SURFACE ENGINEERING POSSIBILITIES IN DMLS TITANIUM IMPLANTS: STRATEGIES FOR ENHANCING BIOLOGICAL PERFORMANCE

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Abstract

Direct Metal Laser Sintering has emerged as a key additive manufacturing technique for producing titanium-based biomedical implants with complex geometries and tailored porous architectures. Despite these advantages, as-built DMLS titanium surfaces exhibit limited bioactivity and insufficient antibacterial properties, which may compromise long-term clinical performance. This review provides a comprehensive and multiscale analysis of surface engineering strategies aimed at enhancing the biological performance of DMLS titanium implants. Mechanical treatments, chemical modifications, bioactive coatings, and antibacterial functionalization approaches are critically evaluated with respect to their mechanisms and biological outcomes. Particular emphasis is placed on the interplay between surface topography, chemistry, and wettability in regulating protein adsorption, cell behavior, and bacterial adhesion. The analysis highlights a fundamental trade-off between promoting osseointegration and preventing bacterial colonization, which remains a central challenge in implant design. Comparative assessment of existing methods indicates that no single modification strategy fully satisfies clinical requirements. Instead, multifunctional and hybrid surface approaches that integrate bioactive and antibacterial features demonstrate the greatest potential. Key challenges, including process variability, coating stability, scalability, and limited clinical validation, are also discussed. Future directions focus on the development of bioinstructive and stimuli-responsive surfaces, as well as the integration of data-driven optimization methods. Overall, advancing DMLS implant performance requires coordinated multiscale surface engineering combined with clinically relevant validation strategies.

Keywords: *DMLS; titanium implants; surface modification; osseointegration; additive manufacturing; biomaterials*

Introduction

Additive manufacturing (AM) technologies have significantly advanced the field of biomedical engineering by enabling the fabrication of patient-specific implants with complex geometries and tailored mechanical properties [1-3]. Among these techniques, Direct Metal Laser Sintering (DMLS), Selective Laser Melting (SLM), and Electron Beam Melting (EBM) are widely used for processing titanium-based biomaterials [4-6]. Compared to SLM and EBM, DMLS typically produces surfaces with higher roughness and a greater number of partially sintered particles, which significantly influence biological response and surface modification efficiency [7-8].

DMLS has emerged as a particularly promising method for processing titanium and its alloys, especially Ti-6Al-4V, which are widely used in orthopedic and dental applications due to their excellent mechanical strength, corrosion resistance, and biocompatibility [4-7]. A key advantage of DMLS lies in its ability to produce controlled porous architectures that mimic the hierarchical structure of natural bone, thereby improving mechanical compatibility and reducing stress shielding [8]. In addition, the inherent surface roughness of as-built DMLS components may enhance initial cell attachment [9-11].

However, despite these advantages, DMLS-fabricated titanium surfaces typically exhibit limited bioactivity and lack intrinsic antibacterial properties, which may compromise long-term implant performance [12-13]. Surface properties – including roughness, chemistry, wettability, and topography – play a critical role in regulating biological responses at the implant-tissue interface [14-16]. Consequently, surface modification has become an essential strategy for improving the performance of DMLS titanium implants [14]. A wide range of approaches has been explored, including mechanical treatments, chemical etching, bioactive coatings, and advanced physicochemical functionalization techniques [14]. Despite extensive research on surface modification of conventionally manufactured titanium, the direct translation of these strategies to additively manufactured materials remains challenging [17-18]. The unique surface morphology, high roughness, and intrinsic porosity of DMLS-produced titanium significantly alter surface-cell and surface-bacteria interactions [9,10,13]. Therefore, approaches developed for wrought or machined titanium cannot be directly extrapolated to DMLS structures [19].

This review provides a critical and multiscale perspective on surface engineering of DMLS titanium implants, with particular emphasis on the interplay between surface properties and biological response. Special attention is given to the balance between osseointegration and antibacterial functionality, as well as to current limitations and future directions for clinically relevant implant design.

Characteristics of DMLS Titanium

Titanium and its alloys fabricated by DMLS exhibit distinct microstructural and surface characteristics compared to conventionally manufactured materials [20]. These features are primarily governed by rapid melting-solidification cycles and process parameters such as laser power, scan speed, hatch spacing, and layer thickness [20,21]. A characteristic feature of DMLS Ti-6Al-4V is the formation of a fine acicular martensitic α′ phase due to extremely high cooling rates [17,18]. This microstructure results in increased strength but reduced ductility relative to wrought materials [19]. Post-processing treatments, such as heat treatment or hot isostatic pressing (HIP), are therefore commonly employed to tailor mechanical properties [20]. In addition to bulk microstructure, DMLS components exhibit high surface roughness caused by partially melted powder particles adhering to the surface [21]. This results in a complex hierarchical topography at micro- and submicron scales. While such roughness can promote cell attachment and mechanical interlocking with bone, excessive irregularity may facilitate bacterial adhesion and biofilm formation

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[9-12]. Porosity represents another defining feature of DMLS materials. Designed porosity enhances bone ingrowth and vascularization; however, process-induced defects such as lack-of-fusion pores may compromise mechanical integrity [21,22]. Importantly, surface-connected porosity can also influence the effectiveness and uniformity of subsequent surface modification treatments [23]. Overall, the combination of unique microstructure, roughness, and porosity makes DMLS titanium highly attractive for biomedical applications, while simultaneously necessitating dedicated surface engineering strategies.

Surface Modification Techniques

Surface modification of DMLS titanium implants is a critical strategy for improving their biological performance. From a biomaterials perspective, surface modification strategies aim to control surface energetics and interfacial interactions at multiple length scales, which determines their effectiveness in additively manufactured titanium [14,24].

Mechanical and Physical Treatments

Mechanical and physical treatments represent the most straightforward approaches for modifying the surface of DMLS titanium. Common techniques include sandblasting, polishing, and shot peening, which are primarily used to remove loosely adhered particles and to adjust surface roughness [25,26].

Sandblasting, typically performed using alumina particles, significantly increases surface roughness and has been shown to enhance osteoblast activity [24,25]. For example, studies have reported up to a ~30-40% increase in cell proliferation on sandblasted Ti-6Al-4V surfaces compared to untreated samples after 7 days of *in vitro* culture [25,27]. This effect is commonly accompanied by increased alkaline phosphatase (ALP) activity, indicating enhanced early osteogenic differentiation [25]. It should be noted, however, that ALP activity reported in the literature may refer either to total ALP or to its bone-specific isoform (B-ALP), with the latter being a more specific marker of osteogenic differentiation. In many cases, studies do not explicitly distinguish between these forms, which may limit direct comparison of reported osteogenic outcomes. However, the relationship between surface roughness and biological response is non-linear. An optimal roughness window has been identified, typically in the range of $Ra \approx 1\text{-}3\ \mu m$, where osteoblast adhesion and differentiation are maximized. [24,28] Surfaces with excessive roughness ($Ra > 5\ \mu m$) tend to promote bacterial adhesion and biofilm formation, while overly smooth surfaces ($Ra < 0.5\ \mu m$) may reduce osteogenic potential due to limited mechanical interlocking [12-13].

Polishing, in contrast, reduces surface irregularities and is often applied in applications requiring low friction or wear resistance. While smoother surfaces may reduce bacterial retention, they generally exhibit lower bioactivity compared to moderately rough surfaces [26]. Laser-based treatments, such as surface remelting, provide additional control over surface morphology by locally modifying the top layer. These processes can reduce surface porosity and improve uniformity; however, they may also diminish beneficial roughness features associated with enhanced osseointegration [23]. Overall, mechanical and physical treatments are effective for tailoring microscale topography, but their impact on biological performance is highly dependent on precise control of surface parameters.

Chemical Treatments

Chemical modification methods are widely used to alter the surface chemistry and wettability of titanium implants. Acid etching, typically involving strong acids such as hydrochloric or sulfuric acid, creates micro- and nanoscale features that improve surface area and promote protein adsorption [29]. Alkali treatment is another effective approach, leading to the formation of sodium titanate layers on the surface. These layers can subsequently induce the formation of apatite when exposed to physiological environments, thereby enhancing bioactivity [30,31]. Combined treatments, such as acid-alkali sequences, are often employed to achieve hierarchical surface structures. Importantly, chemical treatments can also increase surface hydrophilicity, which has been shown to positively influence cell attachment and differentiation [15,32]. However, process parameters must be carefully controlled, as excessive chemical exposure may weaken the surface or introduce contaminants [29].

Bioactive Coatings

The application of bioactive coatings represents one of the most effective strategies for enhancing the performance of DMLS titanium implants [33]. Among these, hydroxyapatite (HA) coatings are the most extensively studied, as they closely resemble the mineral component of bone and promote rapid osseointegration. Various deposition techniques have been employed, including plasma spraying, sol-gel processing, and electrophoretic deposition [33,34]. Each method offers specific advantages and limitations in terms of coating thickness, adhesion strength, and uniformity, particularly on complex porous structures typical of DMLS implants. In addition to HA, other bioactive materials such as bioactive glasses and calcium phosphate coatings have been investigated [35,36]. These coatings can enhance bone bonding; however, their long-term stability and mechanical integrity remain critical concerns, especially under cyclic loading conditions [33].

Antibacterial Surface Modifications

Implant-associated infections remain a major clinical challenge; therefore, the development of antibacterial surfaces is of significant interest. Various strategies have been explored to impart antimicrobial properties to DMLS titanium. One common approach involves the incorporation of antibacterial agents such as silver (Ag), zinc (Zn), or copper (Cu) ions into coatings or surface layers [37,38]. These ions can disrupt bacterial cell membranes and inhibit biofilm formation. However, their concentration must be carefully optimized to avoid cytotoxic effects on surrounding tissues [39]. Another strategy includes the use of antibacterial coatings based on polymers or nanostructured materials. Surface nanotopography itself can also influence bacterial adhesion, with certain nanostructured surfaces exhibiting bactericidal effects while remaining biocompatible [40,41].

Advanced and Emerging Approaches

Recent research has focused on advanced surface engineering techniques that combine multiple functionalities. Plasma treatments, for example, can modify surface energy and introduce reactive functional groups, improving both wettability and coating adhesion. Additionally, laser-based patterning and surface functionalization techniques enable precise control over surface morphology at micro- and nanoscale levels. These approaches can be used to create hierarchical structures that simultaneously promote osteogenesis and inhibit bacterial colonization [42,43].

Hybrid strategies, combining chemical treatments with bioactive or antibacterial coatings, are increasingly being explored to achieve synergistic effects [43,44]. These approaches reflect the need to engineer surface properties across multiple scales to optimize interfacial biological responses [45]. Although these approaches show significant promise, their scalability and reproducibility remain important challenges for clinical translation [46]. Moreover, individual surface modification techniques demonstrate clear benefits, but their effectiveness is highly dependent on the interplay between surface topography, chemistry, and implant geometry [45]. Mechanical treatments primarily influence roughness at the microscale, whereas chemical and physicochemical approaches enable control at the nano- and molecular levels. Importantly, no single technique provides a comprehensive solution [29,47]. For example, sandblasting improves osteoblast activity but may increase bacterial adhesion, whereas antibacterial coatings can reduce infection risk but introduce concerns related to cytotoxicity and long-term stability [12,39]. Therefore, current research increasingly focuses on hybrid and multifunctional approaches that integrate multiple surface modification strategies to achieve synergistic biological effects [48]. A comparative summary of the most commonly applied surface modification strategies is presented in TABLE 1.

Biological Performance

The biological performance of DMLS titanium implants is strongly influenced by their surface properties, including topography, chemistry, roughness, and wettability. Surface modification strategies play a crucial role in regulating cellular responses and determining the overall success of implantation [49].

Osseointegration

Osseointegration is a key factor in the long-term stability of orthopedic and dental implants. The porous architecture achievable by DMLS promotes bone ingrowth and mechanical interlocking; however, surface modifications are often necessary to enhance biological fixation at the bone-implant interface [8,50].

Micro- and nanoscale surface features have been shown to stimulate osteoblast activity and accelerate new bone formation. In particular, chemically treated and bioactive-coated surfaces, such as those modified with hydroxyapatite or calcium phosphate, significantly improve bone bonding. Additionally, increased surface wettability enhances protein adsorption, which is essential for early-stage osseointegration [51,52].

Cell Adhesion and Proliferation

Cellular response to implant surfaces can be understood in terms of structure-property-function relationships, where physical and chemical cues jointly regulate biological outcomes. At the molecular level, cell-material interactions

TABLE 1. Summary and comparison of surface modification strategies applied to DMLS titanium implants, including their mechanisms, advantages, limitations, and biological outcomes.

Modification method	Scale of effect	Typical parameters	Key mechanism	Main advantages	Limitations	Biological outcome	Clinical relevance	Ref.
Sandblasting	Micro	Ra ~1–5 μm	Increased roughness, mechanical interlocking	Enhanced cell adhesion	Bacterial retention	↑ osteoblast activity	Widely used (dental implants)	[24,28]
Acid etching	Micro/ Nano	Micro-pits + nano-features	Increased surface area, wettability	Improved protein adsorption	Risk of surface damage	↑ cell adhesion, differentiation	Standard in implantology	[29,32]
Alkali treatment	Nano	Sodium titanate layer	Induces apatite formation	High bioactivity	Process sensitivity	↑ osseointegration	Experimental/ limited clinical	[30,31]
HA coatings	Macro/ Micro	Thickness: 10–100 μm	Biomimetic composition	Strong bone bonding	Delamination risk	↑ rapid osseointegration	Clinically used	[33,34]
Bioactive glass	Micro/ Nano	Thin coatings	Ion release, bioactivity	Osteostimulation	Brittleness	↑ bone regeneration	Emerging	[35,36]
Ag/Zn/Cu coatings	Nano	Ion concentration-dependent	Antibacterial ion release	Strong antimicrobial effect	Cytotoxicity risk	↓ bacterial adhesion	Limited clinical use	[37-39]
Plasma treatment	Nano	Surface activation	Increased surface energy	Improved wettability	Short-term effect	↑ cell attachment	Pre-treatment step	[42]
Laser modification	Micro/ Nano	Patterning / re-melting	Topography control	High precision	Equipment cost	Tunable response	Experimental	[23,42]
Hybrid coatings	Multi-scale	Combined methods	Synergistic effects	Multifunctionality	Complexity	↑ bio + antibacterial	Future direction	[43,48]

are mediated by integrin receptors, which bind to adsorbed extracellular matrix proteins such as fibronectin and vitronectin [53,54]. This binding activates intracellular signaling pathways, including focal adhesion kinase (FAK), mitogen-activated protein kinase (MAPK), and RhoA (Ras homolog family member A) / ROCK (Rho-associated coiled-coil containing protein kinase) pathways, which regulate cytoskeletal organization, cell spreading, and differentiation [54,55]. Surface roughness and nanotopography influence the spatial distribution and clustering of integrins, thereby modulating focal adhesion formation and downstream signaling. In particular, nanoscale features have been shown to enhance osteogenic differentiation by upregulating the expression of osteogenic marker genes (e.g., *RUNX2, ALPL, BGLAP*), which encode proteins such as RUNX2, alkaline phosphatase (ALP), and osteocalcin.[51,56].

Therefore, the biological performance of DMLS titanium is not solely a function of surface morphology but also of its ability to regulate mechanotransduction pathways at the cell-material interface [55]. Surface roughness at the microscale can promote initial cell attachment, while nanoscale features influence cell differentiation and signaling pathways. These interactions are mediated by interfacial phenomena occurring at the cell- material interface [24,53,56].

Studies have demonstrated that DMLS titanium surfaces subjected to acid etching or alkali treatment exhibit improved adhesion and proliferation of osteoblast-like cells [29,30]. Furthermore, surface functionalization with bioactive coatings can enhance cellular activity by providing biochemical cues that mimic the natural extracellular matrix [33]. However, excessively rough or irregular surfaces may negatively affect cell spreading and lead to heterogeneous cellular responses. Therefore, optimization of surface characteristics is essential to balance mechanical interlocking and biological compatibility [53].

Antibacterial Properties

Bacterial adhesion and biofilm formation on implant surfaces represent a major cause of implant failure. The inherent roughness of DMLS titanium may facilitate bacterial colonization if not properly modified [13,37].

Surface engineering strategies aimed at reducing infection risk include the incorporation of antibacterial agents such as silver, zinc, or copper ions, as well as the development of antibacterial coatings and nanotextured surfaces. These modifications can inhibit bacterial growth through mechanisms such as membrane disruption, ion release, or prevention of bacterial adhesion [37-39]. At the mechanistic level, antibacterial effects are typically associated with three main pathways: (i) disruption of bacterial cell membranes through direct contact with nanostructured surfaces or released ions, (ii) generation of reactive oxygen species (ROS), leading to oxidative stress, and (iii) interference with bacterial metabolic processes and DNA replication [39].

Nanostructured surfaces inspired by natural bactericidal systems, such as cicada or dragonfly wings, have demonstrated the ability to physically rupture bacterial membranes while remaining compatible with mammalian cells. This mechano-bactericidal effect represents a promising alternative to chemical-based antibacterial strategies [40]. Importantly, the challenge lies in achieving a balance between antibacterial activity and cytocompatibility. High concentrations of antibacterial agents may induce cytotoxic effects, while insufficient concentrations may not effectively prevent infection. Therefore, controlled release systems and multifunctional coatings are increasingly being investigated [39,41].

Overall, surface modification of DMLS titanium implants significantly enhances their biological performance. Improved osseointegration, increased cell adhesion, and reduced bacterial activity have been consistently reported across various modification techniques. Nevertheless, the relationship between specific surface parameters and biological outcomes remains complex and not fully understood, highlighting the need for further systematic studies [24,51,53].

Protein Adsorption and Immune Response

Beyond osteoblast activity and antibacterial performance, the initial biological response to DMLS titanium implants is strongly governed by protein adsorption and immune system interactions. Immediately after implantation, proteins such as fibronectin, albumin, and vitronectin adsorb onto the implant surface, forming a conditioning layer that mediates subsequent cell attachment [53,54]. Surface properties such as wettability, charge, and nanoscale topography significantly influence the composition and conformation of adsorbed proteins [15,32]. This highlights the critical role of surface energetics in governing early-stage biological interactions. Hydrophilic and nanostructured surfaces have been shown to promote favorable protein configurations that enhance integrin-mediated cell adhesion [32,56].

In addition, macrophage response plays a critical role in implant integration. Surface-modified titanium can modulate macrophage polarization toward a pro-healing (M2) phenotype, reducing inflammation and promoting tissue regeneration [57,58]. However, the interaction between DMLS-specific surface features and immune response remains insufficiently understood and requires further investigation [20]. Taken together, the interaction between DMLS titanium implants and the biological environment can be described as a hierarchical process involving multiple interconnected stages: (I) initial protein adsorption governed by surface chemistry and energy, (II) cell adhesion mediated by integrin binding and focal adhesion formation, (III) activation of intracellular signaling pathways and gene expression, and (IV) tissue-level responses such as bone formation and immune modulation [53-55]. This multiscale framework highlights that successful implant integration requires simultaneous optimization of surface properties across nano-, micro-, and macro-scales, rather than focusing on a single parameter [45].

The biological response to DMLS titanium implants can be interpreted within a multiscale framework. Surface modifications influence physicochemical properties, which regulate protein adsorption and subsequent cell signaling pathways [53]. At the microscale, topography governs cell adhesion and mechanotransduction, while at the macroscale, porosity determines tissue integration [8,45]. This hierarchical interaction ultimately defines the balance between osseointegration and bacterial colonization [37].

A schematic overview of the multiscale interactions governing the biological performance of DMLS titanium implants is presented in FIG. 1.

Comparative Analysis of Surface Modification Strategies

A direct comparison of surface modification techniques reveals that no single approach provides a comprehensive solution for optimizing the biological performance of DMLS titanium implants. Mechanical treatments primarily influence microscale roughness and are effective in enhancing initial cell adhesion [24,47]. However, their benefits are limited by increased susceptibility to bacterial colonization at higher roughness levels. In contrast, chemical treatments enable control at micro- and nanoscale levels and significantly improve wettability and protein adsorption, which are critical for early-stage cell–material interactions 29,32]. Bioactive coatings, particularly hydroxyapatite, provide the most direct enhancement of osseointegration due to their biomimetic composition. Nevertheless, their long-term mechanical

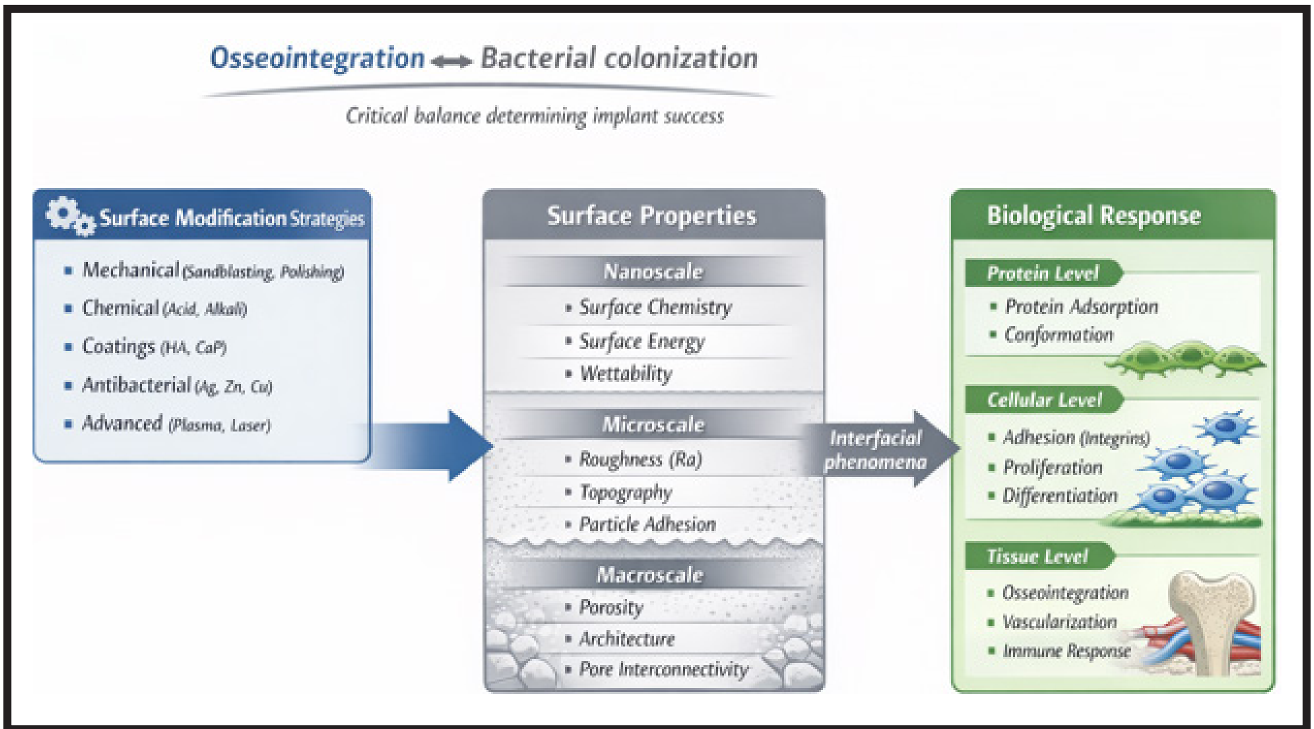

FIG. 1. Multiscale surface biological interaction model for DMLS titanium implants. Surface modification strategies influence physicochemical properties across nano-, micro-, and macro-scales, which regulate protein adsorption, cellular response, and tissue integration. The balance between osseointegration and bacterial colonization determines the overall implant performance.

stability remains a major limitation [33]. Antibacterial modifications offer effective infection control but introduce concerns related to cytotoxicity and controlled ion release. Importantly, the effectiveness of each approach is strongly dependent on the specific characteristics of DMLS surfaces, including inherent roughness and interconnected porosity [20,22, 37-39]. This highlights that strategies effective for conventionally manufactured titanium may not yield comparable results when applied to additively manufactured structures. Overall, current evidence indicates that multifunctional and hybrid surface engineering approaches, integrating bioactive and antibacterial functionalities, offer the most promising pathway toward clinically optimized DMLS implants [43,48]. This comparison underscores that future implant design should not rely on single-parameter optimization but rather on integrated surface engineering strategies that simultaneously address biological and antibacterial requirements. From a design perspective, optimal DMLS implant surfaces should combine moderate microscale roughness ($Ra \approx 1\text{-}3$ µm) with nanoscale features that promote osteogenic differentiation, while simultaneously incorporating antibacterial functionality through controlled surface chemistry or coatings [24,39,56]. Achieving such multifunctionality requires integrated surface engineering rather than isolated modification techniques

Challenges

Despite significant progress in the development of surface modification strategies for DMLS titanium implants, several challenges remain that limit their widespread clinical application [48]. One of the primary issues is the lack of standardization in DMLS processing parameters. Variations in laser power, scan speed, and powder characteristics can lead to significant differences in microstructure, surface morphology, and porosity. As a result, reproducibility of both the base material and subsequent surface modifications remains a critical concern [20-23].

Another important challenge is the long-term stability and durability of surface coatings. While bioactive and antibacterial coatings can significantly enhance initial biological performance, their adhesion strength and resistance to mechanical loading and degradation over time are often insufficient. Delamination or degradation of coatings may compromise implant performance and lead to adverse biological responses [33,37].

The complexity of DMLS-generated porous structures also presents difficulties in achieving uniform surface modification. Coating deposition and chemical treatments may not penetrate effectively into internal pores, resulting in heterogeneous surface properties. This issue is particularly relevant for implants designed with interconnected porosity for enhanced bone ingrowth [8,21-23]. Balancing antibacterial activity with cytocompatibility remains a major challenge. While metallic ions such as silver or copper provide effective antimicrobial properties, their potential cytotoxicity necessitates careful control of concentration and release kinetics [37-39].

Finally, the translation of advanced surface modification techniques from laboratory-scale studies to clinical applications is hindered by economic and technological constraints. Scalability, cost-effectiveness, and regulatory approval processes must be addressed before these solutions can be widely implemented in medical practice [46,59]. Another emerging challenge is the lack of standardized evaluation protocols for biological performance. Variability in *in vitro* testing conditions, including cell type, culture duration, and assessment methods, makes direct comparison between studies difficult. Moreover, discrepancies between *in vitro* and *in vivo* results further complicate the translation of experimental findings into clinical practice [53,59].

Additionally, regulatory considerations pose a significant barrier to clinical implementation. Surface-modified implants, particularly those incorporating bioactive or antibacterial agents, must meet stringent safety and efficacy requirements. This is especially relevant for multifunctional coatings with controlled ion release, where long-term biological effects remain insufficiently characterized [39,49,59]. A critical but often overlooked aspect is the translational gap between laboratory-scale surface modifications and clinically approved implant systems. Many advanced surface engineering strategies remain at a low technology readiness level (TRL), with limited validation in large animal models or human clinical studies [46,60].

Furthermore, regulatory approval of modified implant surfaces requires comprehensive evaluation of long-term safety, including potential ion release, coating degradation, and systemic effects. This significantly increases the complexity and cost of development, limiting the industrial adoption of advanced multifunctional coatings [46,59]. Bridging this gap will require interdisciplinary collaboration between materials scientists, clinicians, and regulatory bodies, as well as the development of standardized testing protocols that better reflect clinical conditions.

Future directions

Surface modification strategies for DMLS titanium implants have demonstrated significant potential in improving biological performance; however, their clinical translation remains limited. Currently, only a small subset of surface treatments, such as sandblasting and acid etching, are routinely used in commercial implants, primarily due to their simplicity, reliability, and regulatory acceptance [24,29,59]. Advanced approaches such as multifunctional coatings and ion-releasing systems remain largely at the experimental stage. Their clinical implementation is constrained by challenges related to long-term stability, reproducibility, and regulatory approval [48,59]. From a clinical standpoint, the most promising strategies are those that provide a balance between enhanced osseointegration, antibacterial functionality, and mechanical reliability, while maintaining manufacturability and cost-effectiveness [33,37,48].

Future research should focus on the development of bioinstructive and multifunctional implant surfaces capable of actively directing biological responses. Particular attention should be given to stimuli-responsive coatings enabling controlled release of bioactive and antibacterial agents, smart surfaces capable of adapting to local biological environments, and hierarchical designs integrating nano-, micro-, and macro-scale features [61,62].

The integration of machine learning and data-driven approaches represents a promising direction for optimizing both DMLS processing parameters and surface modification strategies. Such approaches may enable predictive design of implant surfaces with tailored biological performance [63,64]. Furthermore, there is a critical need for long-term *in vivo* studies and clinical validation, particularly for multifunctional coatings and ion-releasing systems. Bridging the gap between laboratory-scale research and clinical application will require interdisciplinary collaboration and standardized evaluation protocols [60,61]. Ultimately, the next generation of DMLS titanium implants is expected to transition from passive biomaterials to active, biofunctional systems capable of simultaneously promoting osteogenesis and preventing infection.

Conclusions

Surface engineering plays a central role in enhancing the biological performance of DMLS titanium implants. While individual modification strategies offer specific advantages, none fully address the complex clinical requirements. Current evidence indicates that multifunctional and hybrid approaches provide the most promising pathway for optimizing implant performance. However, challenges related to reproducibility, coating stability, and clinical translation remain significant. Future developments should focus on integrated, multiscale surface design combined with clinically relevant validation, enabling the transition toward advanced biofunctional implant systems. This review is limited by the variability of experimental conditions reported in the literature, which makes direct comparison between studies challenging. Additionally, the lack of standardized testing protocols may affect the generalizability of the presented findings.

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