

Biomedical Titanium Alloy for Implant Application: Processing; Biofunction and Failure Mechanisms - Critical Review

Marwa J. Mohammed

College of Biomedical Engineering, University of Technology, IRAQ

Corresponding Author: marwa.j.mohammed@uotechnology.edu.iq



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ABSTRACT

Titanium and its alloys are widely used in orthopedic, dental, spinal and craniofacial devices because they offer a useful combination of high specific strength, corrosion resistance, fatigue capability and a stable oxide surface. Their long-term behavior, however, depends on much more than alloy composition alone. Manufacturing route, microstructure, defect population, architecture and surface condition can all influence elastic compatibility, fatigue, tribocorrosion, osseointegration and infection-related risk. This structured critical review brings together evidence from alloy design, additive manufacturing, surface engineering and implant-failure studies within a processing–microstructure–biofunction–failure framework. Rather than following a formal systematic-review protocol, the review uses a structured narrative search and gives priority to primary quantitative studies and current implant-material standards. Selected β -type alloys have shown elastic moduli of approximately 60–74 GPa, compared with values around 120 GPa for conventional Ti–6Al–4V, while retaining substantial strength. Studies of additively manufactured Ti–6Al–4V further show that defects, post-processing and physiological media can change the mechanism that limits fatigue performance. Overall, the literature does not support simple rules such as lower modulus, higher porosity, greater roughness or stronger antibacterial activity as universally desirable targets, because each can create competing effects on fatigue, corrosion, local stress concentration or host-cell response. A more useful approach is to evaluate alloy chemistry, architecture, defects, surface modification, post-processing, cleaning and sterilization together in relation to the mechanical and biological demands of the intended implant.

Keywords- titanium alloys; biomedical engineering; orthopedic implants; dental implants; additive manufacturing; β -type titanium; porous implants; tribocorrosion; fatigue; surface engineering; implant standards.

I. INTRODUCTION

Titanium and its alloys are central to biomedical engineering because they combine relatively low density with high specific strength, corrosion resistance and a naturally formed oxide layer that supports their use in contact with tissue and body fluids [1,9,11,12]. Commercially pure titanium and Ti–6Al–4V remain the most established choices, while β -type and metastable- β alloys containing Mo, Nb, Ta, Zr and Sn are being explored to reduce elastic modulus and improve implant-relevant combinations of strength, corrosion resistance and biocompatibility [2,9,10,16–18]. The motivation is straightforward: dense metallic implants are much stiffer than bone, so both lower-modulus alloys and porous architectures are being investigated as ways to reduce stiffness mismatch and stress shielding. Yet modulus cannot be considered in isolation from fatigue strength, phase stability, corrosion behavior or biological compatibility [9,10,16–18]. Primary studies illustrate this balance. Gordin et al. reported an elastic modulus of about 74 GPa for β -type Ti–12Mo–5Ta compared with about 120 GPa for Ti–6Al–4V, while Wei et al. reported approximately 60 GPa modulus and about 1250 MPa strength for Ti–30Ta–10Nb [16,17]. These results show that lower stiffness can be achieved without removing the strength needed for load-bearing applications.

Manufacturing adds another layer to this problem. Wrought processing, casting, powder metallurgy and machining remain important, while laser- and electron-beam additive manufacturing have made patient-specific geometries and controlled porous structures increasingly practical [4,5,8,13]. These features are attractive for bone-contacting devices because internal architecture can be used to tune apparent stiffness and create interconnected space for tissue ingrowth. At the same time, layer-wise processing may introduce anisotropy, residual stress, rough surfaces, partially fused particles and internal defects that can become dominant fatigue-initiation sites [4,5,7,13,20,21]. In vivo evidence illustrates both the promise and the limitations of porosity. Bandyopadhyay et al. reported improved early tissue integration in laser-processed porous titanium with 25 vol% porosity in a rat femoral model, and a later comparison found that additively manufactured porous Ti–6Al–4V with surface nanotubes approached the early-stage osseointegration response of porous tantalum at five weeks [19,22]. Such findings support the use of engineered porosity, but they also show why implant performance cannot be judged from a single material property or a polished laboratory coupon. This review therefore follows the full chain from alloy design and manufacturing to microstructure, architecture, surface condition, mechanical durability, biological function and failure.

II. METHODOLOGY AND SCOPE

This article was prepared as a structured critical narrative review rather than a PRISMA systematic review or meta-analysis. That choice reflects the breadth and heterogeneity of the topic: the relevant literature spans alloy development, additive manufacturing, fatigue, tribocorrosion, surface biology, sterilization and implant standards, with substantial differences in specimen geometry, loading conditions, electrochemical environment, surface treatment and biological model. Targeted searches were last updated on 23 August 2026 using PubMed and publisher platforms, including ScienceDirect/Elsevier, SpringerLink, ACS Publications and Frontiers. Backward reference chaining from directly relevant reviews was used to identify additional primary studies, and current implant-material standards were checked through the International Organization for Standardization (ISO). Search combinations included “titanium biomedical alloy”, “ β titanium elastic modulus”, “Ti–6Al–4V additive manufacturing fatigue”, “porous titanium osseointegration”, “titanium tribocorrosion”, “titanium implant surface roughness”, “antibacterial titanium coating”, “sterilization titanium surface”, “titanium implant life cycle assessment” and “surgical implant titanium standard”. Bibliographic identifiers and DOI links were cross-checked against publisher, PubMed or journal records where available.

Studies were included when they directly addressed titanium or titanium-alloy compositions for biomedical use, implant-relevant manufacturing or post-processing, mechanical durability, corrosion or tribocorrosion, surface–cell or surface–bacteria interactions, in vivo osseointegration, environmental assessment of implant manufacture, or recognized material standards. Reviews were used mainly to define the state of each subfield and identify gaps, whereas primary experimental studies were preferred for quantitative comparisons and for claims about the magnitude or direction of an effect [9–15,16–27,32–35]. Study selection was relevance-based and iterative rather than exhaustive. Papers were prioritized when they informed one or more links in the proposed processing–microstructure–biofunction–failure framework, and additional studies were identified by following references from highly relevant reviews and primary articles. Studies focused on non-titanium systems without a direct titanium comparator, aerospace-only applications without transferable biomedical evidence, or claims that could not be traced to a peer-reviewed or standards source were excluded. Because the search was not exhaustive, the review does not claim to represent the total number of eligible studies in the field. No formal risk-of-bias scoring tool was applied; instead, differences in study quality, model relevance and methodological limitations were considered narratively. The purpose was to test whether the proposed framework is supported across representative, high-relevance evidence and to identify areas in which methods and reporting remain inconsistent.

III. POSITIONING OF THE REVIEW AND EVIDENCE GAP

The biomedical titanium literature already includes strong reviews of additive manufacturing, β -type alloys, surface modification, porous scaffolds and tribocorrosion. The contribution of the present paper is therefore not to claim that these topics have been overlooked, but to examine how they interact within a single implant-performance pathway. Bringing them together makes it possible to compare trade-offs that are often missed when alloy chemistry, architecture, surface biology and failure are discussed separately. Table 1 positions the present review against representative prior syntheses and identifies the boundary of each source in relation to the integrated framework used here.

Table 1. Representative literature used to define the review boundary and novelty.

Source	Primary focus	Strength	Boundary relative to this review	Role
Nguyen et al. [4]	AM Ti–6Al–4V	Microstructure and mechanical-property synthesis	AM-focused; limited integration of surface	AM evidence base

Source	Primary focus	Strength	Boundary relative to this review	Role
			biology, tribocorrosion and sterilization	
Calazans Neto et al. [9]	β-Ti dental alloys	Low-modulus alloy design, corrosion and biological considerations	Dental/β-Ti centered; not a full processing-to-failure framework	Alloy design and stress shielding
Raja et al. [10]	Metastable β-Ti	Composition and thermomechanical processing	Does not integrate implant-surface biofunction and failure modes	Composition–processing mechanisms
Xue et al. [11]	Titanium surface modification	Broad physical and chemical surface strategies	Surface-centered; limited integration with bulk defects and fatigue	Surface engineering
Lu et al. [12]	Multifunctional coatings	Osteogenic and antibacterial coatings	Coating-centered; long-term mechanical durability remains a separate issue	Biofunction and infection control
Toosi et al. [13]	AM porous scaffolds	Architecture, permeability, mechanics and bone regeneration	Scaffold-centered; limited integration with alloy chemistry and tribocorrosion	Porous implant design
Dotta et al. [14]	Tribocorrosion	Systematic synthesis of surface-treated titanium	Failure mechanism focused	Wear–corrosion evidence
Park et al. [15]	Cleaning/sterilization	Direct evidence that final processing alters surface and cell response	Single experimental domain rather than a broad review	Final surface-state control

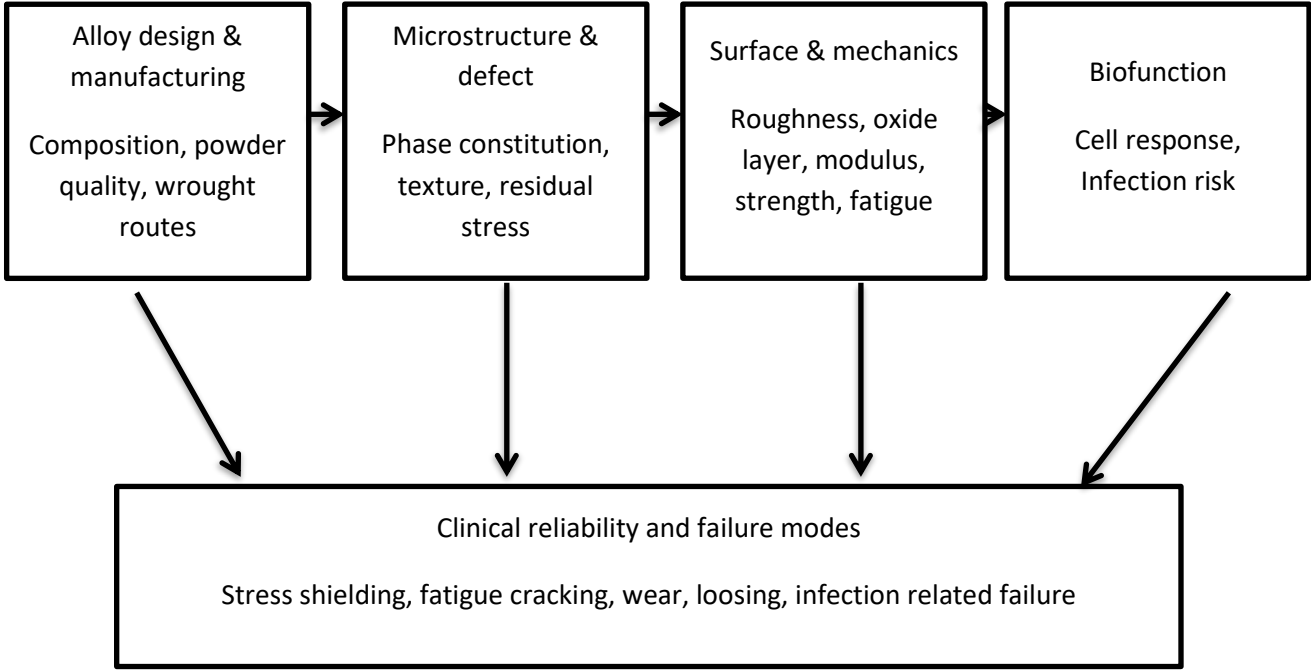


Figure 1. Author-developed conceptual framework for biomedical titanium-alloy qualification, based on relationships synthesized from Refs. [9–14,18–27].

IV. BIOMEDICAL DESIGN REQUIREMENTS, ALLOY SELECTION, AND QUANTITATIVE EVIDENCE

In biomedical titanium, mechanical, electrochemical and biological performances are closely linked. The TiO₂-rich passive film is central to corrosion resistance, yet wear, fretting, crevice conditions and local chemical changes can repeatedly disrupt this layer and alter the balance between material removal and repassivation [1,14,24,25]. In an oral environment, Licausi et al. compared cast and powder-metallurgy Ti-6Al-4V in artificial saliva containing 100 and 1000 ppm fluoride and in a mouthwash containing 112 ppm fluoride. Their results showed that tribocorrosion increased beyond a critical fluoride condition, highlighting why static immersion tests cannot fully represent sliding or fretting contacts [24]. A similar point emerges from modular orthopedic interfaces. Royhman et al. used a 200 N normal load and $\pm 50 \mu\text{m}$ fretting amplitude in bovine serum and observed material-couple-dependent electrochemical behavior [25]. Together, these studies show that tribocorrosion should be treated as part of implant design rather than as a secondary surface phenomenon. Elastic compatibility is similarly multidimensional. β -type titanium alloys can reduce stiffness because the β phase can be stabilized with elements such as Mo, Nb, Ta and Zr, but a clinically useful alloy must still provide sufficient strength, fatigue resistance and chemical stability [9,10,16–18]. Gordin et al. reported a Young's modulus of about 74 GPa for Ti-12Mo-5Ta, while Wei et al. reported approximately 60 GPa modulus and 1250 MPa strength for Ti-30Ta-10Nb [16,17]. Luo et al. extended this idea from alloy chemistry to architecture by additively manufacturing Ti-Nb-Ta-Zr lattices and directly linking porosity to modulus, yield strength and fatigue stress limit [18]. These studies show why stiffness matching cannot be reduced to a single number. Lower modulus may improve load transfer to bone, but excessive porosity or severe manufacturing defects can reduce fatigue resistance, and the acceptable balance depends on the implant geometry and load path [18,20,21,27].

Table 2. Selected primary mechanical studies supporting the quantitative synthesis.

Study	Material/process	Mechanical variable	Quantitative result	Critical implication	Ref.
Gordin et al., 2004	β Ti-12Mo-5Ta	Elastic modulus/hardness	~74 GPa; ~303 HV; Ti-6Al-4V comparator ~120 GPa	Lower stiffness achieved; final implant performance remains surface- and biology-dependent.	[16]
Wei et al., 2018	Ti-Ta-Nb	Composition / annealing	Ti-30Ta-10Nb: ~60 GPa modulus, ~1250 MPa strength, ~3.1 GPa hardness	Shows modulus-strength co-optimization rather than modulus minimization.	[17]
Hedayati et al., 2017	SLM Ti-6Al-4V porous CT specimens	Porosity/fatigue-crack propagation	66–84% porosity; crack path evaluated under cyclic tension	High porosity lowers stiffness but creates architecture-dependent crack paths.	[21]
Hogg et al., 2026	LPBF Ti-6Al-4V: as-built, HIPed and wrought	Corrosion-fatigue	As-built relative endurance ~half wrought/HIP; DMEM reduced endurance ~9% in wrought and ~6% in HIPed material	Defects dominated as-built fatigue; medium effects became clearer after HIP.	[27]

Table 3. Selected primary biological and tribological studies illustrating trade-offs among porosity, roughness, ion release, host response and antibacterial performance.

Study	Surface/implant condition	Biological/tribological variable	Quantitative model result or	Critical implication	Ref.	Evidence level
Bandyopadhyay et al., 2017	LENS porous Ti	Porosity / in vivo bone response	25 vol% porosity; rat distal femur, 4 and 10 weeks	Early integration improved in this rat model; effect remains architecture- and model-dependent.	[19]	Animal in vivo

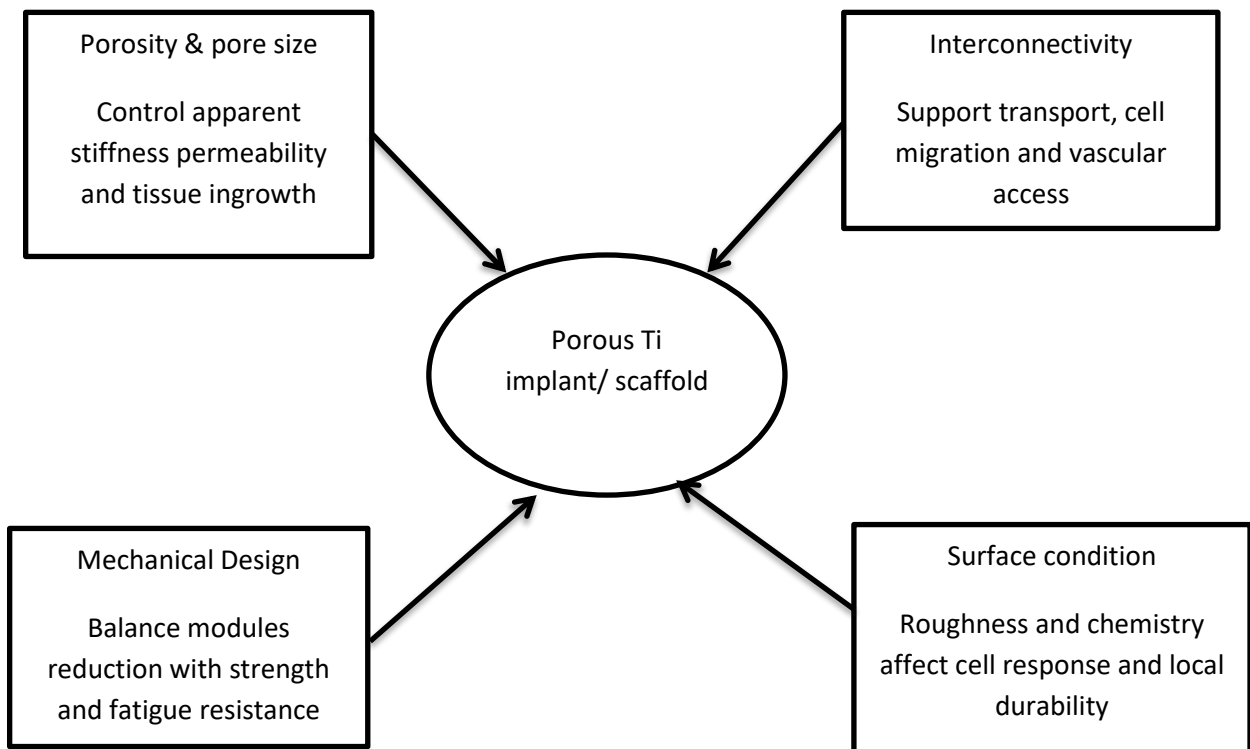
Study	Surface/implant condition	Biological/tribological variable	Quantitative model result	Critical implication	Ref.	Evidence level
Bandyopadhyay et al., 2019	AM porous Ti-6Al-4V vs porous Ta	Porosity / nanotube modification	~30% designed porosity; 5- and 12-week rat endpoints	Surface-modified porous Ti approached porous Ta in early osseointegration.	[22]	Animal in vivo
Schwartz et al., 2008	Ti-6Al-4V pedicle screws	Surface roughness / fixation	Ra 0.2 μm (smooth) vs 2.0, 3.0 and 3.3 μm (grit-blasted disks); 12-week sheep study; removal torque $5.3 \pm 0.4 \text{ Nm}$ (rough) vs $2.3 \pm 0.3 \text{ Nm}$ (smooth); bone-implant contact 73.5% vs 59.6%	Roughness increased in-vivo fixation despite lower in-vitro cell number.	[26]	In vitro + animal
Licausi et al., 2013	Cast and PM Ti-6Al-4V	Fluoride / tribocorrosion	100 and 1000 ppm fluoride; mouthwash 112 ppm	Fluoride altered tribocorrosion; static corrosion tests were insufficient.	[24]	In vitro tribocorrosion
Kontakis et al., 2021	PVD Ti-Ag coatings on Ti-6Al-4V	Ag dose / bacteria / osteoblasts	Ag surface content 27.7 vs 68.5 wt%; 28-day Ag release 4 vs 26.9 ppm; osteoblast alkaline-phosphatase response suppressed by 73% vs 96%	Higher Ag content increased antibacterial effect but produced a larger osteoblast-response penalty.	[32]	In vitro
Zhuang et al., 2021	Ti-6Al-4V-Cu alloy	Cu release / MRSA infection	Cu^{2+} release 0.106 $\text{mg}/\text{cm}^2/\text{day}$; 6-week rat MRSA model; radiological score 1.3 ± 0.4 vs 9.6 ± 1.3 and histological score 2.3 ± 0.5 vs 10.1 ± 1.9 (Ti-6Al-4V-Cu vs Ti-6Al-4V)	Cu-containing alloy reduced infection burden while retaining favorable biocompatibility.	[34]	In vitro + animal infection model

V. MANUFACTURING, ADDITIVE PROCESSING, POROUS ARCHITECTURE, AND FATIGUE

Additive manufacturing is attractive in biomedical engineering because it combines geometric customization with internal architectures that can be designed around both mechanical and biological requirements [4,5,8,13]. Even so, “3D-printed titanium” should not be treated as a single material category. Build orientation, scan strategy, heat input, powder history, post-processing and surface finishing can all change microstructure, porosity, residual stress and roughness, while lattice topology and relative density introduce an additional structural scale [4,5,13,18,20,21]. As a result, two implants made from the same nominal Ti-6Al-4V powder may show very different fatigue behavior if one contains lack-of-fusion defects or rough struts and the other has undergone HIP, chemical etching or machining [20,27]. Van Hooreweder et al. showed that stress relief, HIP and chemical etching after selective laser melting could improve fatigue life by reducing microstructural and surface stress concentrators [20]. Hedayati et al. further showed that fatigue cracks follow architecture-dependent paths in porous AM structures [21]. These findings support evaluating lattice implants as structures in their own right rather than extrapolating directly from bulk coupons.

Recent evidence also helps explain how manufacturing defects and the biological environment interact. Hogg et al. compared LPBF Ti-6Al-4V in as-processed and HIPed conditions with wrought material during high-cycle rotating-bending fatigue in DMEM. The as-processed material had roughly half the relative endurance limit of the wrought and HIPed conditions, with internal AM defects dominating crack initiation. Once defect severity was reduced, the influence of the

physiological medium became easier to detect [27]. This distinction is important. A corrosive medium may appear to have little effect when large manufacturing defects already control failure, yet environmental degradation can become more relevant after those defects are reduced. In this sense, post-processing does more than improve fatigue performance; it can change the mechanism that ultimately limits service life.



VI. SURFACE ENGINEERING, OSSEOINTEGRATION, AND INFECTION CONTROL

The titanium surface is the first part of an implant encountered by proteins, cells, body fluids and microorganisms, so even when the bulk alloy is unchanged, surface engineering can substantially alter biological performance [11,12]. Blasting, etching, plasma treatments, micro-arc oxidation, sol-gel processing and deposited coatings can modify topography, oxide chemistry, wettability and biological signaling [11,12]. The literature, however, does not support a simple rule that rougher surfaces are always better. Schwartz et al. compared smooth Ti-6Al-4V surfaces with Ra around 0.2 μm against grit-blasted surfaces with Ra values of 2.0–3.3 μm . After 12 weeks in sheep, rough screws showed greater bone-implant contact and more than twice the removal torque, even though cell number was lower on the rougher surfaces and markers of osteoblastic differentiation changed with surface condition [26]. This apparent contradiction is biologically meaningful: reduced proliferation may coincide with greater differentiation and stronger fixation, so cell number alone is an incomplete measure of osseointegration.

Antibacterial modification introduces another trade-off because the concentrations needed to suppress bacteria can also influence mammalian cells. The available primary studies show that this balance depends strongly on formulation and dose. Piñera-Avellaneda et al. evaluated a silver-doped calcium-titanate surface and reported inhibition of biofilm formation without a detrimental effect on osteoblast-like cell performance, including in vivo evaluation on porous implants [23]. Kontakis et al., however, found a clearer dose-dependent compromise. Two PVD silver coatings on Ti-6Al-4V released 4 and 26.9 ppm Ag by day 28; the higher-Ag surface reduced bacterial colonization more strongly, but osteoblast alkaline-phosphatase response was suppressed by 96% compared with 73% for the lower-Ag coating [32]. Copper-based approaches provide a similar lesson. A Cu-Co co-doped microarc-oxidized titanium surface supported MG63 osteoblast adhesion, proliferation and differentiation while inhibiting *Staphylococcus aureus* in vitro [33], whereas a Ti-6Al-4V-Cu alloy produced sustained Cu^{2+} release and reduced MRSA infection in a six-week rat model while maintaining favorable biocompatibility [34]. These results do not establish Ag or Cu as universally preferable antibacterial strategies. Instead, they show why efficacy, host-cell compatibility, ion release, coating adhesion and durability need to be assessed together.

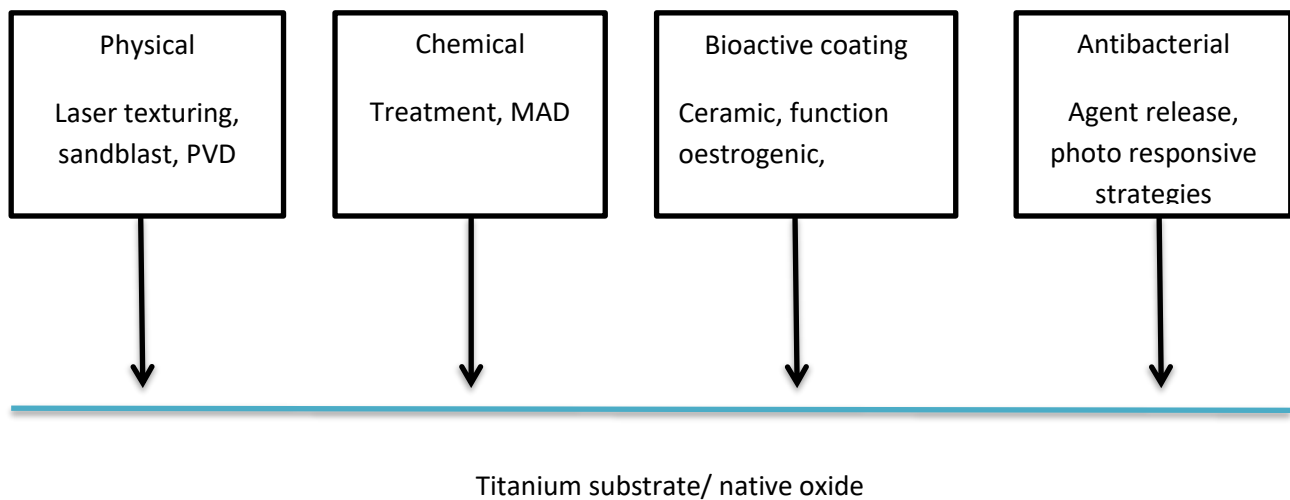


Figure 3. Author-redrawn map of biomedical titanium surface-modification strategies and their competing functional targets, synthesized from Refs. [11,12,23,26].

VII. FAILURE MECHANISMS: FATIGUE, TRIBOCORROSION, LOOSENING, AND STERILIZATION-RELATED SURFACE CHANGES

Long-term implant failure rarely arises from a single material property. Fatigue, wear, corrosion, biological fixation and infection can interact throughout service. Cyclic loading may initiate cracks at pores, lack-of-fusion defects, rough surfaces or geometric stress concentrators, while fretting and sliding can repeatedly disrupt the passive oxide layer and expose fresh metal to the surrounding electrolyte [7,14,20,21,24,25,27]. Dental-like environments add chemical variability, including the influence of fluoride on tribocorrosion [24], whereas modular orthopedic junctions are affected by material pairing, micromotion and contact conditions [25]. These findings help explain why a material may perform well in a static corrosion test but degrade under loaded contact. For devices exposed to both mechanical and electrochemical damage, combined corrosion-fatigue or tribocorrosion testing is therefore more informative than either test alone [14,24,25,27]. Biological fixation and the final processing steps before implantation add another source of variability. Poor or unstable osseointegration can allow micromotion, while bacterial adhesion and biofilm formation can compromise the peri-implant environment [12,23]. Cleaning and sterilization also change the surface presented to cells. Park et al. showed that different cleaning and sterilization procedures altered titanium surface chemistry, hydrophobicity and roughness, and these changes were accompanied by differences in osteoblast-like cell response [15]. Sterilization should therefore be considered part of the material-processing history rather than a neutral administrative step. This is particularly important when comparing studies, because two specimens with the same alloy designation and nominal roughness may not present the same biological surface if their cleaning, sterilization or storage histories differ [11,15].

VIII. BIOMEDICAL APPLICATIONS, STANDARDS, AND TRANSLATIONAL CONSTRAINTS

Orthopedic, dental, spinal and craniofacial devices are exposed to different combinations of load, fixation, contact mechanics and biological conditions. For that reason, “biomedical titanium” is better understood as a family of material–process–surface systems than as a single material choice [9–13]. Dense Ti–6Al–4V remains important where established strength and processing routes are required, β -type alloys are attractive where lower modulus is a priority, and porous AM structures allow stiffness tailoring together with tissue-ingrowth architectures [9,10,13,16–22]. Translation, however, requires more than favorable laboratory data. Composition must be traceable to an appropriate material specification, processing history must be controlled, and the final device should be evaluated in a condition that reflects clinical use, including surface finishing and sterilization.

Current ISO standards make this distinction clear at the material level. ISO 5832-2:2025 specifies characteristics and test methods for unalloyed titanium used in surgical implants, ISO 5832-3:2021 covers wrought Ti–6Al–4V, ISO 5832-11:2024 covers wrought Ti–6Al–7Nb, and ISO 5832-14:2019 specifies wrought Ti–15Mo–5Zr–3Al [28–31]. These standards provide a traceable baseline for material composition and properties, but they do not certify a finished implant. This matters especially for additively manufactured or surface-modified devices, where architecture, internal defects,

machining, heat treatment and coatings can move device-level behavior away from expectations based on the starting material. The ISO 5832 series should therefore be read as material specifications; device-specific mechanical testing, fatigue requirements and regulatory evidence still depend on the implant type and intended use.

Table 4. Selected ISO material standards relevant to titanium surgical implants.

Standard	Material	What it establishes	Critical translational point
ISO 5832-2:2025	Unalloyed titanium	Characteristics and corresponding test methods for surgical-implant titanium	Material specification does not guarantee that a finished device will reproduce standardized specimen properties [28]
ISO 5832-3:2021	Wrought Ti–6Al–4V	Characteristics and test methods for surgical implant alloy	Relevant baseline for conventional Ti–6Al–4V, but AM and surface processing still require device-level qualification [29]
ISO 5832-11:2024	Wrought Ti–6Al–7Nb	Characteristics and test methods	Provides a vanadium-free standardized titanium alloy option for surgical implants [30]
ISO 5832-14:2019	Wrought Ti–15Mo–5Zr–3Al	Characteristics and test methods	Shows that β -rich biomedical alloy systems have recognized implant-material standards, while process route remains application-specific [31]

Sustainability is relevant to titanium implant manufacturing, but broad claims should be made carefully because environmental performance depends on geometry, system boundaries, feedstock production, machine utilization and post-processing. Lyons et al. compared electron-beam melting with milling for a Ti–6Al–4V femoral knee component and reported total CO₂ emissions of approximately 14.94 kg per part for the additive route versus 46.92 kg per part for the conventional route, largely because the machined route required substantially more raw material [35]. This is useful direct evidence that AM can reduce environmental burden for a specific implant geometry, but it should not be generalized to all devices. The result depends on material yield, energy mix, component shape and the full process chain. A meaningful life-cycle comparison should therefore include powder production and conditioning, inert-gas use, failed builds, heat treatment, HIP, finishing, cleaning, sterilization and quality rejection [4,5,8,35]. Performance-aware life-cycle assessment, based on a clearly defined functional unit and service requirement, is more informative than conclusions drawn from material utilization alone.

A more immediate issue is the lack of consistent reporting across studies. Biomedical titanium studies should describe alloy chemistry, powder or feedstock history, process parameters, heat treatment, HIP or other post-processing, defect characteristics, surface roughness and chemistry, architecture and porosity, residual stress where relevant, elastic modulus, strength and fatigue protocol, electrochemical or tribocorrosion conditions, cleaning and sterilization history, and the biological model used [4,5,9–15,18–27]. For porous implants, parameters such as pore size, porosity, interconnectivity, strut thickness and permeability should accompany mechanical data because architecture affects both transport and local stress [13,18–22]. For surface-engineered implants, coating composition, thickness, adhesion, durability and release behavior should be reported alongside osteogenic or antibacterial outcomes [11,12,23]. More complete reporting would make cross-study comparisons more reliable and would also support data-driven optimization across stiffness, fatigue, corrosion, osseointegration, infection resistance, manufacturability and service life.

IX. DISCUSSION

Taken together, the evidence supports the processing–microstructure–biofunction–failure framework and shows why simple design rules are unreliable. β -type alloys demonstrate that elastic modulus can be lowered without necessarily sacrificing high strength, but the resulting performance still depends on phase stability, processing history and surface condition [9,10,16–18]. Porous AM structures can reduce apparent stiffness and support tissue ingrowth, yet the same architecture creates thin struts, local stress concentrations and defect-sensitive fatigue paths [18–22]. Surface roughness can improve in vivo fixation even when cell proliferation is lower, which shows that short-term cell number and long-term osseointegration are not interchangeable outcomes [26]. Likewise, silver-containing coatings can suppress biofilm while preserving osteogenic behavior in some formulations, but this balance depends on concentration and processing rather than

on the presence of silver alone [12,23]. Tribocorrosion studies add another caution: passive corrosion resistance does not predict degradation under loaded contact, especially when fluoride concentration, fretting amplitude or material pairing changes [14,24,25].

Many apparent contradictions in the literature become easier to understand when experimental scale and the dominant failure mechanism are considered. In AM titanium, roughness and internal defects may control fatigue before the chemical environment becomes life-limiting; after HIP or other defect-reduction treatments, corrosion-fatigue effects can become more visible [20,27]. In surface biology, greater roughness may reduce proliferation while promoting osteoblastic differentiation and stronger bone fixation [26]. Likewise, two porous structures with the same nominal porosity can behave differently if pore geometry, strut thickness or defect distribution differs [13,18,21]. These differences do not necessarily indicate poor reproducibility. In many cases, they reflect incomplete reporting of coupled variables. Future studies should therefore describe and control those variables rather than assume that one modulus, pore size, roughness level or coating chemistry will be optimal across applications.

The same caution applies to translation. A favorable result from a polished laboratory coupon, an isolated cell assay or a static corrosion test is not enough to establish the performance of a finished implant. Final geometry, post-processing, surface condition and sterilization can all alter the mechanisms being measured [15,20,24–31]. Material standards provide an essential baseline, but they do not remove the need for device-level mechanical, electrochemical and biological validation [28–31]. Biomedical materials studies are therefore more useful when they make the limits of each test explicit and avoid extending a single favorable endpoint into a claim of overall implant superiority.

Evidence should also be interpreted according to the experimental level at which it was generated. In vitro studies are valuable for isolating mechanisms such as modulus control, ion release, cell differentiation and bacterial suppression [16,17,23,32,33], but they do not reproduce vascularized bone remodeling, immune responses, micromotion or long-term physiological loading. Animal models provide a stronger test of bone integration and infection control and have supported porous titanium osseointegration and Cu-based antibacterial strategies [19,22,34], although differences in species, implantation site and follow-up time limit direct translation to humans. Long-term clinical evidence is still comparatively limited for newer β -type alloys, architected porous implants and multifunctional antibacterial surfaces. These technologies should therefore be viewed mainly as mechanistic or preclinical advances unless device-specific clinical data are available. The review treats these levels of evidence separately rather than combining them into a single claim of clinical superiority.

X. LIMITATIONS OF THE REVIEW

This review has several limitations. It is a structured critical narrative review rather than a formally registered systematic review, so it does not claim exhaustive retrieval of the biomedical titanium literature or support pooled effect estimates. No formal risk-of-bias or study-quality scoring tool was applied; study limitations were considered narratively, which makes cautious interpretation and source-level transparency especially important. The literature is also highly heterogeneous in alloy composition, AM process, surface preparation, defect characterization, loading ratio, fatigue endpoint, electrolyte composition, cell line and animal model, which restricts direct numerical comparison even when studies address similar questions. In addition, much of the evidence for newer β -type alloys, complex porous architectures and multifunctional coatings remains in vitro or preclinical, with less long-term human evidence than is available for commercially pure titanium and Ti–6Al–4V. Finally, the review focuses on material–process–surface interactions and selected implant standards rather than providing a complete regulatory pathway for any one device. These limitations do not invalidate the synthesis, but they do limit how broadly its conclusions can be generalized.

XI. CONCLUSION

Titanium alloys remain foundational biomedical materials because their composition, processing route, architecture and surface state can be tailored to meet demanding mechanical and biological requirements [1,9–13]. The literature reviewed here does not support one-dimensional optimization. Lower modulus, higher porosity, greater roughness or stronger antibacterial activity can be useful in some settings, but each can also introduce penalties through fatigue, corrosion, local stress concentration, coating durability or altered cell response [16–27,32–34]. β -type alloys show that stiffness can be reduced while substantial strength is retained; porous AM architectures can improve mechanical compatibility and osseointegration; post-processing can change the mechanism that limits fatigue; and surface modification can support osseointegration or infection control when chemistry and dose are carefully managed [16–23,26,27,32–34]. Much of the evidence for newer β -type systems, complex porous architectures and multifunctional antibacterial surfaces, however, remains mechanistic or preclinical rather than supported by long-term human outcomes. The main contribution of this review is to show that implant reliability cannot be inferred from alloy chemistry, architecture or surface treatment in isolation. It emerges from the interaction of these factors with defects, post-processing and the biological environment. A practical next step for the field is to evaluate these variables together using implant-specific mechanical, electrochemical and biological endpoints, with transparent reporting of composition, process history, microstructure, defects, architecture, surface

modification, cleaning and sterilization. This approach is more likely to support evidence-based biomedical device design than optimizing individual material properties in isolation.

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