Study on Application Strategies for Surface Modification to Optimize Biocompatibility of Medical Magnesium Alloy Implants

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Keywords: Medical Magnesium Alloy Implants; Surface Modification; Biocompatibility; Regulating Corrosion

Abstract: Medical magnesium alloys have significant application potential in orthopedics and cardiovascular fields due to biodegradability, mechanical property matching, and biosafety. But they still face problems such as insufficient biocompatibility, excessively fast corrosion rate, and poor interfacial bonding. Surface modification is the core means to optimize its performance, and the core goals focus on improving biocompatibility, regulating corrosion behavior, and matching mechanical properties. This paper reviews four main technologies: coatings, chemical conversion films, surface alloying, and physical modification, and proposes application strategies such as material adaptation, process optimization, and element selection. Current technologies face challenges such as insufficient long-term biological stability, difficulty of multi-performance synergy, and the high cost of large-scale application. In the future, breakthroughs are needed in the directions of intelligent modification systems, green environmentally friendly processes, and multidisciplinary integration, to promote clinical application.

1. Introduction

1.1. Advantages of Medical Magnesium Alloy Implants

Medical magnesium alloys have special benefits in the field of implant materials. First, biodegradability is significant. After implantation in the human body, they can gradually degrade after tissue repair is completed, and the degradation products are metabolized and absorbed without the need for secondary surgical removal, thereby greatly reducing patients' postoperative pain and risk of infection, and they are especially suitable for short-term repair scenarios such as fracture fixation and temporary vascular stents.

Second, mechanical properties highly match human bone tissue, with an elastic modulus about 40–45 GPa, far lower than traditional materials such as titanium alloys (110 GPa) and stainless steel (200 GPa). This can effectively avoid the stress shielding effect, which causes bone tissue to atrophy due to long-term lack of stress, affecting the repair effect.

Third, magnesium is an essential trace element of the human body, participating in physiological processes such as bone formation and energy metabolism. The magnesium ions released during degradation have no toxic side effects and can also promote osteoblast activity, laying a natural bio-friendly foundation for the integration of implants and tissues [1]. Therefore, it has broad application potential in orthopedics and cardiovascular fields.

1.2. Existing Problems of Medical Magnesium Alloy Implants

Although magnesium alloys have significant advantages, clinical application still faces difficulties. First, biocompatibility is insufficient. Pure magnesium or ordinary magnesium alloys have high surface chemical activity, and contact with body fluids can easily induce local inflammation, inhibits osteoblast and endothelial cell adhesion and proliferation, and may even leads to apoptosis. Vascular stents may also cause thrombosis due to platelet aggregation.

Second, the corrosion rate is excessively fast. Magnesium has a low standard electrode potential (-2.37 V) and easily undergoes severe electrochemical corrosion in the weakly alkaline environment

  DOI: 10.25236/icceme.2025.013

of body fluid. The corrosion rate of pure magnesium often far exceeds the ideal degradation speed required for bone tissue repair (0.2–0.5 mm/year), leading to loss of mechanical support of the implant before tissue repair is completed, resulting in fracture or deformation.

Third, interfacial bonding is poor. The natural bonding strength of magnesium alloys with bone tissue or vascular walls is low. In addition, corrosion causes irregular changes in surface morphology, which easily leads to loosening and displacement of the implant, and in severe cases, may requires secondary surgical adjustment, affecting the therapeutic effect.

1.3. Significance of Surface Modification for Optimizing Biocompatibility

Surface modification is the core means to solve the problems of magnesium alloy implants and improve biocompatibility, and its significance is mainly reflected in three aspects. First, constructing a biofriendly interface. By forming a functional modification layer that changes the surface chemical composition, micro-morphology, and charge state of the material, such as introducing bioactive components like hydroxyapatite and chitosan to simulate human tissue components, it can actively guide osteoblast adhesion and differentiation, inhibit inflammatory factor release, reduce foreign body reaction, and improve cell and tissue compatibility.

Second, regulating corrosion behavior. The modification layer acts as a physical barrier to isolate the substrate from contact with body fluid, slowing down the electrochemical corrosion rate. Some modification layers (such as fluoride conversion films and composite coatings) can control the release rate of magnesium ions, avoiding sudden increase in local concentration that causes disorder of tissue microenvironment, and ensuring that implants maintain stable mechanical properties during the repair cycle.

Third, expanding clinical application scenarios. Through targeted modification (such as improving blood compatibility for vascular stents and enhancing bone bonding ability for orthopedic implants), magnesium alloys can meet the special needs of different medical scenarios, break through the limitation of traditional applications solely for simple fracture fixation and promote its use in complex orthopedic repair and cardiovascular interventional treatment fields, providing a new direction for the development of medical implant materials.

2. Core Objectives of Surface Modification of Medical Magnesium Alloy Implants

2.1. Objective of Improving Biocompatibility

Improving biocompatibility is one of the core objectives of surface modification, and optimization needs to be achieved across three dimensions: cells, blood, and tissues [2]. In terms of cell compatibility, modification should enable the magnesium alloy surface to have cell friendliness, promote adhesion, spreading, and proliferation of functional cells such as osteoblasts and endothelial cells, and at the same time induce cells to differentiate in the direction required for repair (such as osteoblasts differentiating into bone cells), providing a seed cell basis for tissue regeneration.

In terms of blood compatibility, for implants such as vascular stents that contact blood, it is necessary to reduce platelet adhesion and aggregation, inhibit activation of coagulation factors, and avoid thrombosis, to ensure that blood circulates normally on the implant surface. In terms of tissue compatibility, it is necessary to reduce foreign body reactions caused by implants, reduce the release of inflammatory factors (such as IL-6, TNF-α), promote stable physiological integration between implants and surrounding bone tissue or vascular walls, and avoid repair failure caused by insufficient compatibility.

2.2. Objective of Regulating Corrosion Behavior

The core of regulating corrosion behavior is to synchronize the corrosion rhythm of magnesium alloys with the process of human tissue repair. First, it is necessary to slow down the corrosion rate, and the corrosion rate of magnesium alloys in body fluid needs to be controlled within the ideal range of 0.2–0.5 mm/year, to ensure that implants do not lose mechanical support ability before

tissue repair is completed (such as 3–6 months required for fracture healing). Second, it is necessary to control the release of corrosion products, avoiding sudden increases in the local concentration of magnesium ions and hydrogen. Excessive magnesium ions will disrupt the tissue microenvironment, and excessive hydrogen may form bubbles affecting tissue healing. Modification is needed to allow products to be released slowly and uniformly.

Finally, it is necessary to guide uniform corrosion morphology, avoiding problems such as local pitting corrosion and intergranular corrosion. This type of non-uniform corrosion will cause a sudden decrease of local mechanical properties of the implant and easily lead to fracture. Modification is needed to ensure corrosion occurs uniformly on the surface and maintain the overall structural stability of the implant.

2.3. Objective of Matching Mechanical Properties

Matching mechanical properties needs to balance surface properties and substrate stability, fitting the mechanical requirements of different implantation sites [3]. First, it is necessary to improve surface hardness and wear resistance. The surface hardness of magnesium alloys is relatively low, and after implantation they are prone to produce wear particles due to friction, which stimulate surrounding tissues. Modification (such as coatings and surface alloying) is needed to improve surface hardness, reduce wear, and avoid inflammatory reactions caused by particles.

Second, it is necessary to maintain the mechanical stability of the substrate. The modification layer cannot damage the mechanical properties of the magnesium alloy substrate, and it is necessary to ensure that the modification layer and substrate bear load collaboratively, avoiding overall toughness reduction or brittle fracture of implants caused by excessive coating thickness or excessive alloying.

Third, it is necessary to adapt to the mechanical requirements of tissue repair. For example, implants for load-bearing bones (such as femurs) need higher surface load-bearing capacity, while implants for non-load-bearing bones (such as phalanges) need better toughness. Targeted modification is needed to adjust surface mechanical parameters, ensuring that implants can both support tissue repair and not affect repair effect due to mechanical mismatch.

3. Main Types of Surface Modification Technologies of Medical Magnesium Alloy Implants

3.1. Coating Technology

Coating technology is to cover one or more layers of thin films with specific functions on the surface of magnesium alloys through physical, chemical, or physicochemical methods, so as to improve surface properties [4]. The core of this technology is to use the coating as a functional carrier to achieve improvement of biocompatibility and corrosion resistance.

The common types are mainly divided into three categories: first, inorganic coatings, commonly used materials include hydroxyapatite and titanium dioxide. The components of this type of coating are close to human bone tissue, which can provide good bioactivity, while also having strong corrosion resistance, and can promote the combination of bone tissue and implants. Second, organic coatings, such as polylactic acid and chitosan degradable organic materials, whose advantage is good flexibility, can closely adhere to the surface of magnesium alloys, and can also load bioactive substances such as antibiotics and osteoinductive factors, achieving the dual effect of modification and treatment. Third, composite coatings, combining the corrosion resistance of inorganic coatings with the functionality of organic coatings, such as hydroxyapatite–polylactic acid composite coatings, which can improve mechanical and corrosion resistance through the inorganic phase, and improve biocompatibility through the organic phase, adapting to more complex clinical needs. The preparation process of coating technology is relatively mature, commonly including spraying, dipping, and sol–gel method, with low operational difficulty, suitable for large-scale application.

3.2. Chemical Conversion Film Technology

Chemical conversion film technology is to immerse magnesium alloys in specific chemical

solutions, and through chemical reactions between base metal magnesium and solution components, generate a layer of inorganic compound film closely combined with the substrate surface, without the additional covering of external coating materials [5]. The key of this technology is to use chemical reactions to spontaneously form a protective film. The process is simple, low cost, and the film layer has strong bonding strength with the substrate, not easy to fall off.

It is mainly divided into three categories: first is phosphate conversion film, generated through the reaction of magnesium with phosphate solutions (such as zinc dihydrogen phosphate solution), forming magnesium phosphate salt film layers. The porous structure of this type of film layer is moderate, which can both slow down corrosion and provide anchoring sites for subsequent cell adhesion, and the preparation process does not require high temperature, with little influence on the mechanical properties of the magnesium alloy substrate. Second is fluoride conversion film, immersing magnesium alloys in hydrofluoric acid or fluoride solutions to generate dense magnesium fluoride film layers. Magnesium fluoride has high chemical stability, can effectively isolate contact between body fluid and magnesium substrate, and greatly reduce the corrosion rate. Third is oxide conversion film, using anodic oxidation or chemical oxidation to generate magnesium oxide or magnesium-based composite oxide film layers on the magnesium alloy surface. This type of film has stable structure, can improve surface hardness, and has certain biocompatibility.

3.3. Surface Alloying Technology

Surface alloying technology is to introduce biocompatible alloy elements (such as zinc, calcium, silver, etc.) into the surface layer of magnesium alloys through specific processes, changing the chemical composition and microstructure of the surface, thereby improving surface properties [6]. Its core is to make alloy elements fuse or diffuse with the surface of magnesium alloys, forming a composition-gradient alloying layer, avoiding the problem of separation at the coating–substrate interface.

It is mainly divided into two categories: first, laser surface alloying, using a high-energy laser beam to irradiate the surface of magnesium alloys, rapidly melting the surface area, and at the same time integrating preset alloy powders (such as zinc powder, calcium powder) into the molten pool, forming a uniform surface alloy layer after cooling. This process has concentrated energy and fast heating speed, can precisely control the thickness and composition of the alloy layer, and the alloy layer and the substrate are metallurgically bonded, with very strong bonding strength, which can significantly improve surface hardness and corrosion resistance. Second, plasma surface alloying, using a plasma generator to produce high-energy plasma, ionizing alloy elements (such as titanium, nitrogen) into ionic state, and then accelerating them into the magnesium alloy surface through an electric field, allowing alloy elements and magnesium surface atoms to diffuse with each other, forming an alloying layer. This process can be carried out at low temperature, avoiding damage of high temperature to the toughness of magnesium alloy substrate, and the alloying layer has uniform thickness, which can effectively improve surface biocompatibility and corrosion resistance.

3.4. Physical Modification Technology

Physical modification technology is to change the morphology, structure, or atomic state of the surface of magnesium alloys through physical means, without relying on chemical reagents or the substantial introduction of alloying elements, and only by adjusting the physical state of the surface to achieve performance optimization. The advantage of this technology is that it does not change the main composition of the magnesium alloy substrate, can retain the original properties of the substrate to the greatest extent, and the process is green and environmentally friendly [7].

It is mainly divided into two categories: first, ion implantation, using an ion accelerator to accelerate specific ions (such as nitrogen ions, oxygen ions, silver ions) to a high-energy state, and implanting them into the surface layer of magnesium alloys, making changes to the atomic structure of the surface, forming a doped modification layer. For example, nitrogen ion implantation can form magnesium nitride phase on the surface, improving corrosion resistance; silver ion implantation can endow the surface with antibacterial properties, and the modification layer has no obvious interface

with the substrate, with firm bonding, not easy to fall off. Second, sandblasting treatment, using compressed air to impact fine sand particles (such as alumina sand, quartz sand) at high speed on the surface of magnesium alloys, changing the roughness and micro-morphology of the surface through mechanical action, making the surface change from smooth to rough concave–convex structure. This type of rough surface can increase the contact area between cells and materials, promote adhesion and spreading of osteoblasts, and at the same time does not introduce foreign substances. The process is simple and very low cost, often used as a pretreatment step for subsequent coating or chemical conversion film, improving the bonding strength of subsequent modification layers.

4. Application Strategies of Surface Modification for Medical Magnesium Alloy Implants

4.1. Application Strategy of Coating Technology

The application of coating technology should focus on material selection, process optimization, and interface reinforcement to ensure that the coating functions match clinical requirements. First, in material selection, hydroxyapatite and other inorganic coatings are preferred for orthopedic implants, as their similarity to bone tissue composition promotes osseointegration [8]. For vascular stents, biodegradable organic coatings such as polylactic acid can be selected to balance flexibility and anti-thrombosis properties. If antibacterial effects are required, active agents such as silver ions or antibiotics can be incorporated into the coating.

Second, process parameters should be optimized. By adjusting spraying pressure and temperature or the gelation time in the sol-gel method, the coating thickness should be controlled within 50-200 μm (excessive thickness may lead to cracking, while insufficient thickness may cause incomplete coverage). At the same time, porosity should be reduced (kept below 5%) to minimize body fluid penetration and subsequent substrate corrosion. Finally, interface bonding should be reinforced. Before coating preparation, sandblasting or acid etching can be applied to remove oxide layers and increase surface roughness, producing an anchoring effect between the coating and substrate. This prevents coating detachment after implantation and ensures long-term service stability.

4.2. Application Strategy of Chemical Conversion Coating Technology

The core of chemical conversion coating technology lies in compositional regulation, process control, and performance enhancement, making use of its advantages of simple operation and low cost [9]. The first step is to regulate the chemical composition of the coating. For orthopedic implants, phosphate conversion coatings are selected, and by adjusting the concentration of zinc dihydrogen phosphate solution (5%-15%), coatings with a moderately porous structure can be formed to provide adhesion sites for cells. For higher corrosion resistance, fluoride conversion coatings are used, with fluoride ion concentration controlled at 8%-12% in hydrofluoric acid solution to generate dense magnesium fluoride layers.

The second step is to optimize the preparation process. The reaction temperature should be maintained between 25-60 °C (excessive temperature may cause porous coatings), and the reaction time should be 10-30 minutes. At the same time, the pH of the solution should be adjusted (pH 3-5 for phosphate coatings, pH 2-4 for fluoride coatings) to avoid uneven local reactions. The third step is performance enhancement. Post-treatment sealing, such as immersion in silane solution, can fill micropores in the coating, further improving corrosion resistance while maintaining bonding strength between coating and substrate.

4.3. Application Strategy of Surface Alloying Technology

The application of surface alloying technology requires emphasis on element selection, precise process control, and performance balance, ensuring that no harmful substances are introduced and that the alloyed layer works synergistically with the substrate [10]. First, in element selection, zinc and calcium, which are essential trace elements for the human body, are preferred (zinc at 2%-5%,

and calcium at 1%-3%). These can enhance corrosion resistance without causing toxicity. For antibacterial purposes, silver can be added in small amounts (≤1%) to avoid cytotoxicity from excessive silver ion release.

Second, precise process control is needed. In laser surface alloying, laser power should be set between 500-1500 W and scanning speed at 5-15 mm/s to ensure uniform alloy layer thickness (50–150 μm) while preventing substrate embrittlement due to local overheating. In plasma surface alloying, plasma energy density should be controlled at 10-30 J/cm², and ion implantation doses at 1016-1018 ions/cm², ensuring uniform diffusion of alloying elements. Finally, performance balance should be maintained. The hardness of the alloy layer should be increased by 30%-50% compared with the substrate (to meet wear resistance requirements), while retaining more than 70% of the substrate toughness to avoid implant fracture under stress.

4.4. Application Strategy of Physical Modification Technology

The application of physical modification technology should follow the principles of goal orientation, parameter adaptation, and synergistic adjustment, while preserving the advantages of the magnesium alloy substrate [11]. For ion implantation, the ion species should be selected according to the desired function: nitrogen ions for improving corrosion resistance (dose 10^{17}-10^{18} ions/cm², forming magnesium nitride layers), and silver ions for antibacterial purposes (dose 10^{16}-10^{17} ions/cm², enabling slow release of silver ions). Implantation energy should be controlled within 50-150 keV to avoid excessive lattice damage or increased brittleness of the surface.

For sandblasting treatment, abrasive particle selection should match the intended application. For coating pretreatment, alumina particles (size 50-100μm) are preferred, controlling surface roughness (Ra) between 1-5μm to enhance coating adhesion. For direct improvement of cell adhesion, quartz particles (size 20–50μm) can be used, producing moderate roughness (Ra 0.5–2μm), which increases cell contact area without damaging cells. After sandblasting, cleaning is necessary to remove residual particles and prevent inflammatory reactions after implantation.

5. Challenges and Prospects of Surface Modification for Medical Magnesium Alloy Implants

5.1. Current Challenges in Surface Modification Technologies

At present, surface modification technologies face three core challenges. First, long-term biological stability remains insufficient. Many modified layers, such as organic coatings and thin conversion coatings, are prone to degradation or wear under prolonged exposure to body fluids and tissue friction, leading to reduced biocompatibility and corrosion resistance, which fail to meet the long-term repair requirements of complex tissues such as load-bearing bones [12].

Second, synergistic optimization of multiple properties is difficult. For example, dense coatings that improve corrosion resistance often reduce bioactivity, while roughened surfaces that enhance cell adhesion may accelerate localized corrosion. Achieving simultaneous improvements in biocompatibility, corrosion resistance, and mechanical properties remains challenging. Third, large-scale preparation costs are high. Technologies such as laser alloying and ion implantation rely on advanced equipment and complex procedures, with limited batch processing capacity. As a result, the cost of modified implants is significantly higher than that of conventional materials, restricting clinical application.

5.2. Future Development Directions

Future surface modification technologies can advance in three directions. First, intelligent modification systems should be developed, such as pH-responsive or enzyme-responsive coatings. For example, biodegradable microspheres can be embedded in coatings to release anti-inflammatory agents automatically when inflammation occurs, enabling performance regulation on demand.

Second, green and environmentally friendly processes should be promoted, including the development of water-based coatings and low-temperature plasma technologies that avoid harmful reagents, reduce energy consumption, and lower costs during production. Third, multidisciplinary

integration should be strengthened by combining materials science, molecular biology, and clinical medicine. New systems can be developed in which modified layers synergize with cells and tissues, such as combining stem cell-inducing factors with modified layers, further enhancing tissue repair efficiency. These approaches will drive magnesium alloy implants toward more precise and safer clinical applications.

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