

Review

Research Progress on Biomedical Functional Coatings for Titanium Alloys: A Review

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Highlights

- Nine surface treatments are compared for preparing biomedical coatings on titanium alloys.
- The properties of the coating are affected by its composition, structure, and ion release.
- Ca/P rich coatings promote osteogenesis; Mn/Zn/Cu/Ag rich coatings suppress bacteria.
- Surface treatments should match the intended implant function.
- Ion-release control balances antibacterial effects and cytocompatibility.
- Standardized corrosion, cell, and biofilm tests enable comparisons.

Abstract

Titanium alloys are widely used for implants, yet corrosion, bacterial colonization and incomplete osseointegration remain important causes of interfacial failure. This review critically analyzes major biomedical functional coating fabrication techniques employed to enhance the surface properties of titanium alloys, including micro-arc oxidation, anodic oxidation, magnetron sputtering, electrochemical deposition, electrophoretic deposition, plasma spraying, physical vapor deposition, plasma immersion ion implantation, laser surface treatment, and hybrid (composite) approaches. For each method, key operational principles, structural and functional characteristics, performance advantages and limitations, and representative application domains are critically analyzed. Across these routes, biological performance depends on coating continuity, pore or nanotube geometry, interfacial bonding, phase composition and ion release. Calcium- and phosphorus-rich oxides and hydroxyapatite deposits generally promote cell adhesion, proliferation, alkaline phosphatase activity, mineralization and osteogenic differentiation. Dense oxide, nitride, tantalum and carbon-based films strengthen corrosion barriers, whereas Mn, Zn, Cu and Ag containing surfaces can inhibit bacterial adhesion and biofilm formation. Excessive ion release, however, may compromise cytocompatibility. Reported outcomes also vary with test medium, exposure time, bacterial strain and cell model. Standardized quantitative endpoints and longer-term corrosion, biofilm and osseointegration studies are required to guide clinically reliable multifunctional coatings.

Keywords: titanium alloys; coating; surface modification; corrosion resistance; biocompatibility



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1. Introduction

Titanium alloys remain widely used in orthopedic, dental and spinal implants because they combine high specific strength, toughness, corrosion resistance and a comparatively

low elastic modulus [1]. Beta-type alloys can reduce stiffness further, while Ti-Ta compositions avoid several alloying elements associated with biological concern [2,3]. Additive manufacturing has also expanded the design space by enabling patient-specific components and porous architectures that accommodate bone ingrowth [4,5]. These advantages belong mainly to the bulk material. At the tissue-facing surface, corrosion, wear, bacterial attachment and incomplete osseointegration still govern long-term reliability [6,7].

The implant interface is not chemically static. Proteins, inflammatory species, cells and microorganisms continuously alter the local environment. Physiological and pathological media can change the passive state of Ti-6Al-4V extra-low-interstitial (ELI) alloy [8], and inflammation-mimicking solutions produce similar changes on Ti-29Nb-21Zr [9]. At smaller length scales, surface structure affects protein adsorption, cell attachment and differentiation [10]. Wettability, friction and biocompatibility must therefore be considered together rather than treated as independent surface properties [11].

Oxidation-based treatments create one major family of biomedical surfaces. Micro-arc oxidation (MAO), also known as plasma electrolytic oxidation (PEO), forms porous ceramic layers through localized discharges, with electrical parameters and electrolyte chemistry controlling growth, phase composition and elemental incorporation [12–14]. Anodic oxidation (AO) instead favors nanotubes, nanopores and thin TiO₂ films [15]. Laser nitriding and cladding provide metallurgical bonding [16,17], whereas hydrothermal treatment can increase apatite-forming activity [18]. MAO surfaces have also been linked to osteogenic differentiation [19]. These methods act at different structural scales, so roughness alone cannot explain their biological response.

Physical vapor deposition (PVD) and near-surface modification offer a different balance between dimensional control and function. Carbon implantation and diamond-like carbon (DLC) films reduce friction and restrict ion release [20,21], while Cu containing oxide multilayers add antibacterial activity to barrier protection [22]. Thermal spraying produces thicker hydroxyapatite (HAp)-based bone-contact layers [23], and anodic electrolytes provide direct control over oxide chemistry [24]. Plasma immersion ion implantation (PIII) has been examined for blood-contacting surfaces [25], while corrosion modeling helps screen HAp coating conditions [26]. Particle-based electrophoretic deposition (EPD) supports HAp composites [27], and laser cladding now extends well beyond surface hardening to bioactive-layer fabrication [28,29]. Related work on low-modulus beta alloys, TiO₂ nanotubes and Ti-13Nb-13Zr nanotubes shows that substrate chemistry and nanostructure cannot be separated [30–32]. Cu/Zn modified TiN, graded Ta₂O₅/Ti, metallic-glass films and optimized HAp deposition further illustrate the range of available interfacial designs [33–36].

A useful comparison requires endpoints that match the claimed function. Ca and P rich MAO layers need both compositional and structural evidence [37], while Ag electrodeposition must be read in the context of deposition conditions and electrochemical behavior [38]. For suspension plasma-sprayed HAp, thickness, crystallinity and in vitro reactivity are especially informative [39]. Interlayer design controls corrosion in Ta-based multilayers [40], and the performance of TiN or calcium phosphate (CaP) enriched nanotubes cannot be inferred from a single morphological value [41,42]. Similar dependencies appear in Ti-based ceramic films, HAp deposited on additively manufactured titanium and HAp/TiO₂ coatings on laser-powder-bed-fused substrates [43–45].

Recent studies increasingly combine barrier, antibacterial and osteogenic functions within one surface. Green-synthesized HAp deposited by EPD has been used to couple mineralization with antibacterial action [46]. Zeolitic imidazolate framework-8 (ZIF-8) and graphene additions improve different aspects of barrier performance [47,48], while microplasma and thermal spraying tune porosity, composite architecture and bonding [49,50].

Sr containing MAO layers, thermally treated MAO surfaces, Fe-bearing titanium alloys and MAO coatings on additively manufactured substrates show why processing history and test conditions matter [51–54]. This review organizes the evidence by fabrication route and compares corrosion, antibacterial, cellular and osteogenic outcomes using representative quantitative evidence.

Although recent reviews have addressed titanium surface modification broadly or focused on individual routes, such as MAO, they do not consistently compare the principal coating technologies using the same corrosion, antibacterial, cellular, and osteogenic endpoints [6,7]. The present review contributes a technology-classified synthesis that links fabrication mechanism, coating architecture, interfacial limitations, and study-specific functional outcomes [12,14]. It further provides a comparative evidence map in Table 1 and a cross-technology discussion in Section 3.10 to clarify where direct comparison is valid and where heterogeneous substrates, media, and test protocols preclude ranking. This scope complements, rather than duplicates, route-specific reviews by identifying the design trade-offs relevant to multifunctional titanium-alloy implant surfaces.

2. Literature Search and Analysis Methods

The search and screening process was organized as a structured narrative review guided by PRISMA reporting principles. Figure 1 summarizes source identification and screening. Because the evidence spans heterogeneous coating technologies and outcome families, no quantitative meta-analysis was planned.

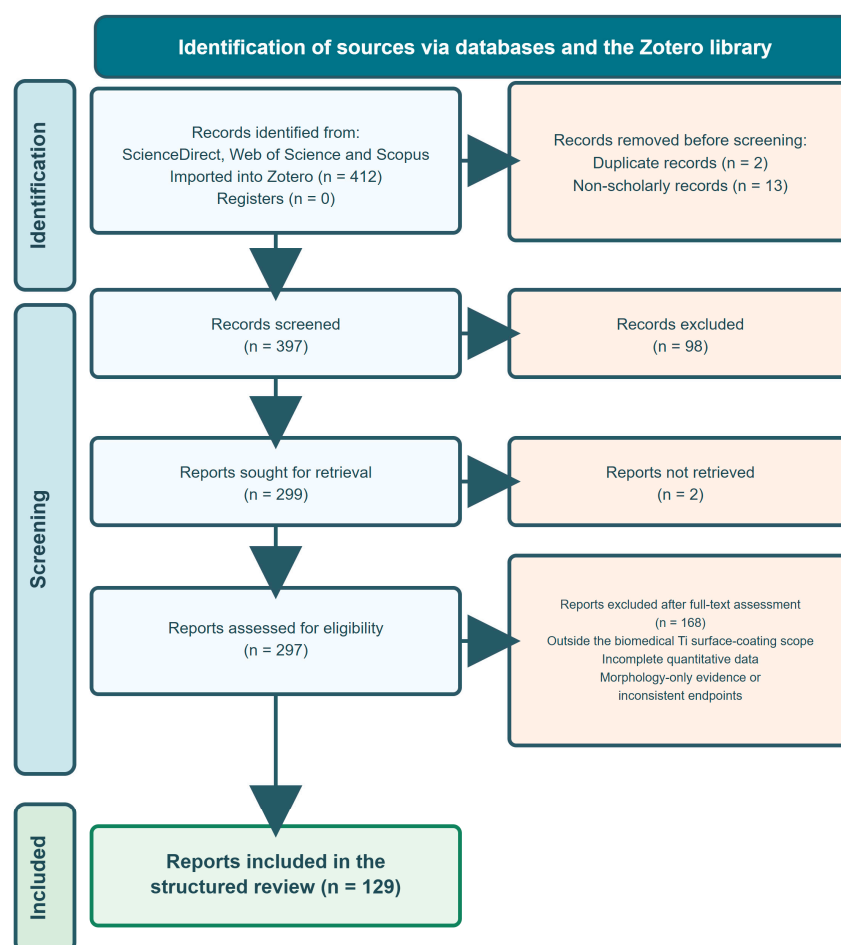


Figure 1. PRISMA-style flow diagram of source identification and screening for the structured narrative review.

Literature search strategy and eligibility criteria. ScienceDirect, the Web of Science Core Collection, and Scopus were searched from 1 January 2015 to 1 July 2026. Earlier studies were retained only when they provided foundational mechanistic information or quantitative evidence not represented in the main search period. The search combined three concept groups: the titanium substrate, the coating or surface-modification route, and the biomedical function or outcome. The Web of Science search was TS = ((titanium OR “titanium alloy*” OR Ti-6Al-4V OR Ti6Al4V) AND (coating* OR “surface modification” OR “micro-arc oxidation” OR “plasma electrolytic oxidation” OR anodiz* OR “magnetron sputter*” OR “electrochemical deposition” OR “electrophoretic deposition” OR “plasma spray*” OR “physical vapor deposition” OR “plasma immersion ion implantation” OR “laser surface treatment”) AND (biomedical OR implant* OR corrosion OR antibacterial OR antibiofilm OR biocompatib* OR cytocompatib* OR osteogen* OR osseointegrat*)). The Scopus search used the same terms within TITLE-ABS-KEY. The ScienceDirect search used (“titanium alloy” OR “Ti-6Al-4V”) AND (coating OR “surface modification”) AND (biomedical OR implant OR corrosion OR antibacterial OR biocompatibility OR osteogenesis), with individual coating-route terms added in successive searches where the platform limited query length.

Primary studies were eligible when they (i) investigated a coating or intentional surface-modification treatment on titanium or a titanium alloy for a biomedical or implant-related application; (ii) reported sufficient information on the substrate, treatment route, and coating or modified-layer composition; and (iii) reported at least one functional endpoint related to corrosion, wear or tribocorrosion, antibacterial activity, cytocompatibility, osteogenic response, or in vivo tissue response. Review articles and methodological or general background sources were retained only to explain mechanisms, terminology, or field context and were not treated as primary experimental evidence. Records were excluded when they concerned non-titanium substrates, non-biomedical applications, bulk-alloy modification without a surface treatment, morphology or composition without a relevant functional endpoint, or were abstracts or conference summaries without a full paper, duplicate reports, non-English full texts, or unavailable full texts.

Screening procedure. Records were imported into Zotero and duplicates and non-scholarly items were removed before screening. Titles and abstracts were screened for scope, followed by full-text assessment against the eligibility criteria. For retained sources, the extracted fields included substrate, coating route and composition, processing conditions, comparator, test medium, bacterial strain or cell model, exposure time, quantitative endpoint, and principal conclusion. The original review process was collaborative but was not prospectively recorded as a dual-independent screening exercise. Consequently, a reviewer-agreement statistic was not calculated, and no retrospective kappa value is presented. This reporting limitation is one reason the article is described as a structured narrative review with PRISMA-informed screening rather than a formal systematic review.

Data extraction and evidence interpretation. The extraction record included fabrication method, substrate, coating or modified-layer composition, processing conditions, performance endpoint, and study conclusion. Electrochemical variables comprised corrosion potential (E_{corr}), corrosion current density (I_{corr}), corrosion rate, polarization resistance (R_p), and electrochemical impedance spectroscopy (EIS). Antibacterial records included strain, inoculation conditions, colony-forming units (CFU), antibacterial rate, and adhesion. Cellular evidence included Cell Counting Kit-8 (CCK-8), 3-(4,5-dimethylthiazol-2-yl)-2,5-diphenyltetrazolium bromide (MTT), optical density (OD), morphology, and fluorescence staining. Alkaline phosphatase (ALP) activity, mineralization, Alizarin Red staining, and related markers were used for osteogenic assessment.

Study appraisal and review classification. The evidence was grouped by fabrication route, including micro-arc oxidation, anodic oxidation, magnetron sputtering, electrochemical deposition, electrophoretic deposition, plasma spraying, physical vapor deposition, plasma immersion ion implantation, laser processing, and combined treatments. Owing to substantial heterogeneity in substrates, coating architectures, experimental designs, comparators, media, exposure times, bacterial strains, cell models, and outcome definitions, this article is presented as a structured narrative review with PRISMA-informed identification and screening, not as a formal systematic review or meta-analysis. No formal risk-of-bias instrument was applied. Evidence was interpreted descriptively according to the transparency of processing conditions, presence and suitability of a comparator, completeness of test-condition reporting, use of direct functional endpoints, and consistency between the reported data and the authors' conclusions. Accordingly, quantitative values are treated as study-specific evidence and are not pooled or ranked across incompatible test systems.

The final reference corpus comprised 129 reports or sources, including primary research articles used for study-specific experimental outcomes and secondary or background articles used only for mechanisms, terminology, and field context. Thus, the terminal number in Figure 1 is a count of sources retained in the structured narrative review and corresponds to the 129 cited references. It must not be interpreted as a count of 129 primary studies included in a formal systematic evidence synthesis.

The evidence was interpreted through four related outcome groups. Electrochemical measurements describe whether the modified surface remains an effective barrier in body fluids. Antibacterial tests address adhesion, growth and biofilm formation. Proliferation and cytocompatibility measurements capture cell survival and expansion, whereas ALP activity, mineralization and related markers provide evidence of osteogenic differentiation. Each Section therefore links the fabrication route to the endpoint actually measured.

3. Fabrication Technologies for Biomedical Functional Coatings on Titanium Alloys

Biomedical surface engineering has moved from single-purpose corrosion barriers toward interfaces that combine electrochemical stability, antibacterial action and tissue integration. The main routes include in situ oxidation, thin-film growth, ECD or EPD, thermal spraying, laser processing and combined mechanical or chemical treatments [6]. Because these routes produce different thicknesses, pore structures, bonding states and release profiles [8], their performance must be judged against both fabrication mechanism and experimental endpoint [55]. Representative antibacterial and electrochemical data from the principal treatment groups are compared in Table 1. Table 1 is intended to map representative outcomes, not to rank technologies. Values obtained with different substrates, media, exposure times, bacterial strains, cell models, and analytical protocols should not be compared as if they were measured under identical conditions.

Table 1. Antibacterial and electrochemical outcomes of representative surface treatments for titanium alloys.

Surface Treatment	Matrix	Reported Antibacterial Outcome	Corrosion Outcome: I_{corr} ($\text{A}\cdot\text{cm}^{-2}$)			Ref.
			Test Medium	Matrix/Control	After Treatment	
MAO	CP-Ti	70.8% (<i>S. aureus</i>)	0.9 wt% NaCl	1.63×10^{-7}	2.05×10^{-8}	[56]
	Ti-6Al-4V	—	0.9 wt% NaCl	7.28×10^{-8} (MAO)	1.91×10^{-8}	[57]

Table 1. Cont.

Surface Treatment	Matrix	Reported Antibacterial Outcome	Corrosion Outcome: I_{corr} ($\text{A} \cdot \text{cm}^{-2}$)			Ref.
			Test Medium	Matrix/Control	After Treatment	
AO	Ti-29Nb-13Ta-7.1Zr	—	PBS, pH 7.4	7.474×10^{-6}	1.40×10^{-7}	[58]
	NiTi	>95% at day 1	PBS	1.66×10^{-7}	6.90×10^{-9}	[59]
MS	Ti-6Al-4V	—	SBF	6.79×10^{-9}	2.92×10^{-9}	[60]
	Ti-6Al-4V	—	0.5 M NaCl + $2 \text{ g} \cdot \text{L}^{-1}$ KF, pH 2	2.6×10^{-8} (TiO ₂ (a))	6.5×10^{-8} (TiO ₂ (a)/graphene)	[61]
	Ti-6Al-4V	30%/40% damaged <i>S. aureus</i> , 2/4 h	—	—	—	[62]
ECD	Ti-6Al-4V ELI	—	HBSS	1.62×10^{-6} (0.01 M)	4.91×10^{-7}	[63]
	Porous Ti	—	PBS	1.83×10^{-7}	9.10×10^{-8}	[64]
EPD	Ti-6Al-4V	<i>E. coli</i> : 15×10^8 CFU·mL ⁻¹ ; <i>S. aureus</i> : 2.2×10^8 CFU·mL ⁻¹	—	—	—	[65]
PIII	Ti-6Al-4V	25.58% (<i>E. coli</i>); 38.20% (<i>B. subtilis</i>)	SBF	9.07×10^{-6}	3.21×10^{-8}	[66]
	Ti-6Al-4V	Reduced <i>S. salivarius</i> adhesion	Artificial saliva/SBF	2.10×10^{-7} / 1.30×10^{-7}	3.00×10^{-8} / 2.00×10^{-8}	[67]
	NiTi	—	Hanks' solution	3.20×10^{-9}	2.50×10^{-9}	[68]
Laser processing	Ti-6Al-4V	—	3.5% NaCl/Hanks/DMEM	1.37×10^{-7} / 5.72×10^{-8} / 3.44×10^{-8}	1.15×10^{-7} / 5.71×10^{-8} / 2.61×10^{-8}	[69]
	Ti-6Al-4V	—	SBF	2.459×10^{-5}	1.051×10^{-7}	[70]
Other: ESD	Ti-6Al-4V	—	3.5% NaCl	2.64×10^{-6}	2.60×10^{-7}	[71]

Note: anodic oxidation (AO); commercially pure titanium (CP-Ti); Dulbecco's modified Eagle medium (DMEM); electrochemical deposition (ECD); electrophoretic deposition (EPD); electrospark deposition (ESD); Hanks' balanced salt solution (HBSS); magnetron sputtering (MS); micro-arc oxidation (MAO); phosphate-buffered saline (PBS); plasma immersion ion implantation (PIII); simulated body fluid (SBF). I_{corr} values were converted to $\text{A} \cdot \text{cm}^{-2}$. Values are reported as study-specific outcomes. Direct comparisons across rows are not valid because substrates, coating chemistry, control definitions, media, reference electrodes, exposure times, temperatures, bacterial strains, inoculum conditions, cell models, and testing protocols differ. Matrix/control denotes the reported baseline, which may be a comparator coating when an uncoated substrate value was unavailable. A dash indicates that the endpoint was not quantitatively reported.

3.1. Micro-Arc Oxidation

MAO uses high-voltage microdischarges to convert the titanium-alloy surface into an adherent oxide-ceramic layer [56,72]. The layer usually contains a compact inner barrier and a more porous outer region [73]. Voltage, frequency, duty cycle, treatment time and electrolyte composition govern thickness, porosity, phase content and defect density. Calcium, phosphorus, manganese and zinc can enter the growing oxide and alter corrosion resistance, apatite nucleation or ion-mediated biological activity [74,75]. The typical process and bilayer structure are shown in Figure 2.

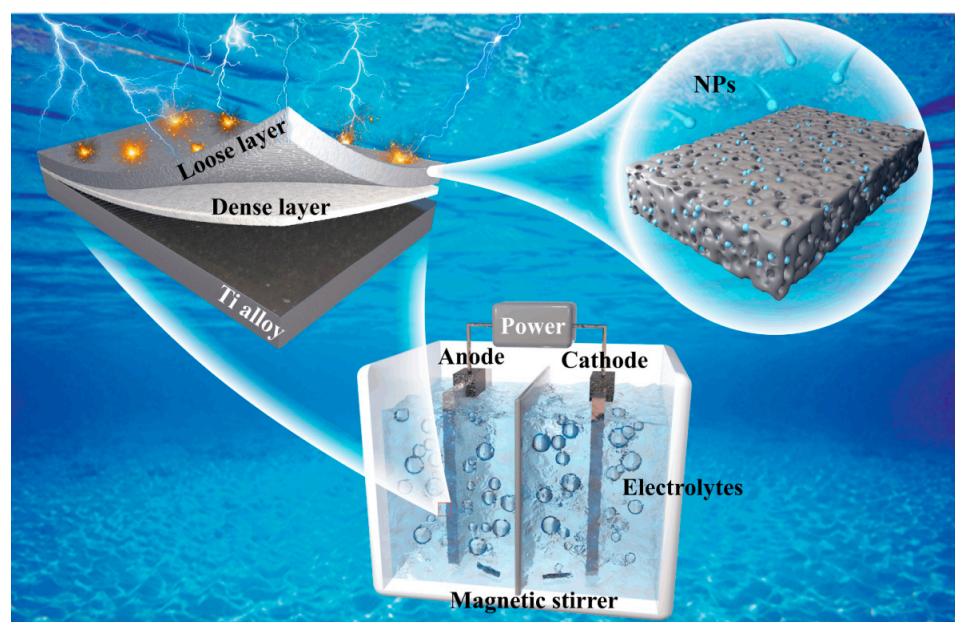


Figure 2. Schematic illustration of a typical MAO process and bilayer coating architecture. Adapted from Ref. [12] with permission.

Calcium and phosphorus containing coatings are primarily designed to improve mineralization at the cell-material interface. On Ti65Zr, the MAO surface examined by Nahum et al. [74] reached a Ca/P ratio of 1.82 after 14 days of immersion. In another system, a 300 V MAO treatment reduced the contact angle to 58.8° and increased bovine serum albumin adsorption to about 2.5-fold at 1 h and 2.1-fold at 24 h [75]. Long-term stability in artificial saliva was also reported for powder-metallurgy Ti-6Al-4V [76]. Across related Ti/Nb, Zn containing and HAp forming systems, electrolyte chemistry determined both oxide composition and crystalline phase development [77–79].

A one-step MAO treatment in the study by Zhang et al. [56] produced a TiO₂ rich inner layer beneath an Mn₃O₄ containing outer region. Compared with the Ti substrate, the Ti-Mn-EDTA coating exhibited a more positive E_{corr} , increasing from -0.18 to 0.19 V, and a lower I_{corr} , decreasing from 1.63×10^{-7} to 2.05×10^{-8} A·cm⁻². Its antibacterial rate against *S. aureus* reached 70.8%, while the ALP activity of MC3T3-E1 cells was approximately 138.5 U·L⁻¹ after 7 days. These endpoints show that Mn incorporation affected the corrosion barrier, bacterial response and early osteogenic activity within the same coating system. Pretreatment provides another means of controlling discharge defects [80]. Zhou et al. [57] applied MAO after pulsed-laser remelting. At 100 W, coating porosity declined from 21.32% to 10.94%, while thickness increased from 8.14 to 19.49 μm. The lowest I_{corr} , 1.91×10^{-8} A·cm⁻², occurred for the 80 W specimen. Here, the corrosion response followed changes in coating compactness and thickness rather than the addition of a functional element.

3.2. Anodic Oxidation

AO forms TiO₂ nanotubes, nanopores or thin oxide films at a finer scale than most MAO coatings [58]. Applied voltage, treatment time, electrolyte and substrate condition determine tube diameter, length and ordering [81]. Annealing, hydrothermal treatment and ultrathin capping layers can then adjust phase composition, wettability, corrosion stability and loading capacity [82].

On Ti-29Nb-13Ta-7.1Zr, Sarraf et al. [58] produced a nanotube layer about 5 μm thick with an average tube diameter of 90.94 nm. Annealing at 450 °C reduced I_{corr} from 7.474×10^{-6} to 1.40×10^{-7} A·cm⁻² and yielded a protection efficiency of 98.12%. The

friction coefficient decreased from 0.51 to 0.28, and the contact angle changed from 89.9° to 45.6° . Raju et al. [81] similarly found that anodization conditions strongly affected Ti-6Al-4V corrosion in SBF. Nanotube surfaces are often modified after anodization. Diameter-dependent corrosion and cell responses have been reported on NiTi [83], while Zn incorporation adds antibacterial and osteogenic functions [59]. Chitosan–bioactive glass layers provide an organic–inorganic alternative [84]. Self-assembly, heat treatment and residual-stress control offer further routes to regulate the nanotube surface [85–87]. Figure 3 outlines the ion-transport and growth stages underlying NiTi anodization.

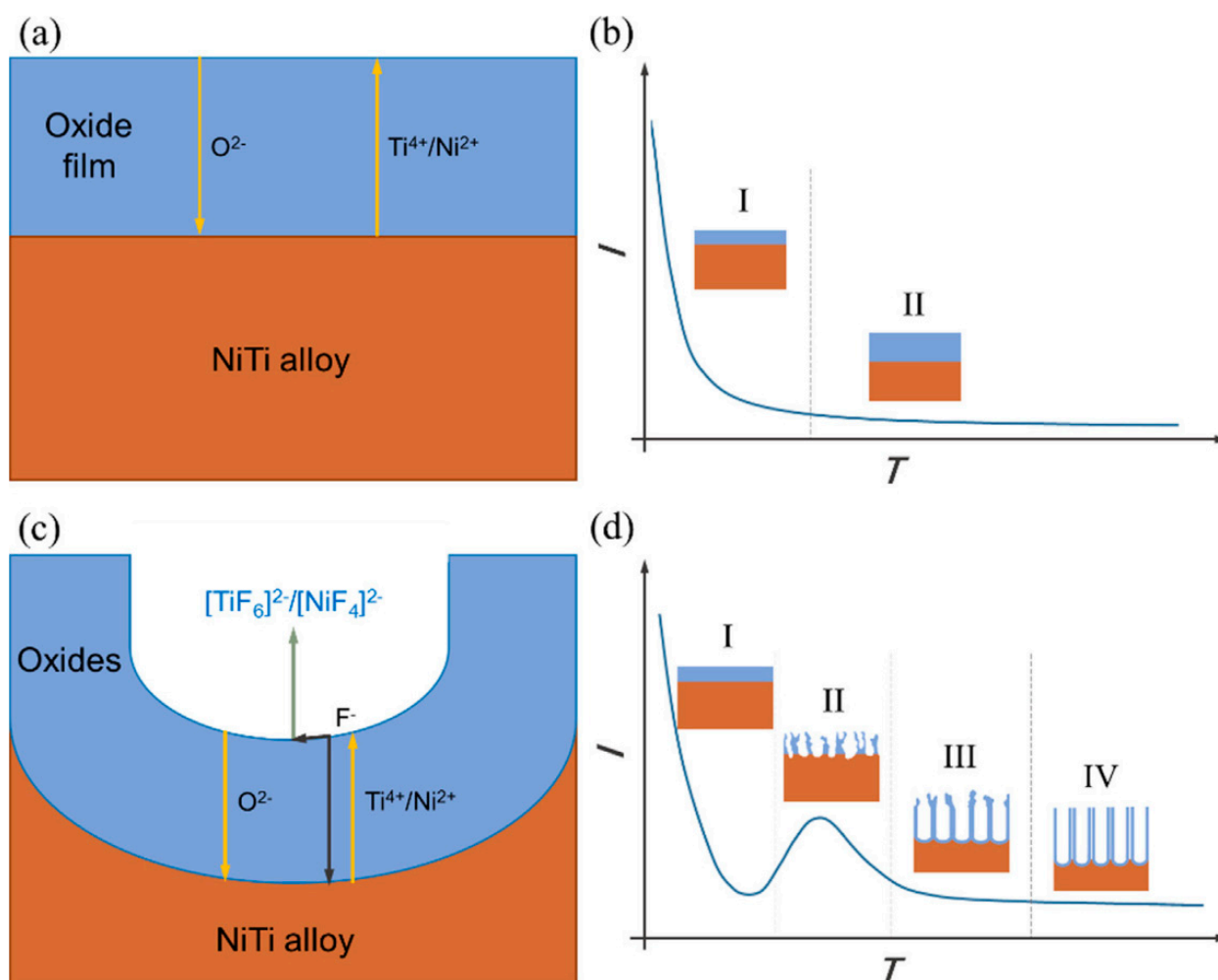


Figure 3. Schematic illustration of electric–field–assisted ion transport and nanotube growth during AO of NiTi (a–d). Adapted from Ref. [85] with permission.

Sepúlveda et al. [82] anodized Ti-6Al-4V at 3.3 and 15 V to obtain nanotubes with internal diameters of about 15 and 50 nm. Ultrathin TiO_2 caps, approximately 0.055 or 0.3 nm thick, reduced exposure to V and F containing species without obscuring the nanotube morphology. MG63 cell density continued to rise over 96 h, and cell elongation increased by about 20%. The response reflects the combined effect of nanoscale geometry and chemical shielding.

3.3. Magnetron Sputtering

MS deposits metals, oxides, nitrides and CaP-related films through plasma-assisted target erosion [88]. Target chemistry, reactive atmosphere, substrate bias, power and interlayer design determine film density, phase composition, residual stress and bonding [89]. Multi-layer and graded structures can interrupt both crack growth and corrosion pathways [60].

The biomedical role of a sputtered film depends on its architecture. Si-HAp layers have been used to adjust Ca/P chemistry and cell adhesion [88]. TiO₂/graphene hybrids, Ti_xNy films and beta-Ti deposited by high-power impulse magnetron sputtering (HiPIMS) target corrosion or mechanical performance through different mechanisms [61,90,91]. Ta-O and Ti-Nb-Zr layers reduce interfacial mismatch [92,93], whereas HAp-Ag-SiO₂-TiN multilayers combine bioactivity with antibacterial action [62].

Reactive sputtering does not give TiO₂ and TiN the same functional profile. In the comparison by Boudjeda et al. [94], TiO₂ showed an I_{corr} of $2.0 \times 10^{-8} \text{ A}\cdot\text{cm}^{-2}$, compared with $8.1 \times 10^{-7} \text{ A}\cdot\text{cm}^{-2}$ for TiN. The TiO₂ surface also retained fewer *S. aureus* and *Escherichia coli* colonies. Film chemistry and defect structure therefore remained decisive even when the deposition route was unchanged.

Ta-based coatings illustrate the value of interlayer design. Ding et al. [89] compared single-, double- and multilayer Ta₂O₅ structures. Ji et al. [60] measured I_{corr} values of 3.54, 2.92 and 3.18 nA·cm^{−2} for Ta, Ti-Zr-Ta and Zr-Ti-Ta coatings, respectively. Each value was below the 6.79 nA·cm^{−2} measured for the substrate. MC3T3-E1 cultures showed no obvious dead cells after 24, 96 or 168 h. Figure 4 presents the sputtering arrangement used for these multilayer films.

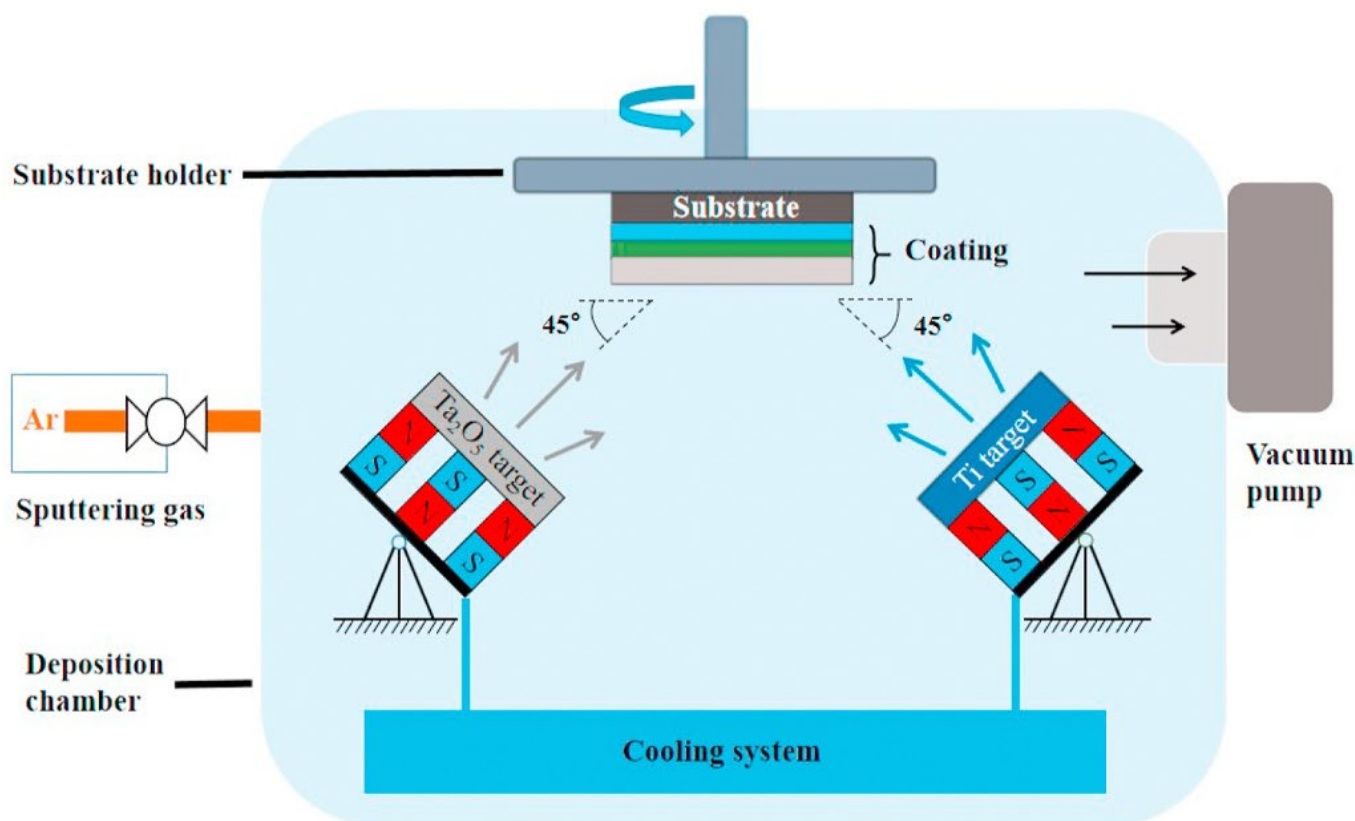


Figure 4. Schematic diagram of the MS system. Adapted from Ref. [89] with permission.

3.4. Electrochemical Deposition and Related Surface Construction

ECD controls nucleation and growth through potential, current, solution concentration, pH and deposition time [63]. It can produce Ag, CaP and HAp based layers at relatively low temperature. Anodic, alkaline or hydrothermal pretreatments are often added to improve coverage, loading, corrosion behavior or mineralization [64]. Valle et al. [63] deposited Ag potentiostatically on Ti-6Al-4V ELI. A 0.1 M AgNO₃ bath applied for 60 s gave reproducible surface morphology. After 300 s, the 0.1 M condition

produced an I_{corr} of $4.91 \times 10^{-7} \text{ A}\cdot\text{cm}^{-2}$, below the $1.62 \times 10^{-6} \text{ A}\cdot\text{cm}^{-2}$ measured for 0.01 M. The corresponding corrosion rates were 0.0058 and $0.0191 \text{ mm}\cdot\text{year}^{-1}$.

A hierarchical route was used by Zhou et al. [64] on porous titanium with 65%–70% porosity and an average pore size near $360 \mu\text{m}$. The surface contained a 60 nm TiO_2 inner layer, an approximately $1 \mu\text{m}$ titanate-gel layer and a $1.2 \mu\text{m}$ electrodeposited HAp layer. Electrodeposition alone lowered corrosion resistance, which was recovered after hydrothermal treatment. The result emphasizes the supporting role of the inner oxide and the final post-treatment.

3.5. Electrophoretic Deposition

EPD moves charged particles through a liquid suspension and deposits them on a conductive surface. HAp, Si_3N_4 , SiO_2 , ZnO, graphene derivatives and proteins can therefore be applied with limited thermal exposure [95]. Suspension stability, particle charge, voltage, time and substrate pretreatment control coating thickness, uniformity, porosity and adhesion [65]. The method is particularly useful on complex geometries and anodized nanotube surfaces [96]. EPD supports markedly different material functions. SiO_2/ZnO composites have been investigated as corrosion and scratch-resistant barriers [96], while 4–5 μm zein layers provide flexible organic carriers [97]. Ultrasound-assisted deposition improves the dispersion of HAp/graphene oxide (GO) particles and their subsequent biological response [98]. Figure 5 shows the sequence of anodization, ultrasound-assisted EPD and SBF exposure used for the HAp-GO system.

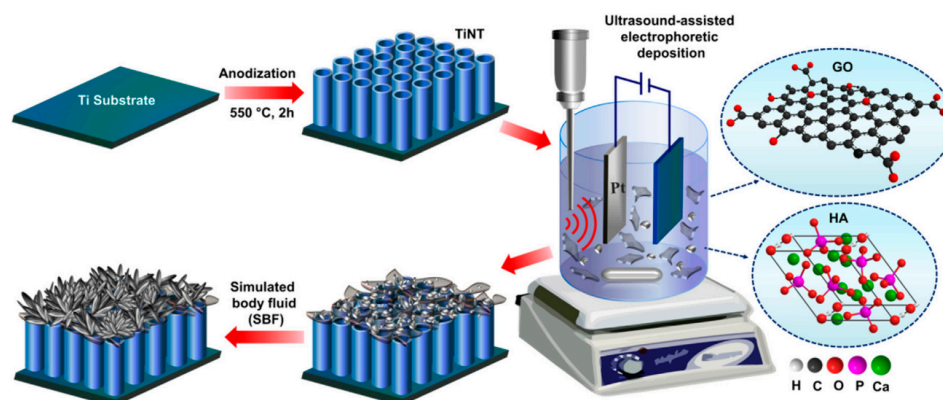


Figure 5. Schematic illustration of anodization, ultrasound-assisted EPD of HAp–GO, and subsequent exposure to SBF. Adapted from Ref. [98] with permission.

Heydarian et al. [65] formed TiO_2 nanotubes on Ti-6Al-4V before loading Si_3N_4 nanoparticles by EPD. After 4 h, the nanotube layer was about $2.6 \mu\text{m}$ thick and contained tubes 80–110 nm in diameter. The reduced elastic modulus fell from approximately 120 to 21 GPa. MG63 viability ranged from 77.66% to 99.79%, and the *S. aureus* count decreased by about $2.2 \times 10^8 \text{ CFU}\cdot\text{mL}^{-1}$. Following SBF immersion, the Ca concentration in solution declined from roughly 50 to $3 \text{ mg}\cdot\text{dL}^{-1}$.

Cu content controlled the balance between adhesion, antibacterial action and cytocompatibility in the HAp–Cu coatings studied by Hadidi et al. [95]. Bond strength rose from 7.3 MPa for pure HAp to 8.4, 9.2 and 10.1 MPa at 1, 3 and 5 wt% Cu. For the 3 wt% coating, minimum inhibitory concentrations were $1400 \mu\text{g}\cdot\text{mL}^{-1}$ against *S. aureus* and $700 \mu\text{g}\cdot\text{mL}^{-1}$ against *E. coli*. This intermediate composition retained a favorable MG63 response while providing measurable antibacterial activity.

3.6. Plasma-Sprayed and Thick Ceramic/Metal Bioactive Coatings

PS rapidly consolidates molten or softened powders into comparatively thick HAp, ceramic-composite or porous-metal coatings. Crystallinity, porosity, residual stress and bond strength depend on feed rate, stand-off distance, particle heating and post-treatment. These variables also affect dissolution and reprecipitation after exposure to body fluids [99–101].

Vacuum plasma spraying produced a 350–380 μm graded porous Ta layer in the study by Kuo et al. [101]. Bond strength reached 54.5 ± 2.3 MPa, surface porosity was about 13%, and Ra was approximately 22.2 μm . Porosity increased from roughly 0.6% near the substrate to 7.6% in the outer layer, while elastic modulus decreased from 148 ± 5 to 123 ± 4 GPa. Alkali-heat treatment accelerated bone-like apatite formation and improved early MG63 attachment.

HAp spray coatings are commonly adjusted through secondary phases, crystallinity and pore-forming treatments. Al_2O_3 and TiO_2 additions have been used to improve residual-stress control and bonding [99,100]. Other studies link powder source and pore development to mineralization and cell response [102–104]. In particular, vapor-induced porosity created a surface favorable for cell adhesion and proliferation. Figure 6 compares conventional atmospheric plasma-sprayed HAp with porous HAp at 100 and 20 μm scales.

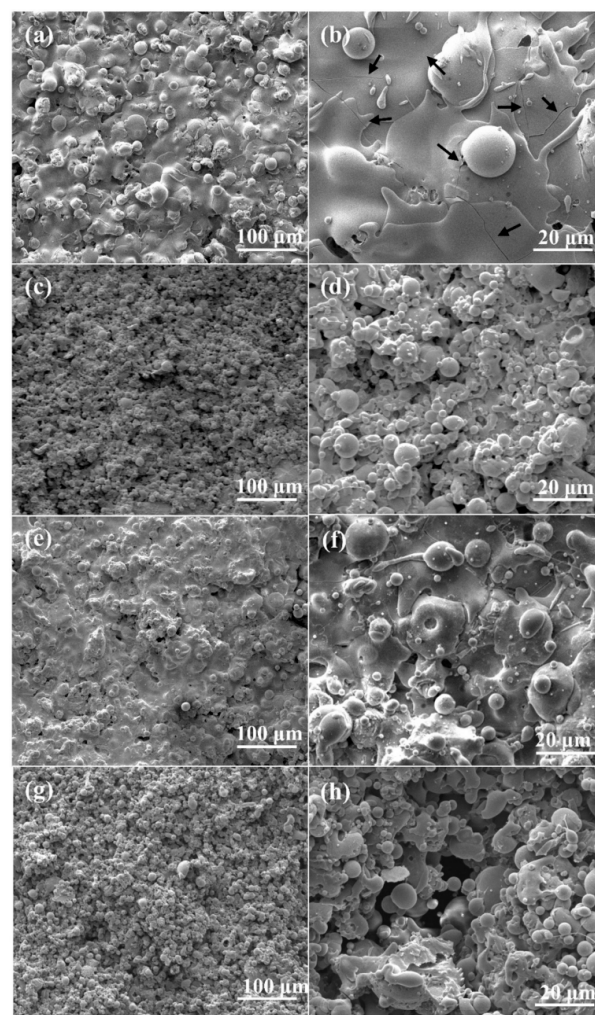


Figure 6. SEM images of (a,b) HAp-APS/ATi, (c,d) HAp-ATi, (e,f) HAp-BTi, and (g,h) HAp-CTi; arrows indicate cracks. Reproduced from Ref. [104] with permission.

Microplasma spraying offers direct control over coating structure through current, gas flow, spraying distance and feed conditions. Alontseva et al. [105] prepared dense CP-Ti, porous CP-Ti and HAp layers on Ti-6Al-4V. Reported thicknesses were 250 ± 30 and 150 ± 30 μm for the principal coatings. Because the Part I study did not include final cellular or corrosion outcomes, it is used here only to describe the process–structure relationship.

3.7. Physical Vapor Deposition, Plasma Immersion Ion Implantation, Diamond-like Carbon and Near-Surface Modification

PVD creates metallic, oxide or nitride cover layers, whereas PIII modifies composition beneath the original surface. DLC mainly acts as a low-friction carbon barrier. Energy input, film thickness, interfacial transition and residual stress govern hardness, adhesion, metal-ion release and tribocorrosion. Because penetration depth and failure mode differ, these treatments should not be treated as a single coating class [66,67,106]. Cover films, implanted layers and carbon coatings address different failure modes. MoN and PVD multilayers reduce tribocorrosion and corrosion current [106,107]. Graded TiN improves adhesion and wear resistance [108], while Ti implantation limits Ni exposure from NiTi [109]. Zr implantation has been assessed through osteogenic-gene expression [110]. DLC deposited by PIII-based routes can reduce the friction coefficient in PBS to approximately 0.06–0.10 [111].

3.7.1. Physical Vapor Deposition

Hussein et al. [66] deposited an approximately 5 μm nanocrystalline TiN film with a hardness of 38.63 GPa and an elastic modulus of 358 GPa. E_{corr} shifted from -0.4407 to -0.2560 V, and I_{corr} decreased from 9.07×10^{-6} to 3.21×10^{-8} $\text{A}\cdot\text{cm}^{-2}$. Antibacterial inhibition increased by 25.58% for *E. coli* and 38.20% for *Bacillus subtilis*.

3.7.2. Plasma Immersion Ion Implantation

N-PIII produced a broader biological and electrochemical response in the system examined by Huang et al. [67]. In artificial saliva, I_{corr} decreased from 2.10×10^{-7} to 4.00×10^{-8} and 3.00×10^{-8} $\text{A}\cdot\text{cm}^{-2}$. The corresponding SBF values fell from 1.30×10^{-7} to 3.00×10^{-8} and 2.00×10^{-8} $\text{A}\cdot\text{cm}^{-2}$. Human mesenchymal stem-cell proliferation and 14-day mineralization increased by at least 1.3-fold. On oxygen-implanted NiTi, MTT OD reached 0.766 at 48 h and 1.042 at 72 h, compared with 0.644 and 0.870 for the untreated alloy [68].

3.7.3. Diamond-like Carbon and Related Near-Surface Modification

DLC coatings act primarily as low-friction, chemically resistant carbon barriers. Their biomedical value lies in reducing friction, tribocorrosion, and metal-ion release rather than in forming a thick bioactive layer. Adhesion and residual stress remain important limitations, particularly for thicker films or mechanically demanding interfaces. DLC deposited by PIII-based routes has reduced the friction coefficient in PBS to approximately 0.06–0.10 [111].

3.8. Laser Cladding, Laser Deposition and High-Energy-Beam Surface Processing

Laser cladding, directed-energy deposition, texturing and laser shock processing all use localized energy, but they modify the surface in different ways [112]. Cladding and deposition create metallurgically bonded alloy or ceramic layers [113]. Texturing changes topography, while shock processing primarily alters the mechanical state. Power, scanning speed, spot size, overlap and feed rate determine dilution, phase transformation, porosity and cracking [114].

Bioactive laser coatings commonly contain glass, HAp or graded ceramic precursors. Bioactive glass, natural HAp and TiO₂-HAp feedstocks have each been deposited on titanium alloys [115–117]. In the graded TiO₂-HAp coating reported by Behera et al. [117], cytocompatibility after 6 days reached $215.6 \pm 17.2\%$. The corresponding value for the 100% HAp layer was $134.6 \pm 9.5\%$. Figure 7 shows the precursor design and cladding arrangement used in that work.

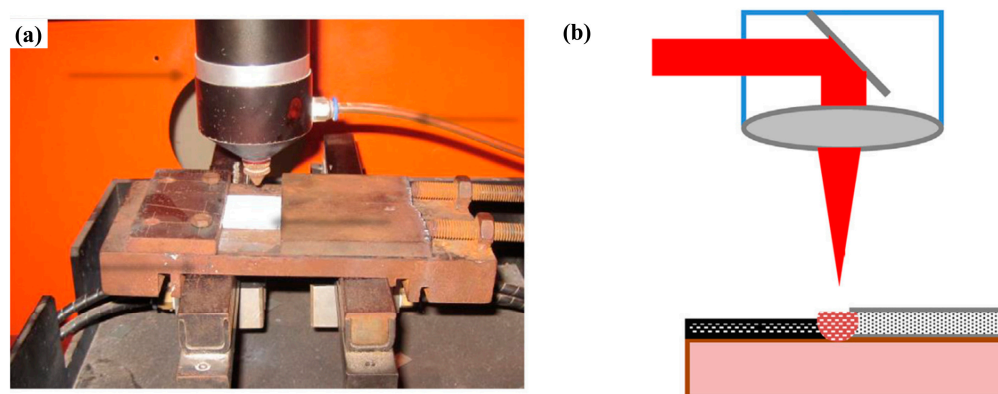


Figure 7. (a) Experimental setup for laser cladding; (b) schematic diagram of laser cladding with HAp and graded TiO₂-HAp precursors. Adapted from Ref. [117] with permission.

The corrosion response of laser-cladded Ti-6Al-4V depended on the test solution. In the bioactive-glass system studied by Meng et al. [69], I_{corr} decreased from 1.37×10^{-7} to $1.15 \times 10^{-7} \text{ A}\cdot\text{cm}^{-2}$ in 3.5 wt% NaCl. In DMEM, it fell from 3.44×10^{-8} to $2.61 \times 10^{-8} \text{ A}\cdot\text{cm}^{-2}$. The change in Hanks' solution was much smaller, showing that one medium cannot represent all service conditions.

Texturing, deposition and shock processing also differ in their immediate targets. Femtosecond texturing followed by hydrothermal treatment creates hierarchical oxide features [112]. Laser texturing and oxidation can strengthen a TiZrTa film interface [113], whereas build orientation affects the surface produced by direct laser deposition [114]. Laser shock processing, by contrast, is used mainly for mechanical strengthening [118].

Xu et al. [70] produced Ti-35Nb-xZr claddings with a thickness of $844.6 \pm 51.2 \mu\text{m}$. At 5 wt% Zr, hardness reached 406 HV0.2 and elastic modulus was approximately 89.67 GPa. I_{corr} values for the Zr containing coatings ranged from 1.051×10^{-7} to $3.063 \times 10^{-7} \text{ A}\cdot\text{cm}^{-2}$. These values were about two orders of magnitude below the $2.459 \times 10^{-5} \text{ A}\cdot\text{cm}^{-2}$ measured for TC4. Ca/P containing deposits also formed during SBF immersion.

3.9. Other Combined Surface-Functionalization and Boundary Methods

Several useful treatments sit outside the main coating categories. Shot peening, etching, electrospark deposition (ESD), biomimetic deposition and friction-stir processing act through plastic deformation, chemical dissolution, transient discharge or near-surface mixing [119]. They produce different modification depths, roughness levels and microstructures [120]. Such routes are valuable when complex geometry, wear strengthening or near-surface compositing is more important than a conventional deposited film [71]. Mechanical and chemical combinations mainly reshape topography and the near-surface microstructure. Shot peening followed by HF/HNO₃ etching improved cell attachment on Ti-6Al-4V [119]. Cold rolling and shot peening produced distinct tribological responses on Ti-13Nb-13Zr [120]. Biomimetic deposition on CP-Ti instead targeted tribocorrosion and cellular behavior [121].

ESD produced CuTi₃, CuTi, Cu₄Ti₃ and Cu₃Ti phases in the coatings studied by Burkov et al. [71]. Microhardness reached 6.7 GPa. Wear rates ranged from 0.67 to

$5.35 \times 10^{-5} \text{ mm}^3 \cdot \text{N}^{-1} \cdot \text{m}^{-1}$, compared with approximately $7.5 \times 10^{-5} \text{ mm}^3 \cdot \text{N}^{-1} \cdot \text{m}^{-1}$ for the substrate. The equiatomic Cu/Ti composition gave the best wear resistance, whereas lower Cu contents provided stronger corrosion resistance in 3.5% NaCl.

Friction-stir processing modifies the substrate itself rather than adding a discrete coating. TiO_2 particles, phase transformation and grain refinement have been introduced into Ti-Nb-Zr and Ti-6Al-4V surfaces by this route [122,123]. Post-treatment of selective-laser-melted Ti-6Al-4V provides a related strategy. Thermal, acid, chemical and thermochemical treatments produced rutile-rich micro- and nanoscale surfaces with improved hardness and scratch resistance [124].

3.10. Cross-Technology Comparison, Evidence Limits and Clinical Translation

Across technologies, the appropriate route depends on the required coating thickness, geometry, interfacial load, and release function [56,72,73]. MAO and AO create in situ oxides with strong substrate continuity and controllable micro- or nanotopography, but pores and discharge defects can compromise long-term barrier stability [58,89,95]. MS and PVD provide dense, compositionally controlled thin films suited to corrosion and wear barriers, although residual stress, pinholes, and interfacial adhesion constrain thick or highly loaded films [66,101,106]. ECD and EPD operate at relatively low temperature and accommodate HAP, polymers, particles, and ion carriers on complex geometries, but coating uniformity and adhesion depend strongly on suspension or electrolyte conditions [67,112,113]. PS and laser cladding generate thicker load-bearing or bioactive layers with mechanical anchoring or metallurgical bonding, but thermal history, residual stress, cracks, and phase transformation require control [114–116]. PIII and DLC are most suitable when near-surface chemistry, metal-ion release, friction, or tribocorrosion is the primary target rather than thick bioactive-layer formation [117].

These distinctions also limit cross-study ranking. Table 1 combines outcomes obtained under different substrates, media, exposure times, bacterial strains, cell models, and analytical protocols. The table should therefore be read as an evidence map of representative endpoints, not as a direct performance league table. A defensible comparison requires matched controls, standardized units, and method-specific context.

For clinical use, functional performance must persist under mechanical and electrochemical loading. Adhesion strength, wear, fatigue, scratch resistance, and tribocorrosion determine whether a coating remains attached and whether cracks, delamination, or particle release expose the substrate [99–101]. The available studies provide bond-strength, hardness, wear, and tribocorrosion data for selected PS, PVD, DLC, ESD, and combined systems, but these endpoints are reported less consistently than corrosion and short-term biological outcomes [106–108]. Long-duration cyclic loading and standardized adhesion testing are therefore priorities rather than established strengths of the current evidence base [71,120,121].

Most evidence synthesized here is laboratory-scale and is dominated by in vitro electrochemical, bacterial, and cell assays. These tests establish mechanism and initial performance, but they do not reproduce long-term bone remodeling, immune response, mixed-species biofilms, cyclic mechanical loading, sterilization, manufacturing variability, or regulatory qualification. Claims of clinical benefit should therefore remain conditional until validated in long-term in vivo studies and, ultimately, controlled clinical investigations.

Antibacterial efficacy and cytocompatibility are not independent design objectives. Increasing Ag, Cu, Zn, or Mn loading can strengthen bacterial inhibition, yet excessive or rapid ion release may impair mammalian-cell viability or osteogenic function [22,33,46]. The most informative studies evaluate release kinetics, antibacterial

activity, cell response, corrosion stability, and coating integrity within the same material system [59,95]. Controlled-release, graded, or multilayer architectures offer a practical route to separate an inner barrier function from an outer bioactive or antibacterial function.

4. Conclusions

Overall, biomedical functional coatings improve titanium-alloy implant interfaces by strengthening corrosion barriers, limiting bacterial colonization and regulating cellular and osteogenic responses. Their effectiveness depends on coating composition, surface architecture, interfacial bonding and ion release rather than on fabrication route alone. Evaluation must therefore link processing and structure to quantitative electrochemical and biological endpoints. Three conclusions can be drawn.

- (1) Oxide, calcium-phosphate, metallic, nitride, carbon and composite-ceramic layers can improve the stability of titanium implant interfaces. Their value cannot be established from thickness, pore size or morphology alone. Corrosion, antibacterial activity, cell proliferation and osteogenic response must be measured directly.
- (2) No fabrication route is universally superior. MAO and EPD are effective for incorporating Ca/P, Mn, Sr, HAp and other functional constituents. MS and PVD provide dense TiO₂, TiN, Ta/Ta₂O₅ and multielement films. PS and laser cladding are better suited to thicker ceramic or composite layers. Ion implantation, DLC and ESD mainly strengthen near-surface barriers, wear resistance and tribocorrosion performance.
- (3) Functional claims should follow the endpoint measured. E_{corr} , I_{corr} , R_p and EIS describe electrochemical behavior; CFU counts, antibacterial rates and strain information support antibacterial claims. CCK-8, MTT and OD measurements address proliferation or cytocompatibility, while ALP activity, mineralization and related tests support osteogenic interpretation. Stronger conclusions require agreement among these endpoints and longer-term models that include corrosion, bacterial challenge and tissue-relevant conditions.

Taken together, clinically useful coatings will require complementary barrier, antibacterial, and osteogenic functions that are matched to implant geometry and mechanical loading. Translation will depend on demonstrating coating integrity, controlled release, and biological safety in relevant long-term models. These requirements define the evidence threshold for selecting a coating technology; the research opportunities required to meet that threshold are considered in the following Section.

5. Future Perspectives and Outlook

Future research should move beyond isolated short-term endpoints toward coating systems and evaluation frameworks that address the remaining translational gaps. The following priorities concern how new coating concepts should be designed, tested, and validated, rather than restating the conclusions above:

- (1) Current studies use different test media, bacterial strains, cell models and evaluation periods, making direct comparison between coating systems difficult. Standardized methods should therefore be established for corrosion current density, antibacterial activity, cell viability, alkaline phosphatase activity and mineralization. Dynamic experimental models that incorporate inflammatory conditions, bacterial challenge and mechanical loading are also needed to better reproduce the implant environment.
- (2) Antibacterial elements, such as Ag, Cu, Zn and Mn, can inhibit bacterial adhesion and biofilm formation, but excessive ion release may reduce cytocompatibility. Future coating design should focus on controlled release and suitable elemental

concentrations. The balance between antibacterial activity, cell proliferation, osteogenic differentiation and corrosion stability should be evaluated within the same experimental system.

- (3) Pores, cracks, residual stress and weak interfacial bonding may cause coating degradation, delamination and particle release. Further research should optimize coating composition, thickness, pore structure and adhesion strength. Graded, multilayer and composite coatings may combine a dense corrosion-resistant inner layer with a bioactive or antibacterial outer layer. Their long-term durability should be examined under wear, cyclic loading and tribocorrosion conditions. For additively manufactured titanium alloys, surface and coating strategies must also accommodate process-dependent roughness, porosity and complex geometry. Comparative work on DMLS- and EBM-fabricated Ti-6Al-4V showed that integrated sandblasting, acid etching and anodization can generate micro/nanoscale topographies with favorable in vitro cell responses [125]. Tribocorrosion-resistant designs likewise require testing under simultaneous mechanical and electrochemical loading. In titanium matrix composites, the response to MAO or thermal oxidation depended on reinforcement architecture and the stability of the oxide-supported subsurface.

Smart or stimuli-responsive coatings are a further emerging direction because they may couple local infection, inflammation or external stimulation to controlled therapeutic release. A dual-stimuli-responsive titanium coating has combined near-infrared photothermal activation and an inflammatory microenvironment to regulate carbon monoxide release while supplying Sr^{2+} for osteogenic support [126]. Immunomodulatory surface design should similarly assess macrophage phenotype together with antibacterial activity and tissue integration. A smart carbon-monoxide-releasing titanium coating has been reported to promote a pro-regenerative macrophage response while providing switchable antibacterial activity [127]. Data-driven design is also beginning to support coating optimization. For example, machine-learning models have been used to predict the wear rate of HVOF-sprayed hydroxyapatite coatings on Ti-6Al-4V and to identify influential spray parameters [128]. These approaches remain early-stage and will require standardized, shareable datasets, externally validated models, and performance testing that integrates corrosion, wear, release and biological endpoints.

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