a high-power laser that completely melts the Ti powder; this technique can be categorized into complete melting and partial sintering of the powder [54]. This technique is used for high precision and densification, with excellent mechanical strength and complex geometries. The development of AM techniques has given rise to powder bed fusion, which has a wide application in the field of biomedical implants. Liu *et al.* [55] studied the mechanical properties of β -type Ti-24Nb-4Zr-8Sn alloy, which resulted in an enhanced tensile compression ratio, which is linked to increased porosity and enhanced superelastic characteristics.

Powder bed fusion (PBF) technique is categorized into two main processing approaches, *i.e.* selective laser melting (SLM) and electron beam melting (EBM); these two techniques not only have high forming efficiency and accuracy, but also have a wide range of applications. In SLM, it manufactures parts with high precision, complex geometries and mechanical properties similar to those achieved through conventional methods. TA15 (Ti-6.5Al-2Zr-1Mo-1V) Ti alloy was fabricated using SLM technology, which resulted in a uniform microstructure and improved mechanical properties through precise control of the melt pool and accurate layer deposition[56]. EBM is an advanced additive manufacturing technique, where the process is performed under a vacuum environment to fabricate on a powder bed using 3-D computational design methods, which produces parts with higher density when compared to SLM [56]. Applications including hardening, welding, drilling and melting are advantages of electron beam machining. DED has attracted increasing interest in biomedical applications, particularly for the fabrication of patient-specific Ti-alloy implants with complex geometries and porous structures that can facilitate osseointegration.

Directed energy deposition

Directed energy deposition (DED) is a metal AM technique for the fabrication of patient-specific implants with complex geometry and produces a porous structure that promotes osseointegration [57]. In the DED technique, a laser or electron beam is used to locally melt the supplied metal powder and merge it with the material that was already deposited [58]. DED enables the in-situ repair and cladding of three-dimensional components, as well as the fabrication of functionally graded materials and alloys through the real-time adjustment of powder composition and deposition parameters [59,60]. DED provides several important advantages, including the ability to process a wide range of materials and achieve high deposition rates. Nevertheless, maintaining stable and accurately controlled processing conditions remains essential for producing defect-free components. Milošev *et al.* [61] have conducted a comparative study of Ti-6Al-4V alloy using techniques such as wrought processing and DED laser bed technique, and reported the technique to be highly promising for complex shapes. Variations in process parameters can lead to defects such as porosity and residual stresses [62], both of which may compromise the mechanical performance and overall reliability of the final component [62]. Laser engineered net shaping (LENS) and wire arc AM (WAAM) are classified as DED. LENS uses higher laser power, enabling the modification, repair and addition of new features to existing parts. The LENS process produces a more defect-free interfacial bond between layers, whereas WAAM uses an electric arc as the melting source. Both techniques have high deposition rates, making them suitable for large parts and repair applications, while also promoting bone ingrowth and biological fixation. DED offers several advantages, such as high deposition rates, material versatility, the capability to produce complex geometries and efficient repair of expensive parts. Challenges like residual stresses, porosity and poor surface finish can hamper the quality and performance of fabricated components [63].

Binder jetting

While material jetting directly deposits and cures the build material itself as droplets (typically photopolymers or waxes), a closely related jetting technology takes a different approach: instead of jetting the structural material, it jets a liquid binder into a powder bed to selectively glue particles together. This process is known as binder jetting. Binder jetting is the combination of powder and a liquid binding agent, which is cost-effective and produces porous parts [64]. Binder jetting produces complex-shaped parts inside a powder bed through a repeating, layer-by-layer sequence: a thin layer of Ti powder is spread, and inkjet printing selectively deposits a liquid adhesive to bind the

particles locally. Once this printing process builds the component, the powder bed is cured to form green parts. Finally, these green parts undergo de-powdering and sterilization at a specific temperature to achieve the desired density [65]. The features of high efficiency and minimal cost make binder jetting widely adopted in biomedical applications and for metal implants. Based on the composition of the binder, it can be classified into organic and inorganic types; polyethylene is a non-toxic and commonly used binder for biomedical applications. Dashtgerd *et al.* [66] investigated NiTi alloy using binder jetting followed by sintering at different temperatures, in which he reported high cell viability, indicating favourable conditions for cell growth and survival. However, binder jetting offers cost-effectiveness and high efficiency without residual stress, though it has drawbacks such as shrinkage during sintering, compromised quality, and reduced mechanical properties. Therefore, the technology is optimal for applications prioritizing speed and design freedom over maximum material strength.

Material extrusion

Extrusion-based AM (EB-AM) is a 3D fabrication technique wherein a material is heated and extruded through a nozzle and deposited layer by layer. It employs two types of technologies: filament extrusion and pellet extrusion. Filament extrusion systems use pre-formed spooled filament, which limits the range of usable materials. Pellet extrusion printers are more complex, incorporating a screw inside a barrel that feeds, transports, melts, and forces the material through the nozzle at the required temperature. Despite their greater complexity, pellet extruders offer lower material costs, higher deposition rates, and access to a wider variety of materials. Material extrusion deposits a polymeric filament containing metal powders, as printed by a system like that used for thermoplastics. After printing, the green parts undergo rebinding to remove the polymeric binder, transforming them into brown parts. Subsequently, these parts are heated at a high temperature for metallurgical bonding and densification [67]. Lorenzi *et al.* [68] mentioned its principal disadvantage during the formation of distinctive macro-defects, which usually result in a periodic and interconnected network, potentially undermining the ultimate material properties. Later, he investigated a comparative study of Ti-6Al-4V alloys by using different AM techniques and material extrusion. Xu *et al.* [69] were inspired by the cortical-cancellous architecture of natural bone; hierarchically porous Ti6Al4V scaffolds were fabricated *via* material extrusion (MEX) with tuneable Young's modulus. A bio-inspired cortical-cancellous bone implant was produced using dual-syringe MEX, with an outer layer of Ti648Mn2 ink at 10 % structural porosity and an inner core of Ti646Mn4 ink at 70 % structural porosity, achieving simultaneous consistency in structural porosity and Young's modulus with natural bone. In conclusion, the material extrusion technique provides a cost-effective and geometrically flexible route to fabricate Ti alloys for biomedical applications, with clear advantages in equipment cost, feedstock handling, and contamination control compared with powder-bed and beam-based AM.

The selection of an additive manufacturing technique for biomedical application of Ti alloys depends primarily on the intended application, material requirements, geometry, mechanical loading, porosity, dimensional accuracy, production volume, and post-processing needs. Among the available techniques, laser PBF (LPBF) is generally preferred for high-resolution, patient-specific, dense, and porous implants, while EBM is more suitable for large, highly porous orthopedic structures. DED is advantageous for repairing or manufacturing large components, whereas material extrusion, material jetting, and binder jetting are better suited to polymeric, indirect, or cost-sensitive applications. Therefore, the AM technique can be selected according to the microstructure and current density of Ti alloys for biomedical applications, as shown in Table 5.

Table 5. Effect of additive manufacturing technique on the microstructure, corrosion behaviour, and biomedical applications of titanium alloys

AM technique	Source	Microstructure	Polarization resistance	Biomedical application
Powder bed fusion	High energy laser	Acicular α' martensite; textured fine columnar grains	Lower corrosion resistance	Orthopedic Implants; Acetabular cups, Femoral stems [54]
Directed energy deposition	High energy laser, electron beam, plasma / electric arc	Mixed α' and $\alpha + \beta$	Lower corrosion resistance	Spinal implants; Interbody cages, pedicle screw [70]
Binder jetting	No direct heat source	Equiaxed α grains with intergranular β	Higher corrosion resistance	Cranial plates, mandibular implants [64]
Material extrusion	Heated nozzle, thermal energy	Primary α with segregated β at grain boundaries	Higher corrosion resistance	Dental implants [67]

Recent approaches and future perspective

AM has considerably increased the design and manufacturing possibilities of Ti-based biomedical implants by enabling personalised and complex geometries. However, the advancement of implant performance cannot be achieved solely through improvements in manufacturing processes. Challenges such as the high elastic modulus of conventional Ti alloys, limited control over the strength-ductility-biocompatibility balance, surface degradation and tribocorrosion, and the need for adaptive and long-term implant monitoring continue to drive research toward integrated material, computational, manufacturing, and digital strategies. Consequently, recent research has increasingly shifted from optimizing the fabrication of Ti implants toward designing advanced materials, predicting and controlling their performance, introducing adaptive functionality, and managing implant behaviour throughout its lifecycle.

Emerging methods such as high-entropy titanium alloys, artificial intelligence, machine learning, 4D printing, smart implant technologies, and digital twinning are attracting increasing interest. These methods cover various stages in the Ti-implant development process, including material creation, manufacturing efficiency, functional integration, and predictive lifespan. Integrating them with AM opens new possibilities for enhancing Ti implants, offering improved mechanical compatibility, corrosion resistance, wear resistance, personalized functionality, and consistent long-term performance. Table 6 presents various smart approaches and their functions as implants.

Table 6. Various smart approaches and their function as an implant

Smart approaches	Example	Function as an implant
Embedded sensors	Electrochemical sensors	Monitors loading, micromotion, passive-film breakdown, and changes in the implant environment [71]
Smart implant	Sensor-integrated hip, knee, dental or spinal implant	Measures implant performance and detect implant loosening, abnormal loading, or damage [72,73]
4-D	Additively manufactured shape-changing structure	Changes geometry or functionality with time or external stimulation [74]
Digital twin	Patient-specific virtual implant model	Combines sensor data and simulations to predict degradation and remaining implant life [75]
Shape memory alloy	NiTi	Recovers its shape; responsive to external stimuli [76]
Artificial intelligence	Machine-learning model for corrosion or tribocorrosion prediction	Identifies degradation patterns and estimates corrosion rate, wear rate, or failure probability [77]

High-entropy titanium alloys

High-entropy titanium alloys (HEAs) were developed by applying the principle of mixing elements to increase the configurational entropy of the system. Such HEAs enhance solid solution alloy formation stabilization, distinguishing HEAs from conventional alloys. These materials are commonly referred to as multifunctional element alloys (MPEAs) or compositionally complex alloys (CCAs) because of their diverse elemental compositions [78]. Conventionally, alloy design has primarily centered on microstructures composed of single solid phases with the aim of achieving desired mechanical properties. However, recent research indicated that microstructures made up of two intermetallic phases can display a beneficial combination of structural attributes when their morphologies and distributions are precisely managed. Consequently, expanding the investigation of compositionally complex alloys (CCAs) to include multiphase microstructures represents an important direction for future research [79]. The high configurational entropy, lattice distortion and sluggish diffusion effects of HEAs contribute to their desirable mechanical properties and structural stability. With the multi-principal component design framework, various HEA systems have been developed, including transition metal systems (Fe, Co, Cr, Ni, Mn, *etc.*), Al-transition metal systems, and refractory metal systems (Ti, Zr, Nb, Ta, Hf, W, Mo, V, *etc.*). These diverse HEA systems provide a broad range of possibilities for developing novel structural and functional metallic materials.

Ibrahim *et al.* [80], compared the conventional Ti-6Al-4V with high-entropy alloys (HEAs), which have emerged as promising candidates for next-generation biocompatible metallic materials because of their combination of high strength and hardness, wear and corrosion resistance, temperature stability, and biocompatibility. The equiatomic TiNbTaZrMo HEA initially showed high strength but was limited by low compressive ductility ($\sim 9\%$) and high density ($>9\text{ g cm}^{-3}$). Removing Mo produced the quaternary TiNbTaZr HEA, improving compressive ductility to $\sim 48\%$; however, its tensile ductility remained limited to $\sim 1\%$. Further compositional modification resulted in Ti₄₀Zr₂₅Nb₂₅Ta₁₀, which achieved a tensile ductility of $\sim 18\%$, a yield strength of 875 MPa, and a density of 7.2 g cm^{-3} in the as-cast state. These properties make it a potential candidate for orthopedic implants such as bone screws and fixation plates, while meeting the mechanical requirements of medical-grade Ti-6Al-4V ($\sigma_{0.2} \geq 759\text{ MPa}$ and $\epsilon_f \geq 10\%$) [80].

Cheng *et al.* [81] investigated bio-corrosion and biocompatibility of high-entropy alloys, specifically TiNbZrTa and TiNbZrTaMo high-entropy alloys (HEAs), using Ti-35Nb-7Zr-5Ta as a comparison. Both HEAs have a BCC structure with lattice distortion due to their equal atomic composition. This resulted in better corrosion resistance for TiNbZrTa than for TiNbZrTaMo and showed better cell proliferation, whereas adding Mo reduced the cell proliferation rate. Wang *et al.* [82] have reported that Al-Cr-Fe-Co-Ni HEA alloys have shown great mechanical properties governed by variations in microstructural and chemical composition. In the Al_xCrFeCoNi HEA alloy system, increasing the Al content can transform the crystal structure from a single-phase face-centered cubic (FCC) structure to a dual-phase FCC-BCC structure and ultimately to BCC/B2 phases. The microstructural changes can also be made by varying the proportion of the transition metals. Yan *et al.* [83] reported that the addition of elements like Fe, Co, and Ni in particular amounts favors the stabilization of FCC, whereas adding chromium promotes the formation of BCC. The duplex FCC-BCC structure offers a good strength-ductility balance, making it a promising candidate for future structural applications. Lower Cr content in Al-Cr-Fe-Co-Ni alloys reduces Cr in the body-centered phase (BCC), leading to pitting corrosion [84], whereas adding Ti to Al-Cr-Fe-Co-Ni HEAs enhances corrosion resistance, but high concentrations can increase pitting corrosion [85]. Popescu *et al.* [86] reported that the TiZrNbTaFe HEA alloy system produced by the powder metallurgy route showed

greater microstructural uniformity than HEAs produced by melting and casting. This new alloy system showed better mechanical biocompatibility, which has applications in orthopaedics and dental implants. HEAs containing elements such as Fe, Co, and Ni can form stable borides, making boronization a suitable surface modification method. Boronization forms hard and protective metal borides on the surface, which reduce surface damage during wear. As a result, the boronized HEA showed lower wear volume loss under different wear environments [87]

The corrosion, wear, and their synergistic effects in FeCrNiCo, FeCrNiCoAl, and FeCrNiCoMo HEAs exposed to artificial seawater. The findings showed that wear significantly influenced the corrosion rate, with the increase in corrosion due to wear being about 100 times greater than the rise in wear rate caused by corrosion [88].

The studies indicate that the corrosion and tribological performance of HEAs are significantly impacted by their composition, microstructure, processing route, and surface modifications. Ti-based high entropy alloys display promising corrosion resistance and biocompatibility, with strategies like controlled phase formation and boronization capable of enhancing resistance to wear and surface degradation. Despite the synergistic effects of corrosion and wear, this remains a critical consideration, especially in physiological environments where mechanical and electrochemical degradation occur together. Therefore, evaluating corrosion, wear, tribocorrosion, and biocompatibility together is crucial for determining the suitability of HEAs for biomedical implants.

AI-assisted alloy design

Increased use of AI-based methods, with a particular focus on machine learning (ML), has emerged with the advent of new technologies in biomedical research. With data at its core, ML identifies patterns, learns from observations, and facilitates decision-making. In healthcare, these approaches can assist with disease diagnosis, treatment planning, surgical support, and monitoring of biomedical implants. For implant structural health monitoring, AI-based systems could improve post-surgical assessment, which currently relies mainly on routine clinical visits or patient-reported discomfort. Such systems may enable earlier identification of implant-related problems. The traditional method is slow, and trial-and-error approaches increase the risk of error, making development time-consuming. Computational materials modelling offers valuable insight into the structural characteristics, dynamic behaviour and physical properties of materials and how they interact with external factors like temperature, pressure, and electric and magnetic fields, which are essential for the advancement of intelligent materials and sensing technologies [89]. The datasets contain information on elemental characteristics, thermodynamic properties, and processing conditions that act as important inputs for predictive models. When density functional theory and calculation of phase diagrams (CALPHAD) are combined with machine learning, it can predict phase behaviour and assist in selecting compositions with desired properties [90]

A variety of machine learning (ML) techniques includes support vector machines, random forests, neural networks, and generative models that can capture complex non-linear relationships between properties, composition and processing parameters [91]. Traditional methods of material design often involve labour and face limitations in controlling internal factors. ML overcomes this limitation by forming relationships linking inputs and outputs without requiring existing knowledge of the underlying mechanism, making it highly useful for intricate material systems where physical principles may be incomplete or calculations intractable [92].

Phase equilibria and thermodynamic principles are necessary to understand the material behaviour and phase transformation during processing and service. Phase diagrams are

fundamental tools for predicting phase stability and designing new materials. Most of the engineering alloys are multicomponent; therefore, graphical representation and experimental determination become more difficult. Hence, to overcome these issues, computational thermodynamics are applied to predict phase relationships, phase stability and thermodynamic properties; this new approach can greatly enhance the efficiency of material design and enable the study of complex multicomponent alloy systems [93].

CALPHAD works on the principles of minimizing the total Gibbs free energy to find a material in its equilibrium state. It works in four simple steps: 1) Model - the energy of each phase is expressed through mathematical equations. 2) Fit - unknown parameters in these equations are fitted to experimental and quantum-mechanical data. 3) Combine two- and three-element data are combined into a database to predict complex, multi-element materials. 4) Minimize - the program will cycle through all possible sets of phases and choose the set that results in the lowest total energy. Many studies have been done for CALPHAD-guided alloy development, such as the Al-Ti-Nb alloy [94] revealing an equal distribution of an FCC Al-rich matrix and in situ-formed tri-aluminum-nitride intermetallic phases that act as strengthening reinforcement. This alloy exhibited a high melting temperature of 647 °C, indicating superior thermal stability. Another study focused on the $\text{Al}_{87.5}\text{Ti}_{6.25}\text{Ta}_{6.25}$ (at.%) alloy, which has a low density of 3.75 g cm³ and contains 50 vol.% in-situ formed trialuminide reinforcement, which contributed to superior mechanical and thermal properties.

Despite their widespread application, CALPHAD has some underlying drawbacks, as the calculation for quaternary components fully depends on the data provided for reliable prediction. If the important phases are missing or phase boundaries are inaccurate, then the prediction provided by CALPHAD may be misleading for multi-component alloys. Direct extrapolation from binary to quaternary systems is possible only when there are no ternary phases present, the alloy system should have a simple phase relationship, a miscibility gap must be absent and only a few intermetallic compounds should exist. The development of new alloys will likely depend on the synergistic blend of CALPHAD, first-principles calculations and machine learning, thus allowing the exploration of large compositional spaces through data-driven approaches and enabling the faster discovery of advanced materials with target properties [95].

4D printing

Four-dimensional (4-D) printing has evolved since the introduction of 3D printing technologies, smart materials and computational design tools. Unlike 3D printing, 4D printing technologies incorporate time as an additional dimension, which changes their shape, properties, or function in response to external stimuli such as temperature, pH, light and moisture. This unique characteristic of 4D printing technologies has led to their application in smart electronics and biomedical devices [96]. The design for 4D technologies requires three key factors: 1) choice of stimuli-responsive material, 2) nature of triggering stimulus, 3) geometric and topological design of 4D printing. 4D printing technology is divided into two: Shape memory alloys (revert to their original configuration in response to external stimuli) and shape-shifting alloys (show temporary change and revert to their original form when the stimulus is removed) [97].

Smart biomaterials are materials that can sense physiological changes in the body and adapt their behaviour in response. Smart biomaterials are advanced and have applications in the biomedical field, such as drug delivery, advanced wound care and cell regeneration. Smart hydrogels, shape memory alloys, self-healing biomaterials, and liquid crystal materials. Smart hydrogels have been

extensively studied for 4D printing because of their characteristic to expand or swell in the presence of water, causing a change in their shape [98]. Several reports state that the shape-memory effect, along with self-healing, is triggered by body temperature, making them suitable candidates for biomedical applications.

4D printing enhances conventional additive manufacturing through the inclusion of responsive materials that can undergo programmed variations in shape, properties, or functionality after fabrication [74]. Depending on the material system, these changes can be triggered by stimuli such as temperature, moisture, pH level, light exposure, magnetic fields, or electrical signals. Shape-memory alloys, shape-memory polymers, hydrogels, liquid-crystalline elastomers, and responsive composites have therefore been investigated for developing dynamically adaptable structures for biomedical applications. Among these materials, NiTi is particularly relevant to titanium-based biomedical applications because its shape-memory and superelastic behaviour can be combined with additive manufacturing to produce complex structures capable of thermally or mechanically induced shape recovery. Additive manufacturing also enables the fabrication of porous and lattice-based NiTi structures with geometries suitable for biomedical applications [99]. Consequently, 4D-printed structures have been investigated for potential use including orthopedic implants, tissue-engineering constructs, drug-delivery systems, stents, and other adaptive medical devices.

Smart approaches

Implants provide mechanical support in case of leg or hip injury, helping in tissue regeneration and overall well-being of the patient. Smart implants are next-generation biomedical devices that combine therapeutic benefits with integrated sensing and monitoring capabilities, enabling both treatment and real-time assessment of physiological conditions [72]. Studies on hip replacement implants report that the number of hip replacement surgeries, along with secondary surgeries, is increasing because implant loosening, caused by mechanical overload and wear particles, is a major issue. Conventional implants support mechanical strength, but they cannot detect forces, loads, or any change acting on the implant after the surgery. Therefore, complications are only detected after the patient experiences symptoms and implant damage has already occurred. Therefore, the need for smart implants integrated within the implants is crucial for real time-monitoring to enhance the functionality and lifetime of the implant [73]. The incorporation of embedded sensors in orthopaedics and dental implants has emerged as a vital feature of smart implant technology [100]. These miniature sensors are incorporated within the implants to enable real-time monitoring of both mechanical and biological parameters. In the real-time healing process, changes may occur in implant load and strain; therefore, real-time monitoring offers valuable information about bone regeneration and osseointegration. Successful healing may be demonstrated by the progressive increase in load sharing by the surrounding bone. Conversely, abnormal stress distributions and excessive micromotion of the implant suggest delayed healing, implant instability and potential loosening. Similarly, pH and temperature sensors can detect early sign of inflammation or infection before clinical symptoms appear.

Currently, the most established clinical application of smart implants is in joint arthroplasty. An example is the U.S. Food and Drug Administration (FDA) approved persona IQ smart knee implant, which incorporates embedded sensors, including accelerometers and gyroscopes, to continuously monitor implant kinematics and transmit patient mobility data for clinical assessment [101]. This technology demonstrates the growing integration of digital twins and smart implants in personalized orthopaedic care. Over the last three decades, various sensor-integrated systems

have been developed for total hip arthroplasty (THA) and total knee arthroplasty (TKA). In 2021, a significant milestone was achieved with the first FDA-cleared sensor-enabled knee, Persona IQ, being introduced into clinical practice [102]. This development demonstrated the potential of connected orthopedic implants to integrate with secure digital health platforms, facilitating communication between patients and clinicians. These smart prostheses are intended to complement conventional postoperative assessment by providing continuous *in vivo* information on the biomechanical and physiological condition of the implant, thereby supporting timely and data-informed clinical decision-making that can be achieved through periodic clinical evaluations alone. These smart prostheses improve implant functionality through continuous monitoring; however, smart approaches can also be achieved through materials that respond directly to changes in their surrounding conditions. Among these materials, shape-memory alloys (SMAs) present a promising approach due to their capability to revert to a predetermined shape after being exposed to the right external stimulus. Shape-memory behaviour therefore provides an important basis for developing adaptive orthopedic implants with responsive functionality [103]. Among SMAs, Ni-Ti are extensively used in biomedical applications because of its high strength, biocompatibility and tribological properties [76]. The Ni-Ti alloy has 55 wt.% of nickel and 45 wt.% titanium, is often called NITINOL. Due to their excellent flexibility, superelasticity, and biocompatibility, Ti-Ni stents are widely utilized in surgical treatments such as constricted arteries, recurrent urethral obstructions, biliary obstructions and malignant esophageal stenosis. Shape-memory alloy possesses superior corrosion resistance compared with commercially pure titanium and Ti-6Al-4V, along with excellent Biocompatibility, making it a promising material for biomedical use [104]. However, the high Ni content (30 to 60 %) drives inflammation and immune responses; therefore, researchers are developing Ni-free Ti alloys by adding safer elements like Nb, Ta and Zr [105]. Among these, β -Ti alloys, like Ti-Nb-based shape memory alloys, have become a promising alternative for biomedical applications by providing shape memory and superelastic properties without the use of nickel. In addition, these alloys provide exceptional resistance to corrosion, outstanding biocompatibility, significant strength, and minimal elastic flexibility, making them attractive candidates for next-generation orthopaedic and dental implants. The future of smart implants lies in artificial intelligence (AI), wireless communication and sensing technologies, which are capable of real-time diagnosis. The integration of machine learning and AI could help detect abnormalities and infection in real time.

Digital twin technology

Digital twinning is an emerging technology that creates the virtual replica of any physical entity, system or process by exchanging real-time data between the digital and physical counterparts. It comprises of three essential components: the physical twin, the virtual twin and the twin data. The physical twin is the actual physical asset, such as a titanium implant, whereas the virtual twin is the digital model, which may be physics-based and data-driven and serves as the basis for simulation. The digital twin data generated through simulation and acquired from sensors are used to support decision-making [75]. Digital twin technology has been increasingly adopted in biomedical engineering, particularly in orthopaedic applications, where patient-specific three-dimensional models are generated from clinical imaging data such as computed tomography (CT) scans and magnetic resonance imaging (MRI). The individualized approach enables clinicians to evaluate different therapeutic strategies before intervention, thereby supporting more informed clinical decision-making.

A corrosion prediction system plays a crucial role in anticipating material degradation and supporting the development of effective corrosion protection strategies. Karkadakattil *et al.* [106] studied digital twin technology governed by physical laws by linking friction and corrosion to a single, irreversibly evolving surface aging state, keeping prediction realistic to the real world, which is represented in Figure 6. Conventional methods rely on empirical data and experimental tests, which may require high time and resource investment. Cai *et al.* [107] reviewed various atmospheric corrosion prediction methods, modelling of corrosive environments, calibration of corrosion effects, and establishment of corrosion kinetics. Models of short-term processes use power functions and there is a need for more accurate long-term predictions capable of reliably capturing long-term corrosion behaviour under varying environmental conditions [106]. However, digital twinning technology addresses the need for faster and more efficient corrosion prediction by integrating the data. This enables early identification of corrosion-prone areas, supports informed design decisions, reduces reliance on physical prototypes and facilitates cost-effective corrosion protection throughout the product development process.

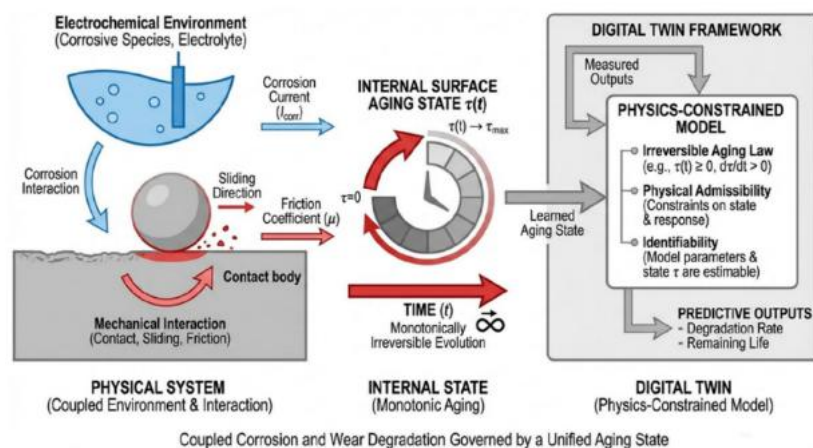


Figure 6. Conceptual representation of a physics-governed digital twin. Reprinted from ref. [106], under the Creative Commons by-nc-nd 4.0 license. No changes were made

Beyond corrosion prediction, digital twinning can also be used to study and monitor the mechanical properties during the implant service life. Current digital twinning implementation in orthopaedic trauma surgery relies on 3D models of patient anatomy obtained from clinical imaging to support patient-specific design and research application [106]. Digital twinning technology also addresses stress shielding, which is a common issue caused by the high stiffness mismatch between bone and the implant material. Therefore, digital twinning helps design and evaluate porous lattice structures to lower the elastic modulus of the implant, thereby mitigating stress shielding and bone resorption. Digital twinning also supports the assessment of fatigue life and durability by conducting virtual fatigue tests under physiological loading conditions. The simulation predicts the long-term cyclic performance of the implant and identifies potential micro-fractures and failure of the implant. Denkena *et al.* [108] reported digital implant lifecycle management (DILM), which extends conventional product lifecycle management by integrating the implant's full lifecycle, from development and manufacturing to implantation, monitoring, and decommissioning. It connects patient-specific information, manufacturing data, and clinical monitoring throughout the implant lifecycle, providing a framework for the application of digital twin technologies to monitor and optimize implant performance, as shown in Figure 7.

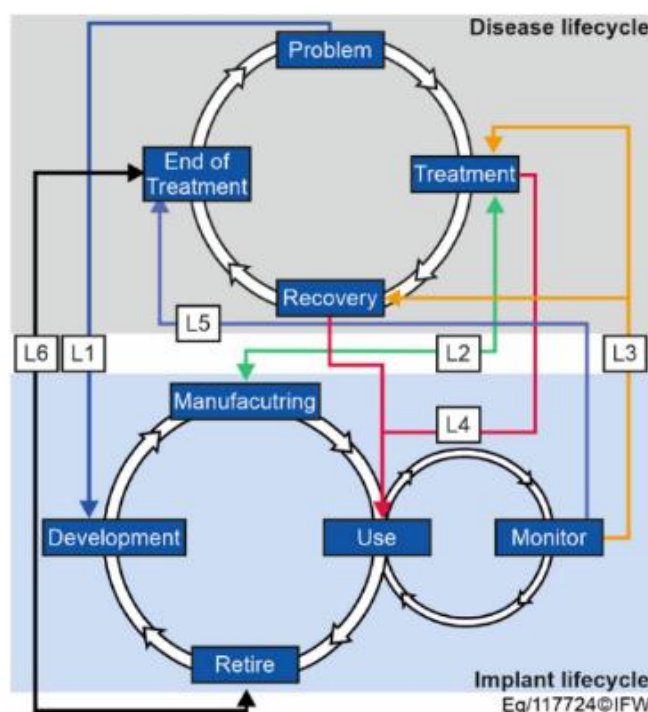


Figure 7. Flow diagram of the link between the disease and the implant lifecycle. Reprinted from ref [108], under the Creative Commons by-nc-nd 4.0 license. No changes were made

In addition to this, digital twinning enhances the prediction of implant-bone interface behaviour by simulating the osseointegration process through analysing micromovements, shear stress and micro-strain to predict the risk of implant loosening and the long-term stability[109]. Therefore, digital twinning technology offers valuable insight for optimizing surgical decisions and for the improvement of implant life. The application of digital twin technology in orthopedics has expanded across arthroplasty, spinal surgery, trauma, sports medicine, and rehabilitation, with different twin categories serving distinct clinical functions. Glinkowski *et al.* [110] stated that in arthroplasty, simulation-based digital twins primarily integrate CT/MRI data to support patient-specific biomechanical assessment, implant alignment, and preoperative planning. Simulation twins have also been investigated in spinal and trauma contexts for assessing load distribution, corrective measures, fixation techniques, and stress scenarios. Monitoring-oriented twins enhance these capabilities through the use of wearable sensors, motion capture, surface scanning, and repetitive measurements, allowing for longitudinal evaluation and rehabilitation monitoring. However, most orthopaedic digital-twin applications remain focused on anatomical reconstruction, biomechanical simulation, surgical planning, clinical decision-making, and functional monitoring, whereas material degradation mechanisms of metallic implants have received comparatively limited attention. In particular, the integration of electrochemical corrosion, mechanical wear, tribocorrosion, fatigue damage, and biological responses into a patient-specific digital twin could provide a more comprehensive representation of implant degradation throughout its service life.

Conclusion

Conventional manufacturing techniques, including casting, powder metallurgy, forging, rolling and heat treatment, are key to the advancement of titanium-based implants as they offer predictable mechanical properties and scalability to industry. However, the methods have several limitations in terms of their capability to produce complex geometries, patient-specific designs and controlled porous structures for optimal osseointegration. With the advent of recent developments

in AM technologies, EBM, SLM and DED, customized implants with complex architectures and tailored mechanical properties have been revolutionized. These technologies enable the creation of porous constructs that closely resemble natural bone, providing improved bone ingrowth and implant fixation. Corrosion performance of Ti alloys is still a challenging topic of research because implants are constantly subjected to various physiological environments and mechanical loading conditions. The composition of the alloy, manufacturing process, surface condition and environmental factors all have significant effects on the overall stability of the passive oxide layer and long-term performance of the implants. Significant improvements in corrosion resistance, osseointegration and overall implant functionality have been achieved. New technologies like AI for alloy design, CALPHAD for material optimization, high-entropy alloys (HEAs), bio-functional surfaces, 4D printing, and smart implants are new approaches that are expected to bring new possibilities for the development of future implants. These innovations provide opportunities for creating implants that exhibit enhanced mechanical properties, biocompatibility, real-time monitoring and improved patient-specific treatment outcomes. A combination of advanced alloy design and innovative manufacturing techniques is anticipated to be the upcoming research on titanium implants. Interdisciplinary efforts across materials science, biomedical engineering, and clinical outcomes are essential for creating safer, more durable, and personalized titanium-based solutions.

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