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Perspectives on Metallic Biomaterials for Implants: The Emerging Role of Ti-Nb Alloys and Additive Manufacturing

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Abstract: Metallic biomaterials have long supported orthopedic and dental implants, progressing from stainless steels and Co–Cr alloys to titanium-based systems. Despite decades of clinical success, challenges such as cytotoxic alloying elements, stress shielding from stiffness mismatch, and high production costs remain unresolved. β-stabilized titanium alloys, particularly Ti–Nb, are emerging as strong candidates for next-generation implants. Niobium stabilizes the β-phase at room temperature, enabling lower elastic modulus values that are closer to those of bone, while ensuring corrosion resistance and biocompatibility. When processed by laser powder bed fusion (LPBF), these alloys allow microstructural tailoring, texture control, and improved mechanical compatibility. This perspective reviews the evolution of metallic biomaterials, outlines the limitations of conventional alloys, and discusses the advantages of advanced β-Ti systems. Future directions emphasize integrating alloy design, additive manufacturing, and cost-effective strategies to enable safer, more reliable, and patient-specific biomedical implants.

Keywords: Biomedical implants; β-Ti alloys; Ti–Nb; Stress shielding; Additive Manufacturing; LPBF.

1. Introduction

The performance of engineering materials is inherently linked to the interplay between their microstructure, defined by processing, and the properties required for service [1, 2]. In this context, additive manufacturing (AM) has emerged as a disruptive technology, enabling the design of components with unprecedented complexity and tailored properties [3, 4].

Among AM techniques for metals, laser powder bed fusion (LPBF) has gained particular attention due to its ability to produce customized components with reduced raw material consumption and the capacity to process high-melting-point alloys [5-7]. These advantages, combined with the flexibility to design microstructures layer by layer, enable them to possess specific properties for different types of mechanical stress, as illustrated in Figures 1 [8] and 2 [9], positioning LPBF as a promising route for biomedical applications.

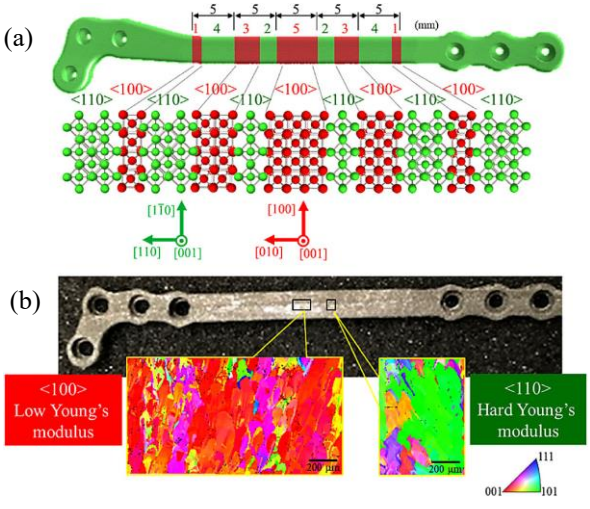


Figure 1. (a) Conceptual illustration showing localized control of texture (mosaic-like texture) in bone plate implant used for fixating fractures. (b) The real product in which mosaic-like texture was developed, fabricated by Hagihara and Nakano [8].

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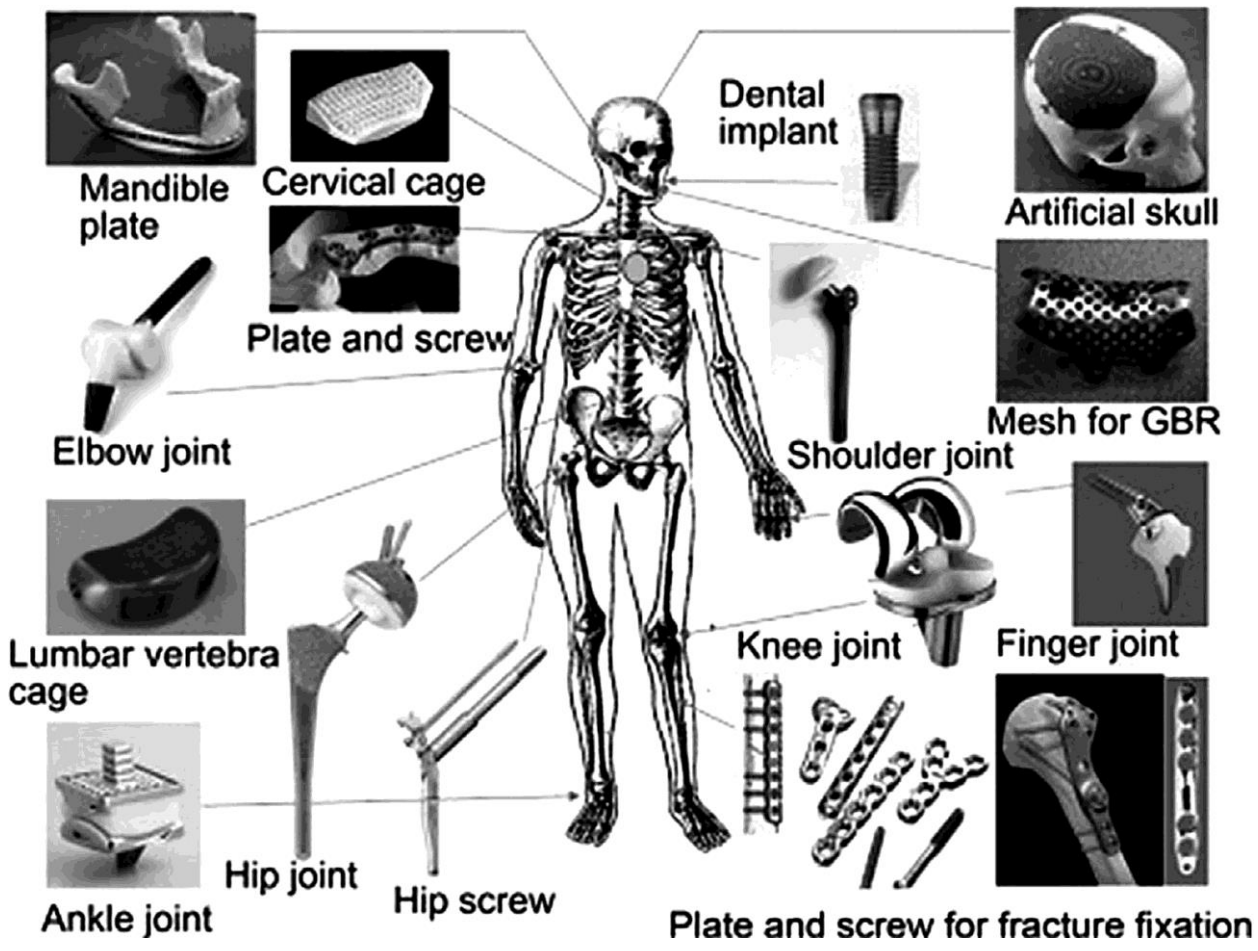


Figure 2. Various load-bearing metallic medical devices used in orthopedic surgery for mechanical support, fixation, and repair of bone tissues [9].

Biomedical implants impose stringent requirements, not only in terms of mechanical performance but also in terms of biological compatibility [2, 10, 11]. Traditional alloys, such as stainless steels, cobalt–chromium alloys, and Ti-6Al-4V, have been widely used in orthopedic and dental applications due to their mechanical strength and corrosion resistance [10, 12]. However, concerns remain regarding their long-term clinical performance. In particular, the release of potentially toxic elements (e.g., Al and V in Ti-6Al-4V), together with the mismatch in stiffness between metallic implants and bone tissue, has motivated the search for new alloy systems with improved clinical relevance [11-12].

One of the significant clinical challenges associated with metallic implants is the phenomenon of stress shielding, which occurs due to the mismatch in elastic moduli between the bone and the implant material [3, 11]. Conventional alloys typically exhibit modulus values significantly higher than those of cortical bone (~30 GPa), resulting in reduced mechanical stimulation and subsequent bone resorption around the implant. Overcoming this problem requires not only new alloy compositions but also processing strategies that can tune the crystalline texture and mechanical response of the material. LPBF, with its unique thermal gradients and high solidification rates, offers opportunities for microstructural customization to reduce stiffness.

Another critical issue concerns the fatigue performance of LPBF-produced implants. Even when porosity is below 0.5 vol.%, surface roughness and microstructural inhomogeneities can significantly compromise fatigue resistance, resulting in scatter in mechanical properties [13, 14]. Thus, the definition of optimized “processing windows” is essential, balancing densification and microstructural control to minimize defects while promoting bone-compatible stiffness [3, 5]. At the same time, economic aspects cannot be overlooked:

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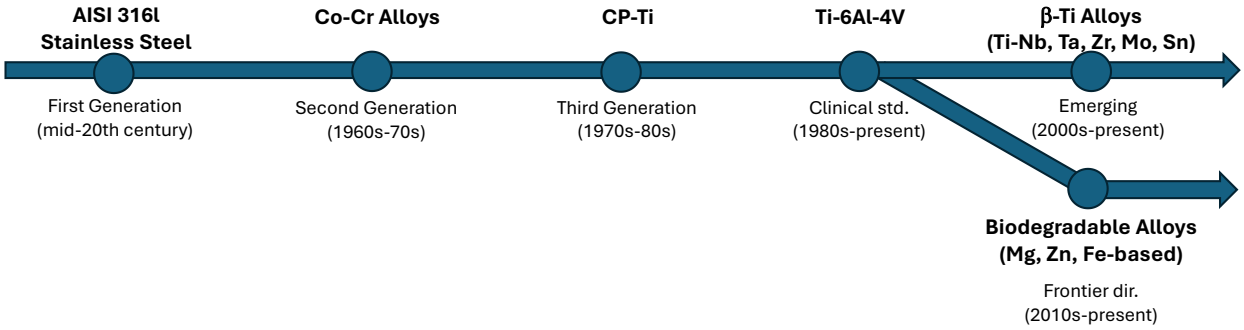


Figure 3. Timeline of metallic biomaterials for biomedical implants.

while spherical powders offer excellent processability, their high cost remains a barrier, particularly in biomedical applications where unitary, patient-specific production dominates. Alternative powder routes, such as hydrogenation–dehydrogenation (HDH), offer cost reductions [15] but demand careful adaptation of processing parameters to avoid undesirable performance [16].

Within this scenario, Ti–Nb alloys emerge as up-and-coming candidates for next-generation implants [16]. Niobium acts as a strong β-phase stabilizer, ensuring phase stability at room temperature while lowering the elastic modulus. Alloys such as Ti–36Nb (at.%) can exhibit modulus values near 60 GPa along specific crystallographic directions—almost half that of Ti-6Al-4V—while maintaining excellent biocompatibility [10, 17]. When combined with LPBF processing, these alloys offer the potential for producing dense, fatigue-resistant, and textured implants with mechanical properties that are more closely aligned with those of bone. This paper aims to provide a perspective on the opportunities and challenges associated with Ti–Nb alloys and LPBF, pointing toward their role as enabling technologies for advanced biomedical implants.

2. Historical Background

The development of metallic biomaterials has closely followed the clinical demands for durable and reliable implants that can restore mobility and function [10]. As shown in Figure 3, the materials development timeline, as discussed throughout this section, occurred in the following order [9, 18]:

1. AISI 316L Stainless Steel → First generation,
2. Co–Cr Alloys → Second generation,
3. CP-Ti → Third Generation,
4. Ti–6Al–4V → Clinical standard,
5. β-Ti Alloys (Ti–Nb, Ti–Ta, Ti–Zr, Ti–Mo, Ti–Sn) → Emerging/future Direction, and,
6. Mg, Zn, Fe-based → biodegradable alloys.

Each category of these metallic biomaterials exhibits distinct advantages and limitations when assessed against the stringent requirements of biomedical applications, including biocompatibility, mechanical reliability, corrosion resistance, and long-term clinical performance. The evolution aimed to develop biodegradable metals, thereby reducing stress shielding and enhancing the performance and durability of resorbable implants.

The earliest generation of metallic implants relied on **austenitic stainless steels**, particularly AISI 316L, due to their relatively low cost, adequate corrosion resistance in physiological environments, and ease of processing [9]. While successful in temporary fixations and specific orthopedic devices, stainless steels presented long-term limitations associated with localized corrosion and the release of nickel ions, raising concerns about hypersensitivity and biocompatibility.

To overcome these issues, **cobalt–chromium alloys (Co–Cr–Mo, Co–Cr–W–Ni)** became the next step in the development of implants. These alloys provided superior wear resistance and mechanical strength, making them suitable for demanding applications such as hip and knee prostheses and heart valves. Despite their advantages, their relatively high elastic modulus—significantly greater than that of bone—limited their biomechanical compatibility. This mismatch often resulted in stress shielding, compromising long-term osseointegration [9].

The search for improved biocompatibility and lower stiffness led to the widespread adoption of **titanium and its alloys**. Commercially pure titanium (CP-Ti) emerged as a milestone due to its remarkable corrosion resistance and excellent tissue compatibility [9, 10]. However, its mechanical strength was insufficient for many load-bearing applications, prompting the introduction of the

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alloy **Ti-6Al-4V**, which offered significantly enhanced strength and reliability. Ti-6Al-4V soon became the clinical standard and remains widely used today [19].

Nevertheless, the presence of aluminum and vanadium in Ti-6Al-4V raised new concerns. Both elements have been associated with cytotoxicity and potential long-term health risks, including neurological and bone-related pathologies [11, 12, 19]. Additionally, the elastic modulus of Ti-6Al-4V (~110 GPa) remained considerably higher than that of cortical bone, perpetuating the stress shielding problem. These limitations have driven research toward new titanium alloys based on **non-toxic β stabilizers**, such as niobium, tantalum, zirconium, molybdenum, and tin [8, 12, 19].

Among these, **Ti–Nb alloys** have gained prominence for their unique combination of biocompatibility, phase stability, and reduced stiffness [17, 19]. By stabilizing the body-centered cubic (BCC) β-phase at room temperature, niobium enables alloys with elastic modulus values closer to those of bone, particularly along specific crystallographic directions. This historical progression—from stainless steels and Co–Cr alloys to advanced β-Ti systems—illustrates the continuous pursuit of materials that not only withstand mechanical loading but also integrate harmoniously with the human body.

Finally, **biodegradable metallic systems**, such as magnesium, zinc, and iron-based alloys, are gaining attention as temporary implants that gradually resorb in vivo, eliminating the need for a second removal surgery [18]. Their main advantage is controlled biodegradation; however, challenges remain in terms of managing corrosion rates, local tissue responses, and ensuring sufficient mechanical integrity during the healing period. Despite these limitations, they represent an exciting frontier for specific biomedical applications, such as cardiovascular stents and pediatric implants.

Taken together, the observed evolution highlights a continuous trade-off between mechanical performance, biological response, and clinical reliability. Although stainless steels and Co-Cr alloys are currently used in cost-sensitive or high-strength applications, titanium alloys—especially β-Ti systems—are emerging as the most promising class for next-generation implants due to their balance of biocompatibility, mechanical properties, and adaptability to additive manufacturing processes [11, 12].

3. Current Challenges in Biomedical Metallic Alloys

Despite the widespread clinical success of metallic implants, several challenges remain unresolved, limiting their long-term performance and clinical reliability. Table A shows an overview of the main issues.

The first and most critical issue is biocompatibility. Conventional alloys such as Ti-6Al-4V, although widely used, incorporate aluminum and vanadium, which have been associated with cytotoxic effects and linked to neuropathies, osteomalacia, and even Alzheimer's disease [11, 12, 19]. Similarly, stainless steels may release nickel, and cobalt–chromium alloys may release cobalt and chromium ions under physiological conditions, raising concerns regarding allergic reactions and systemic toxicity [9]. These limitations underscore the urgent need for alloys based on non-toxic elements, which can ensure safe, long-term interactions with surrounding tissues [20].

Table A. Current challenges in metallic biomaterials for implants and possible strategies.

Challenge	Description	Possible Strategies
Biocompatibility	Release of toxic ions (Al, V, Ni, Co, Cr) leading to cytotoxicity and adverse biological responses.	Use of non-toxic alloying elements (Nb, Ta, Zr, Mo, Sn); development of β-Ti alloys; surface functionalization.
Stress shielding	Elastic modulus of conventional alloys (~110–220 GPa) much higher than cortical bone (~30 GPa), causing bone resorption.	Design of β-stabilized titanium alloys (e.g., Ti–Nb); texture tailoring via LPBF; architectured porosity design.
Fatigue resistance	Porosity and surface roughness in AM parts reduce fatigue life and increase scatter in S–N curves.	Definition of optimized processing windows; surface finishing; HIP; microstructural refinement.
Economic feasibility	High cost of spherical powders limits large-scale clinical adoption.	Use of cost-effective powders (HDH); improvement of powder spreading techniques; recycling strategies.

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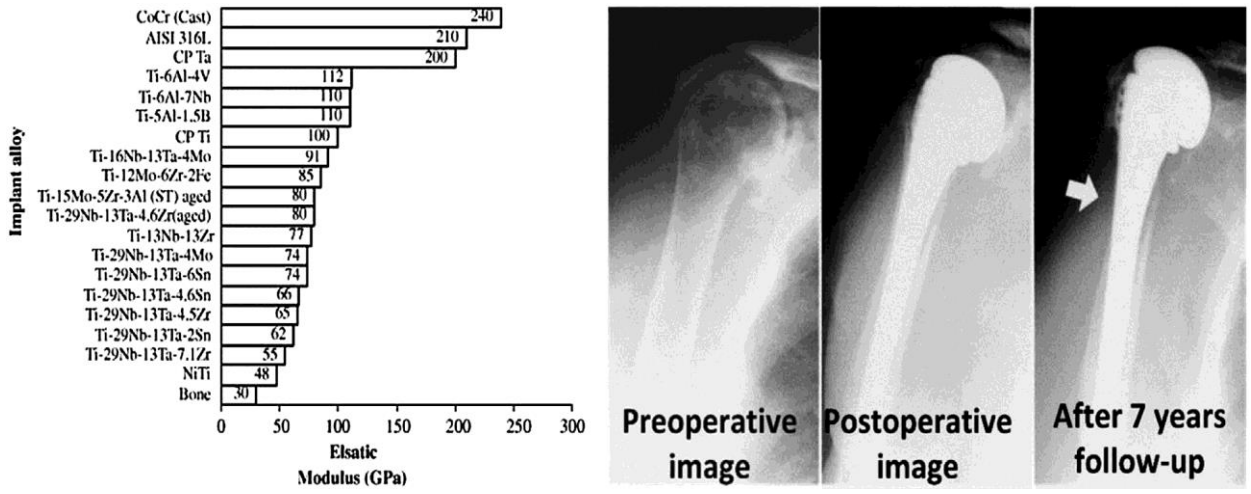


Figure 4. Effect of higher E of implant materials on stress shielding of surrounding bones: (A) evolution of E compatibility of biomedical grade alloys and (B) example of stress-shielding effect in joint replacement implants were the white arrow indicates the region of bone resorption due to stress-shielding effect [10].

Another major challenge is the phenomenon of stress shielding, which occurs due to the mismatch in elastic modulus between metallic implants and natural bone. Whereas cortical bone exhibits a modulus of around 30 GPa, traditional alloys present values well above 100 GPa, thereby reducing the mechanical stimulus required for bone remodeling [11, 12, 19, 20]. As a result, bone resorption occurs around the implant, ultimately compromising fixation and longevity. Strategies to mitigate this issue include not only the development of new alloy compositions, as shown in Figure 4, but also the engineering of crystallographic textures through processing routes such as LPBF, which can tailor elastic anisotropy to achieve stiffness values closer to bone [11, 21].

Fatigue resistance represents a further limitation, particularly in load-bearing orthopedic implants produced by additive manufacturing. Even with porosity levels below 0.5 vol.%, microstructural heterogeneities and surface roughness can severely compromise fatigue life [14, 16]. This issue introduces large scatter in S–N behavior, reducing reliability in service. Addressing this challenge requires defining optimized processing windows that minimize porosity, control microstructural features, and ensure mechanical performance under cyclic loading. Post-processing treatments such as surface polishing or hot isostatic pressing (HIP) can improve fatigue resistance, but they increase costs and reduce the economic advantage of AM [16].

Ultimately, economic considerations are playing an increasingly important role in the clinical adoption of advanced biomaterials. Spherical powders, which offer excellent flowability and reproducibility in LPBF, are prohibitively expensive, especially when considering patient-specific production in relatively low volumes. Alternatives, such as hydrogenation–dehydrogenation (HDH) powders, can reduce raw material costs by up to 80% [15]. However, their irregular morphology affects powder spreading and increases the risk of defects [16]. Developing robust strategies to use cost-effective powders without compromising implant performance is therefore essential to expand the accessibility of advanced metallic implants.

Taken together, these challenges illustrate the complex balance between biological safety, mechanical reliability, and economic viability. Overcoming them requires an integrated approach that combines alloy design, advanced processing technologies, and cost-effective production strategies. In this regard, Ti–Nb alloys produced by LPBF stand out as an up-and-coming solution, addressing biocompatibility, modulus compatibility, and fatigue resistance, while offering opportunities to reduce production costs through optimized powder usage [21].

4. Perspectives for Advanced Titanium Alloys

The growing demand for metallic biomaterials that combine biocompatibility, mechanical compatibility, and long-term reliability has placed titanium alloys at the center of research efforts. While CP-Ti and Ti-6Al-4V have historically dominated clinical applications, their limitations in stiffness mismatch and the presence of cytotoxic alloying elements have accelerated the exploration of **advanced β-stabilized titanium alloys**.

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Systems alloyed with niobium, tantalum, zirconium, molybdenum, and tin have emerged as the most promising candidates due to their non-toxic nature and ability to stabilize the body-centered cubic (BCC) β-phase at room temperature. Several β-stabilized titanium systems have been investigated to address the limitations of conventional implant materials.

Ti–Ta alloys are among the most attractive due to tantalum's exceptional corrosion resistance and proven biocompatibility. Tantalum acts as a potent β stabilizer, allowing the development of alloys with relatively low elastic modulus values, while simultaneously enhancing resistance to physiological degradation. Although pure tantalum is prohibitively dense and challenging to process, alloying it with titanium combines the mechanical and weight advantages of Ti with the biological performance of Ta. These alloys are therefore of particular interest in load-bearing implants where long-term stability is required [22].

Ti–Zr alloys represent another vital class, benefiting from zirconium's strong biocompatibility and corrosion resistance. Zirconium does not pose cytotoxic risks and is already utilized in biomedical applications, such as dental implants and orthopedic devices. In titanium alloys, Zr contributes to solid solution strengthening without significantly altering density, while also reducing the elastic modulus. Ti–Zr alloys are especially attractive in cases where enhanced corrosion resistance is crucial, such as in implants exposed to aggressive physiological environments [19, 20].

Ti–Mo alloys utilize molybdenum as a strong β-stabilizer, resulting in stable body-centered cubic (BCC) structures at room temperature. Molybdenum not only reduces the elastic modulus but also contributes to improved corrosion resistance. Ti–Mo alloys have been studied for orthopedic and cardiovascular applications, though their relatively high density and cost compared to Nb and Zr remain considerations for large-scale clinical adoption. Nonetheless, they remain a promising option when superior mechanical and corrosion properties are prioritized [11, 19].

Ti–Sn alloys have recently attracted attention as an alternative β-stabilized system. Tin is considered a neutral or weak β stabilizer, but it plays a critical role in enhancing ductility and corrosion resistance while maintaining biocompatibility. Ti–Sn alloys can be combined with other β-stabilizers, such as Nb or Mo, to fine-tune their mechanical response. Their ability to promote stable mechanical performance and cytocompatibility has positioned them as candidates for new generations of biomedical alloys [19].

Ti–Nb alloys, in particular, offer unique advantages. Niobium acts as a strong β stabilizer, enabling alloys with lower elastic modulus values, in some cases approaching ~60 GPa along the <100> crystallographic direction. This modulus is nearly half that of Ti-6Al-4V and significantly closer to cortical bone (~30 GPa), making Ti–Nb alloys especially effective in mitigating stress shielding. Beyond their mechanical compatibility, Ti–Nb alloys exhibit excellent corrosion resistance in physiological environments and high cytocompatibility, supporting their long-term clinical viability. These characteristics align them with the next generation of biomaterials, which not only resist degradation but also actively integrate with surrounding tissues [16, 17].

A further advantage of Ti–Nb systems is their suitability for **additive manufacturing (AM)**, particularly laser powder bed fusion (LPBF) [8, 17]. The rapid solidification and steep thermal gradients characteristic of LPBF enable the engineering of microstructures and crystallographic textures tailored to meet biomedical requirements. For instance, the induction of <100> texture in β-Ti alloys can further reduce the effective elastic modulus, while careful control of the processing window can minimize porosity and enhance fatigue resistance. This synergy between alloy design and processing flexibility highlights the potential of LPBF to unlock unprecedented levels of property customization [21].

Looking forward, research on advanced titanium alloys must address not only microstructural tailoring but also **economic and clinical feasibility**. Strategies to reduce powder costs, such as employing hydrogenation–dehydrogenation (HDH) routes without compromising processability, are crucial for broadening the clinical adoption of these materials [15, 16]. Additionally, long-term *in vivo* studies will be essential for validating the performance of Ti–Nb implants under complex physiological loading conditions. Beyond these aspects, the convergence of digital design, alloy development, and additive manufacturing represents a robust integrated approach, enabling the creation of implants that are not only biocompatible but also optimized for patient-specific biomechanical requirements [9, 18, 19].

Taken together, as summarized in Table B, these advanced titanium alloys expand the design space for next-generation implants. Each element brings unique benefits: Ta for corrosion resistance, Zr for enhanced biocompatibility, Mo for modulus reduction and corrosion resistance, and Sn for ductility and stability. When

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Table B. Comparative overview of advanced titanium alloys for biomedical implants obtained by LPBF.

Alloy System	Main Alloying Element	Key Benefits	Limitations / Considerations
Ti–Nb	Niobium (β stabilizer)	Low elastic modulus (~60 GPa in <100>); strong biocompatibility; corrosion resistance; stable β phase at room temperature.	Cost of Nb; requires optimized LPBF processing to minimize porosity.
Ti–Ta	Tantalum (β stabilizer)	Excellent corrosion resistance; high biocompatibility; stable modulus reduction.	High density; high cost; pure Ta is difficult to process.
Ti–Zr	Zirconium (neutral to β stabilizer)	Biocompatible and non-toxic; enhanced corrosion resistance; solid solution strengthening.	Limited modulus reduction; requires combination with other stabilizers.
Ti–Mo	Molybdenum (strong β stabilizer)	Significant β-phase stabilization; improved corrosion resistance; modulus reduction.	Higher density and cost compared to Nb and Zr.
Ti–Sn	Tin (weak β stabilizer)	Enhances ductility; improves corrosion resistance; cytocompatible; synergistic with other stabilizers.	Weak β stabilizer alone; best used in combination with Nb, Mo, etc.

combined with additive manufacturing technologies, these alloys offer unprecedented opportunities for customizing implant performance to match patient-specific biomechanical needs. However, further research is needed to balance mechanical properties, processing feasibility, and cost, ensuring their effective translation into clinical practice.

Particularly, Ti–Nb alloys stand out as **enabling materials** for the next generation of biomedical implants. Their combination of low stiffness, superior biocompatibility, and compatibility with AM technologies positions them as frontrunners in overcoming the limitations of traditional alloys. The convergence of materials science, processing innovations, and biomedical engineering places these alloys at the forefront of efforts to reduce stress shielding, enhance fatigue resistance, and ensure the long-term success of metallic implants [7, 8, 9].

5. Future Directions and Clinical Relevance

The future of metallic biomaterials for biomedical implants lies in the convergence of alloy design, advanced processing techniques, and clinical translation. In particular, β-stabilized titanium alloys, such as Ti–Nb and related systems, stand out as frontrunners due to their superior biocompatibility, corrosion resistance, and elastic modulus values that are closer to those of bone. When processed by additive manufacturing, these alloys offer unprecedented opportunities to tailor microstructures, induce crystallographic textures, and design patient-specific implants with optimized biomechanical compatibility.

A key direction for future research is the integration of additive manufacturing with alloy development, allowing simultaneous control of geometry, microstructure, and mechanical response. For example, the induction of <100> textures in Ti–Nb alloys by LPBF can significantly reduce effective stiffness, directly addressing stress shielding [21]. Coupling such advances with controlled porosity and surface functionalization will enable implants that not only mechanically integrate with bone but also promote osseointegration and long-term stability.

Another critical aspect is the translation of laboratory results into clinical validation. While many β-Ti alloys have demonstrated promising properties in vitro and in pre-clinical studies, long-term in vivo evaluations remain scarce. Clinical trials assessing the performance of Ti–Nb and other advanced alloys under real physiological conditions will be essential to confirm their safety, durability, and functional benefits compared to current standards, such as Ti-6Al-4V. Establishing robust databases of clinical outcomes will also accelerate regulatory approval and widespread adoption.

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From an economic and practical perspective, future developments must also address the cost-effectiveness and scalability of advanced implant production. The high cost of spherical powders limits the accessibility of AM-based implants. Expanding the use of alternative feedstocks, such as HDH powders, in combination with optimized processing strategies, could significantly reduce production costs without compromising performance. Achieving this balance will be vital to ensure that advanced implants are not only technologically superior but also financially viable for healthcare systems worldwide.

In summary, the clinical relevance of advanced titanium alloys lies in their potential to create the next generation of metallic implants: safer, more reliable, and tailored to the biomechanics of individual patients. By combining biocompatible alloy chemistries, processing innovations through LPBF, and cost-efficient production strategies, Ti–Nb and related β-Ti systems are poised to redefine the standard of care in orthopedic and dental implants. The coming years will determine how quickly these laboratory advances can transition into clinical practice, ultimately shaping the future of metallic biomaterials in modern medicine.

6. Conclusions

The evolution of metallic biomaterials reflects the constant pursuit of safer and more reliable solutions for biomedical implants, as summarized in Figure 5. While stainless steels, cobalt–chromium alloys, and Ti-6Al-4V have provided decades of clinical success, their limitations—particularly stress shielding and the release of potentially toxic elements—underscore the need for next-generation alloys. β-stabilized titanium systems, especially Ti–Nb, represent a paradigm shift by offering lower stiffness, superior biocompatibility, and compatibility with additive manufacturing.

Looking ahead, the convergence of **biocompatible alloy design**, **advanced processing technologies such as LPBF**, and **cost-effective production strategies** will be decisive in defining the future of biomedical implants. Ti–Nb and related β-Ti alloys stand out as enabling materials, not only to overcome the drawbacks of conventional alloys but also to establish a new standard for patient-specific, long-lasting, and clinically relevant implants. The translation of these advances from laboratory studies to clinical practice will be essential to realize their full potential in modern medicine.

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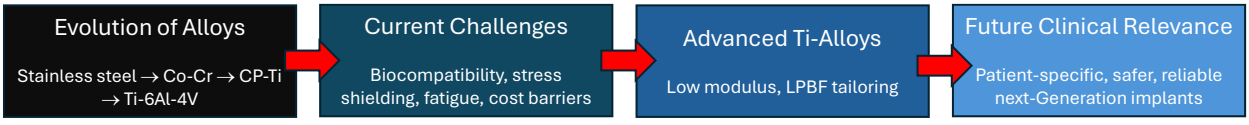


Figure 5. Roadmap for the Development of Advanced Titanium Alloys in Biomedical Implants.

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