Current Developments in Surface Coatings of Magnesium Alloys for Biomedical Applications: A Review

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Review
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Magnesium alloys have attracted considerable attention, particularly in the biomedical field, due to their excellent biocompatibility and bone-like properties, which make them suitable materials for biomedical applications. However, their low corrosion resistance and rapid degradation lead to poor functional performance. Therefore, this review focuses on the biomedical applications of magnesium alloys and examines the corrosion mechanisms and corrosion types affecting these alloys when used in the medical field. Among the many techniques, this review focuses on surface coating preparation. Surface coating techniques have helped increase corrosion resistance and biocompatibility while reducing the susceptibility of magnesium alloys to premature degradation. These coatings represent a modern approach for improving the clinical performance of these alloys. However, several challenges remain, including balancing degradation rates with structural integrity and enhancing their compatibility with biological tissues. Accordingly, this review summarizes recent advances in surface coating technologies and provides an overview of the corrosion mechanisms associated with coated magnesium alloys, thereby highlighting future directions and opportunities for their broader application in the biomedical field.

Keywords
magnesium alloy, corrosion mechanism, corrosion types, surface modification, bioimplants

Introduction

In recent years, magnesium alloys have gained significant attention because of their unique properties, including high specific strength and light weight[1–4]. Their lightweight nature makes them attractive for a wide range of industries, including automotive, aerospace, biomedical, electronics, and sporting goods applications. Magnesium possesses a close-packed hexagonal crystal structure, which limits its malleability at room temperature[5]. Conventional magnesium alloys, like AZ31, ZK60, and AZ61 exhibit several limitations, including poor creep resistance, low tensile strength, and susceptibility to corrosion, which restrict their broader industrial utilization. The increasing demand for lightweight materials capable of withstanding harsh operating conditions, particularly in the aerospace and aviation sectors, has stimulated extensive research into the development of magnesium alloys[6]. Their high demand is largely attributed to their low density, which has driven the widespread use of both wrought and cast magnesium alloy products[6,7]. In recent years, the demand for alloys with improved corrosion resistance has further accelerated the development of advanced magnesium alloys. Owing to improvements in their mechanical properties and corrosion resistance, magnesium alloys have become increasingly attractive for biomedical applications, in addition to many other specialized uses[8]. Given the global emphasis on reducing energy consumption and enhancing sustainability, magnesium alloys are expected to play a key role as structural materials in the development of future technologies, including electric cars, modern aircraft, and biomedical applications[9]. Besides permanent implant materials, there is increasing interest in long-lasting biomaterials that slowly degrade within the body supporting bone or tissue regeneration to restore their original roles. This review highlights current magnesium alloy systems used in biomedical applications and discusses their widespread adoption in recent years. Particular attention is given to corrosion mechanisms and the factors influencing corrosion behavior. Strategies for enhancing corrosion resistance, including alloy design and surface treatment are critically examined, highlighting some modern coating technologies possessing unique functional properties. Finally, current challenges and future opportunities in this area are discussed.

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Methodology

A literature search was conducted by meticulously reviewing resources from 2002 to 2026. The search was performed using leading databases such as Web of Science™, Scopus®, ScienceDirect, and Google Scholar, selected for their extensive coverage of materials science, biomedical engineering, chemical engineering, and pharmaceutical sciences. Boolean operators (AND/OR/NOT) were applied in combination with the keywords «magnesium alloys, magnesium alloy, corrosion mechanism, corrosion types, surface modification, bioimplants». The inclusion criteria comprised peer-reviewed original research articles, scoping reviews, and meta-analyses focused on magnesium alloys for biomedical applications. Particular emphasis was placed on studies investigating methods for improving corrosion resistance, such as the development of magnesium alloys and surface treatments. Included were studies reporting on recent progress in strategies and methods for overcoming challenges in adopting magnesium alloys, particularly those emphasizing new surface modification techniques to reduce corrosion and enhance biocompatibility. Additionally, research exploring chemical conversion coatings, biomimetic deposition, micro-arc oxidation (MAO), sol-gel coatings, and ion implantation were considered, all in the context of their effectiveness and contribution to improved corrosion resistance.

The exclusion criteria included conference abstracts, editorials, articles with restricted access, technical reports, doctoral/master's theses, news articles, and non-English publications.

Non-peer-reviewed studies, articles lacking experimental data, and studies not directly relevant to the objectives of this review were also excluded.

A systematic multi-phase strategy was used for study selection and data extraction to ensure methodological rigor and reproducibility. The first step included an extensive database search using specific Boolean operators and filters, such as publication date and article type (peer-reviewed). Relevance to the objectives of this review determined the inclusion of full-text articles.

Included studies offered significant insights into recent advances in addressing the challenges associated with magnesium alloys, especially regarding surface modifications aimed at reducing corrosion and hydrogen evolution while enhancing mechanical properties and biocompatibility. Extracted data included magnesium alloy types, surface treatments for corrosion, mechanical properties, biocompatibility, results, and potential outcomes. The data was synthesized thematically to identify key themes, such as the application of coatings on magnesium alloys in the biomedical field.

Biomedical applications of magnesium alloys: A long history and a promising future

Magnesium alloys have a long history of biomedical use, with their first reported application in 1878 as ligatures to control bleeding from blood vessels[3]. Since then, these alloys, as shown in Table 1, have been extensively investigated for various bio-

Table 1 – *Biocompatibility, corrosion resistance, and mechanical properties of the alloying elements commonly used in magnesium alloys*[4]

Elements	Biocompatibility	Corrosion resistance	Mechanical performance
Al	Al exhibits neurotoxicity and, in some cases, has been associated with Alzheimer's disease and muscle fibre damage.	Improves corrosion resistance.	Enhances the strength and ductility of magnesium alloys.
Zn	Biocompatible and non-toxic.	Increasing Zn content generally improves corrosion resistance.	Contributes to solid-solution strengthening and increases alloy strength.
Mn	Toxic to both cells and nerves.	Improves corrosion resistance.	Increases yield strength but may reduce tensile strength and ductility.
Ca	An essential element involved in bone formation.	Improves corrosion resistance with increasing Ca content.	Increases yield strength but may reduce ductility.
Sr	An essential and vital element in bone formation.	Excessive Sr content may reduce corrosion resistance.	Increases yield strength with increasing Sr content.
Zr	Biocompatible and contributes to strengthening bone bonds.	Excessive Zr content may decrease corrosion resistance.	Acts as a grain refiner, thereby improving strength and ductility of the alloy.
Nd	At high concentrations, Nd exhibits cytotoxic effects, whereas low concentrations are generally considered safe.	Improves corrosion resistance.	Enhances mechanical performance of the alloy.

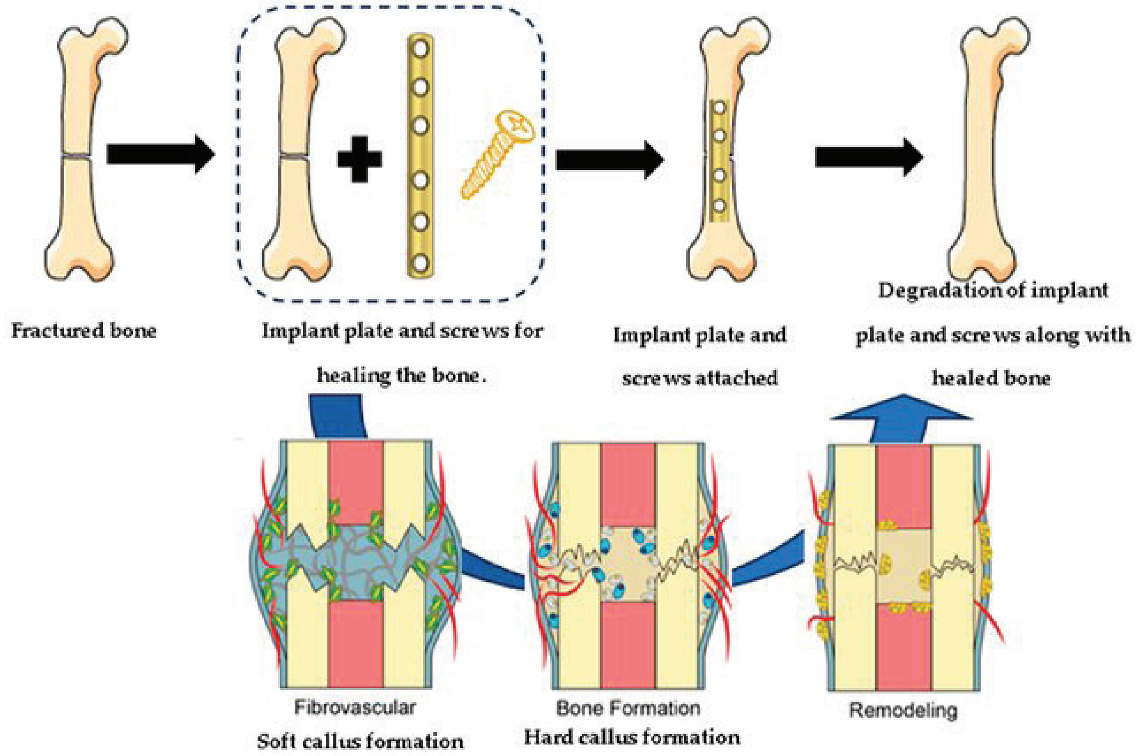

Fig. 1 – *Biodegradable implants of Mg alloy*[11]

medical and surgical applications, including cardiovascular surgery, treatment of musculoskeletal disorders, general surgery procedures, and biodegradable implants[4,10]. Fig. 1 illustrates the principal areas of application for magnesium alloys in medicine[11]. Currently, the most prominent biomedical applications of magnesium alloys include vascular stents and orthopedic implants[11].

To improve the strength of magnesium, alloying is the main method for improving its mechanical properties. However, the addition of alloying elements may lead to the formation of secondary phases that accelerate corrosion. Magnesium alloys such as AE21, AZ32, and AZ91 are examples of Al-containing Mg alloys that exhibit excellent mechanical properties and improved corrosion resistance, making them attractive for biomedical applications[6,11,12]. Nonetheless, concerns regarding the potential role of aluminum in neurodegenerative disorders, such as Alzheimer's disease, highlight the need for balance in biomedical materials. Conversely, zinc (Zn) is an essential trace element for human health and contributes significantly to the strengthening of Mg-Al alloys. Mg-Zn alloys are considered highly biocompatible and suitable for implantation in the human body[7]. An Mn-containing oxide film prevents penetration of Cl^- ion, making it a valuable alloying element in biomedical Mg alloys. Mg-Ca alloys help bone implants, minimizing discrepancies and enhancing corrosion resistance. In addition, strontium (Sr) exhibits similar chemical properties to Mg and Ca. Other elements in magnesium alloys, such as zirconium (Zr), silicon (Si), and lithium (Li), help enhance osteoconjugation strength and improve corrosion resistance in the human body. They also improve the ductility of vascular stents implanted in the human body[10,11]. Overall, alloying elements play a crucial role in explaining and understanding how magnesium alloys degrade and corrode. Consequently, optimizing alloy composition is essential for achieving the desired balance between mechanical integrity and the degradation rate, ensuring that the implant maintains sufficient strength while degrading at a rate compatible with the healing process and tissue regeneration.

Cardiovascular stents

One of the most important biomedical applications of magnesium alloys is the fabrication of biodegradable vascular stents[13–20]. These stents have the potential to overcome the problem of narrowed or blocked arteries by providing temporary mechanical support to maintain adequate blood flow. This leads to the remodeling of the blood vessel as the magnesium stent gradually dissolves and is replaced with normal vascular tissue, thus eliminating the need for surgical intervention to treat narrowed arteries[7,8]. Fig. 2 shows the development of biodegradable magnesium alloy stents[8].

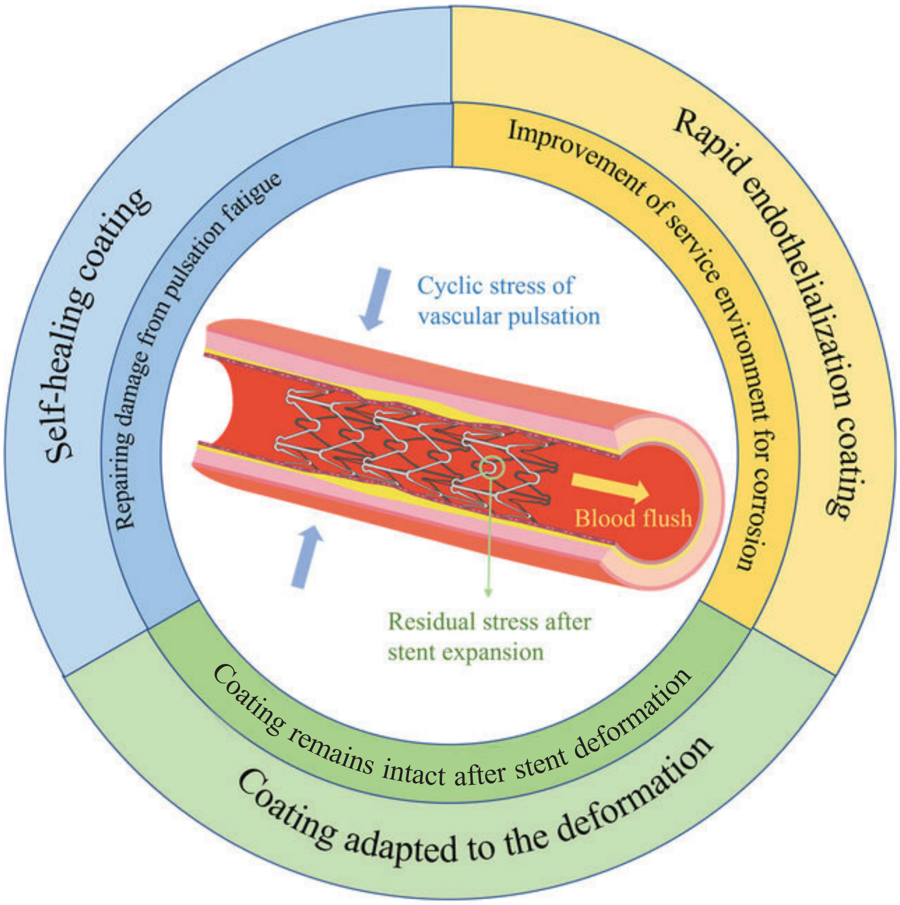

Fig. 2 – *Development of Mg alloy stents*

Sub-bone implants and orthotics

Magnesium alloys have proven their effectiveness as implant materials because of their excellent biocompatibility and mechanical properties that closely resemble those of natural bone[21–23]. They are therefore promising alternatives to conventional implant materials such as titanium (which may cause stress due to higher stiffness). Furthermore, magnesium alloys are non-toxic and capable of stimulating and accelerating the growth of new bone tissue[24–26]. With the advancement of scientific research in this field, it is expected that magnesium alloys will have great importance in the medical field.

Main factors contributing to the corrosion susceptibility of Mg alloys

Magnesium alloys exhibit high electrochemical activity, making them susceptible to corrosion[27]. Several factors contribute to their relatively poor corrosion resistance as discussed further herein.

Negative reduction potential of Mg alloys

Mg alloys possess negative reduction potentials, which result in their high electrochemical activity, especially in aqueous environments, leading to spontaneous reactions and thus increasing their susceptibility to corrosion[4,28].

Formation of microscopic corrosion cells

The electrode potential of magnesium is low, and the hydrogen exchange current density at its surface is relatively low. However, when magnesium is combined with other metallic elements in alloys, microscopic corrosion cells can form between different regions of the alloy, increasing the corrosion current density and accelerating the rate of deterioration[29–33].

Unstable corrosion product layer

During corrosion, a layer of oxidation products forms on the magnesium surface. However, this layer is uncoupled and has a weak structure, making it unable to provide effective protection to the metal surface. Furthermore, the released hydrogen gas destabilizes this layer, leading to the exposure of fresh alloy surfaces to the surrounding environment thereby accelerating the corrosion process[34].

Corrosion mechanisms of Mg alloys

Magnesium alloys are extensively used in the biomedical field[25], because of their favorable combination of high strength, low density, and good biocompatibility. However, a major challenge is their rapid degradation in the body due to their highly negative electrode potential (–2.37 V), resulting in the premature loss of structural integrity before bone regeneration is complete. In addition, the corrosion behavior of magnesium alloys in the human body is highly complex. Although extensive research has been conducted to understand these processes, the corrosion mechanisms of magnesium alloys under physiological conditions remain an active area of investigation. This review explains the electrochemical surface reactions.

When magnesium is exposed to an aqueous environment, it oxidizes and loses its electrons according to the following anodic reaction in Eq. (1)[5,7,8]:

$$Mg \rightarrow Mg^{2+} + 2e^- \quad (1)$$

Magnesium ions are released into the solution, contributing to the disintegration of the alloy over time. The electrons produced by magnesium oxidation react with water or hydrogen ions in the solution, according to the cathodic reaction given in Eq. (2)[5,7,8]:

$$2H_2O + 2e^- \rightarrow H_2 + 2OH^- \quad (2)$$

Hydrogen bubbles are released, which hinder the formation of a stable protective layer on the alloy surface, accelerating the corrosion process according to the reaction given in Eq. (3)

$$Mg + 2H_2O \rightarrow Mg(OH)_2 + H_2 \quad (3)$$

The reaction produces magnesium hydroxide, which forms a weak surface layer that provides insufficient protection for the metal, allowing corrosion to continue over time.

The corrosion of magnesium in the human body is more intricate. Therefore, researchers have made significant efforts to study the behavior of magnesium alloys in the human body. Equations 1 to 3 describe electrochemical reactions occurring randomly on the surface, causing galvanic coupling due to varying potentials between the Mg substrate and intermetallic phases/grain boundaries. In addition, there are other effects resulting from organic molecules adsorbed on the surface of the magnesium alloy. The hydroxyl ions (OH^-) formed during the reactions make the environment alkaline, resulting in a layer of $Mg(OH)^-$ covering the surface of the magnesium alloy. This layer weakens the adhesion of the magnesium alloy, leading to failure of the implantation process.

However, the resulting $Mg(OH)_2$ film's looseness and porosity allow the corrosive medium to reach the Mg substrate, creating pits and more $Mg(OH)_2$. About 150 mmol L^{-1} of Cl^- ions are present in the human body, and these Cl^- ions can degrade the $Mg(OH)_2$ layer, leading to localized corrosion damage. It is important to note that the presence of numerous chloride ions (Cl^-) in the human body causes $Mg(OH)_2$ to dissolve into $MgCl_2$ once the chloride concentration reaches 30 mmol L^{-1}. The following reaction, given in Eqs. (4) and (5), represent the product formed.

$$Mg(OH)_2 + 2Cl^- \rightarrow MgCl_2 \quad (4)$$

$$Mg^{2+} + 2Cl^- \rightarrow MgCl_2 \quad (5)$$

When the magnesium hydroxide layer erodes, the basic mineral substrate, which is magnesium, will be directly in contact with body fluids, and will therefore react with the elements present in the body fluid, which are ions of calcium (Ca^{2+}) and phosphate (PO_4), which act as nucleation and deposition centers, forming a calcium-phosphate apatite layer that helps in bone growth[35]. In addition, carbonates are formed when carbonic acid or xxx carbon dioxide is present. These components can be deposited on the magnesium surface, thus producing a layer composed of corrosion products. The longer the implantation time, the more cells appear on the surface of the magnesium substrate, and thus tissues are created that are close to the corrosion products. Thus, the magnesium suffering from corrosion may separate from the basic basic substrate, as the surrounding fibrous tissues absorb the separated magnesium molecules until complete decomposition is achieved.

Types of corrosion in magnesium alloys

Galvanic corrosion in magnesium alloys

Galvanic corrosion is one of the most common types of corrosion in magnesium alloys, occurring as a result of differences in electrochemical potential between the various components of the alloy[34,36]. During the corrosion process, the magnesium matrix acts as the anodic (positive) electrode, while secondary phases and metallic impurities within the alloy act as the cathodic (negative) electrodes[11]. Galvanic corrosion occurs when magnesium comes into direct contact with another metal of different electrochemical potential, resulting in the flow of electrons between them, accelerating the corrosion of Mg alloys[5,37–39].

Electrochemical mechanism of galvanic corrosion

The primary cause of magnesium corrosion in aqueous environments is that hydrogen ions act as oxidants in the reaction. Since the equilibrium potential for the reduction reaction of these compounds is higher than the oxidation potential of magnesium, the resulting free energy difference is negative, making the reaction thermodynamically spontaneous. In this process, certain regions of the magnesium matrix undergo an oxidation reaction, where magnesium atoms lose their electrons and become magnesium ions (Mg^{2+}). The released electrons move to the cathodic regions, where they react with hydrogen ions (H^+) in the solution, resulting in a reduction reaction and the formation of hydrogen gas[5,36–42].

Factors influencing galvanic corrosion

The severity of galvanic corrosion in magnesium alloys depends on several key factors, most notably the composition of the corrosive medium. Corrosion rates vary depending on the acidity (pH), the presence of aggressive ions such as chloride, and oxygen concentration. Alloy composition, the ratio of elements added to magnesium, such as aluminum, zinc, or rare earth elements, affects the formation of secondary phases that may enhance or reduce the corrosion rate. Also, environmental factors include temperature, humidity, the nature of protective coatings, or mechanical factors that may affect the durability of the protective oxide layer[41–43].

Localized corrosion

Localized corrosion includes several types of corrosion that are concentrated in specific points or regions of the alloy[44–48] as shown in Fig. 3 [45].

Pitting corrosion

Although some metals exhibit good resistance to general corrosion, they are more susceptible to pit corrosion when exposed to aggressive solutions, which can lead to material failure[49]. Magnesium alloys are naturally non-inert metals, and when exposed to chloride ions in a non-oxidizing medium, they tend to corrode and crack outside their normal corrosion range. Pit corrosion occurs due to the formation of plugged cells within the alloy, leading to self-acidification within the corroded point[50]. In the initial stage, the uneven distribution of oxygen leads to the formation of electrochemical cells, which increases the acidity of the local solution within the cavities. As the cation concentration increases, the solution attempts to maintain its electrical neutrality by dragging the chloride ions into the gap, where they react with the metal ions to form soluble chlorides. This reaction increases the corrosion rate in the affected areas, forming small holes in the alloy surface that enlarge over time[51]. Pit corrosion occurs in acidic, neutral, and alkaline solutions, and heavy metal impurities in the alloy accelerate its occurrence[52].

Filamentary corrosion

Filamentary corrosion occurs when the active corrosion cell moves along the grain boundary surface, with the head acting as the anodic (positive) electrode and the tail acting as the cathodic (negative) electrode. This type of corrosion is character-

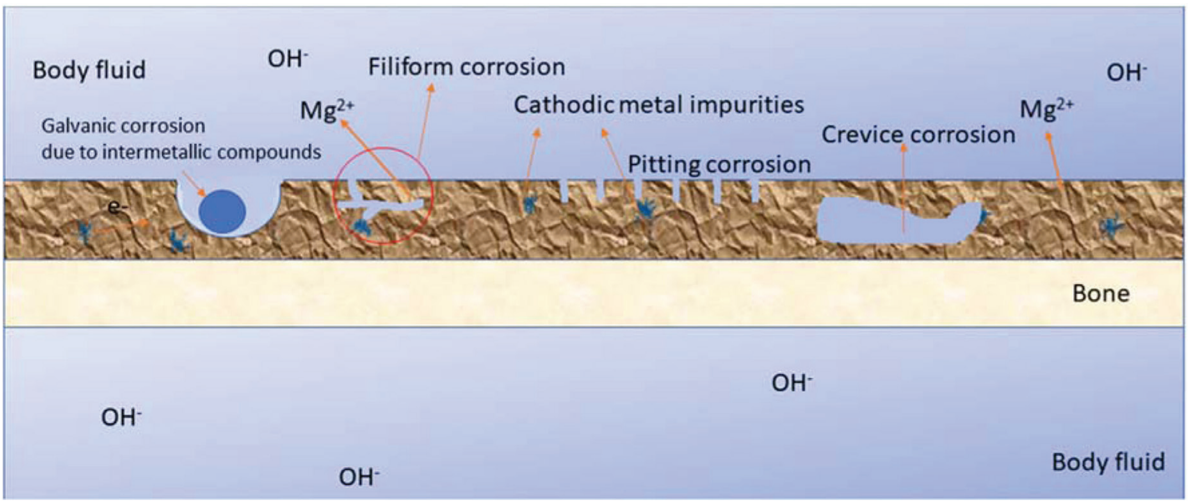

Fig. 3 – *Common localized corrosion types at the bone-implant interface: An overview*[7]

ized by its appearance on protective and oxidized layers, while uncoated magnesium is less susceptible to this type of corrosion[47]. In AZ91 magnesium alloy, pit corrosion begins at specific locations on the surface, then filamentary corrosion develops as the reaction moves along the grain boundaries. The severity of this corrosion is influenced by factors such as temperature, material structure, and surface treatment. It is well known that magnesium alloys oxidize at high temperatures, as this oxide layer does not protect against corrosion, which leads to an increase in the rate of corrosion[53]. Also, the metal impurities present in the alloys may contribute to the acceleration of corrosion, while in some alloys, such as the Mg-RE system, impurities contribute to improving corrosion resistance, especially in applications that require high temperatures[54].

Surface modification

Surface modification plays an important role in improving the performance of magnesium alloys by enhancing their chemical and biological properties[55]. Surface modification techniques not only improve corrosion resistance but also enhance the biocompatibility and bioactivity of magnesium alloys, making them suitable for biomedical applications. Extensive efforts are underway to develop and improve the surface of magnesium alloys. This review focuses on surface coating preparation.

Surface coating preparation

The oxide layer formed on the surface of magnesium alloys is porous and insufficient to render magnesium alloys corrosion-resistant. This has led to numerous efforts to create a highly effective corrosion-resistant layer[56]. Fig. 4 shows that apatite and organic polymer coatings are possible via chemical and physical methods[57]. There are several methods for modifying the surface of magnesium alloys. In this review, we will discuss some of them.

Chemical conversion coatings

One of the most common methods used to protect magnesium alloys from corrosion is chemical conversion coatings. These coatings can be obtained through electrochemical or chemical reactions. These reactions result from the interaction of the alloy surface with chemical solutions containing various ions, such as fluorides, phosphates, carbonates, and chromates. This leads to an increase in the corrosion resistance of the alloys by forming a layer with high cohesion and adhesion[58]. Following are the most popular chemical conversion coatings systems.

Fluoride coatings

Fluoride coatings on Mg alloy surfaces have become a preferred choice in medical research due to their compatibility with the simulated body fluid (SBF), high corrosion resistance, and insoluble properties[59]. Yan *et al.*[60] demonstrated that a durable fluoride coating on the surface of AZ31B alloy improved biocompatibility and corrosion resistance, and the resulting coating also possessed antibacterial properties. Meanwhile, a study conducted by performing surface coating of magnesium alloys with an ultrasonic cleaner resulted in a color change and accelerated degradation of the fluoride coating. This was due to the increased layer thickness, which increased the alloy's corrosion resistance. However, further studies are still needed to examine the degradation of these coatings in specific medical applications, such as cardiovascular diseases.

Phosphate coatings

Phosphating is a surface treatment for magnesium alloys. Phosphate coatings are a thin, non-degradable phosphate layer formed by reacting magnesium with phosphate. This coating is corrosion-resistant, economical, and easy to apply. Liu *et al.*[61] applied a dicalcium phosphate dihydrate coating to protect magnesium alloys from corrosion, and the results showed an increase in the alloy's corrosion resistance. The layer formed was leaf-like, compact, dense, and highly porous. Zai *et al.*[62] also studied the formation of a layer on the surface of AZ31 Mg alloys used in surgical implants through chemical phosphate conversion. Advanced chemical conversion coating techniques need to be developed due to the urgent need to improve the functional performance of magnesium alloy implants. Further studies are needed to investigate the combination of these coatings with ceramic and polymer coatings to obtain coated surfaces with superior mechanical properties and corrosion resistance. In general, chemical conversion coating methods are among the most widely used coating techniques due to their simple processing, and strong adhesion of the resulting coating. These coatings have demonstrated significant success in improving both mechanical properties and corrosion resistance due to the stability of the produced protective layers. However, the main limitations of this method are the difficulty of producing high thickness coating, and the high toxicity of the excess materials used, such as phosphates and fluoride.

Biomimetic deposition

Biomimetic bioprecipitation is a state-of-the-art method that mimics the body's natural apatite formation. Bioceramic films naturally appear on bio-

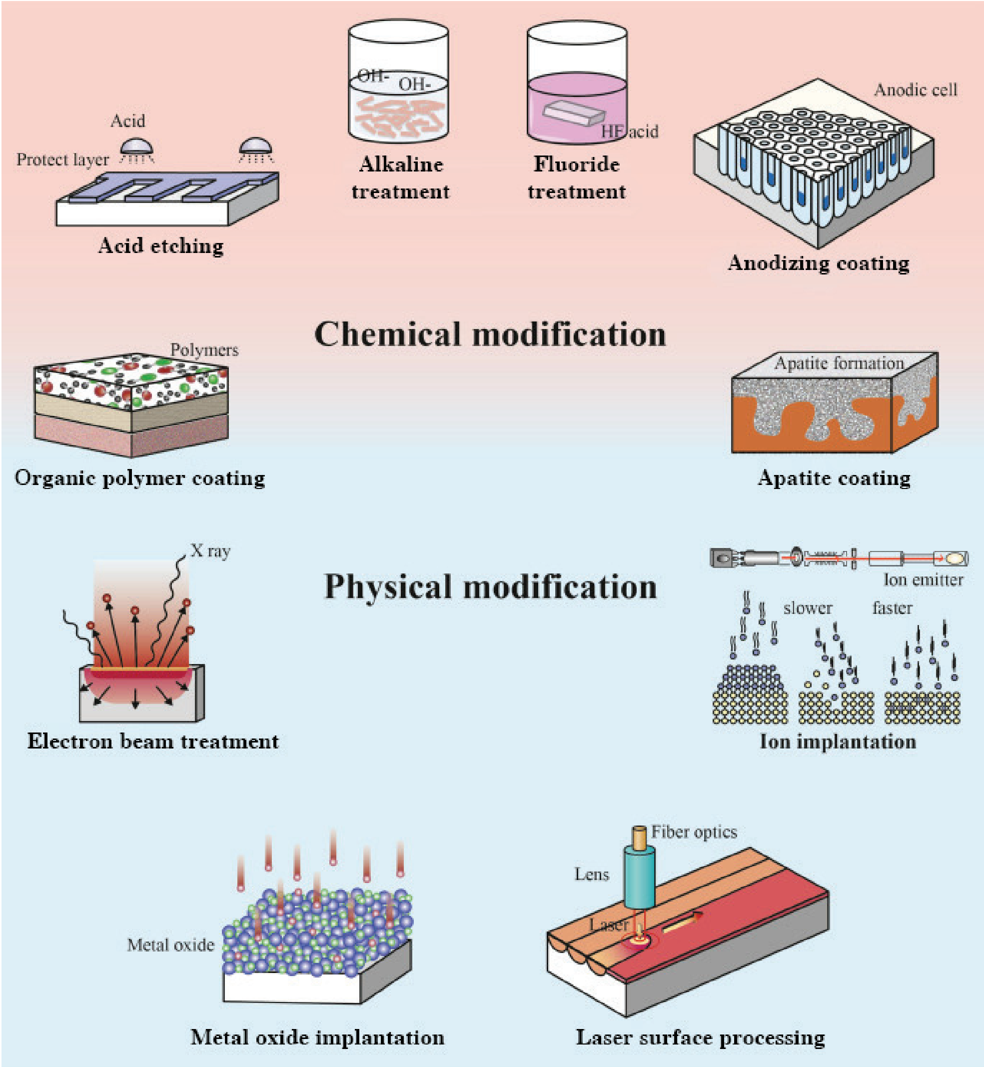

Fig. 4 – *Surface modification of Mg and its alloys using chemical and physical methods*

material surfaces. This method offers numerous benefits, making it a viable choice for medical and bioengineering applications[63]. This supports bone and dental implants and biomaterial modification, as with magnesium and its alloys. Al-Zoubi *et al.*[64] developed a self-assembled hybrid material consisting of graphene oxide (GO), 8-hydroxyquinoline (8-Hq), and an inorganic coating, with the aim of using it as an effective protective layer to reduce the corrosion of the base metal. This technology can produce bioapatite coatings on even complex and porous geometric substrates, such as three-dimensional bone implants. This method also increases the therapeutic efficacy of the precipitation layer by introducing biological agents during precipitation, a feasibility not found in traditional methods that rely on surface adsorption to the layer. Hernandez *et al.*[65] immersed pure magnesium surfaces in a supersaturated calcination solution (SCS) to coat them with bioactive hydroxyapatite (HAp). Utilizing a polydopamine (PDA) primer, Dong *et al.*[66] created composite coatings of calcium phosphate (CaP) and graphene oxide (GO) on AZ60 Mg alloy. Using water instead of ethanol as a dispersant significantly improved corrosion resistance and biocompatibility, as shown by the results. Although biomimetic deposition technology offers significant advantages, it remains under research and development. To improve the stability of the coatings and expand the options for depositable materials, further studies are required to improve bioperformance and expand their use in future medical and engineering applications. Based on its previously mentioned character, biomimetic deposition is considered one of the best methods for natural apatite formation (bioactive coating) and enhancing integration. In addition, it can be applied to complex geometries that are related to living tissues and is able to incorporate biological agents. However, this technique is often limited by its relatively poor mechanical stability, which restricts its use to very small areas. This limitation may also result in lower corrosion resistance

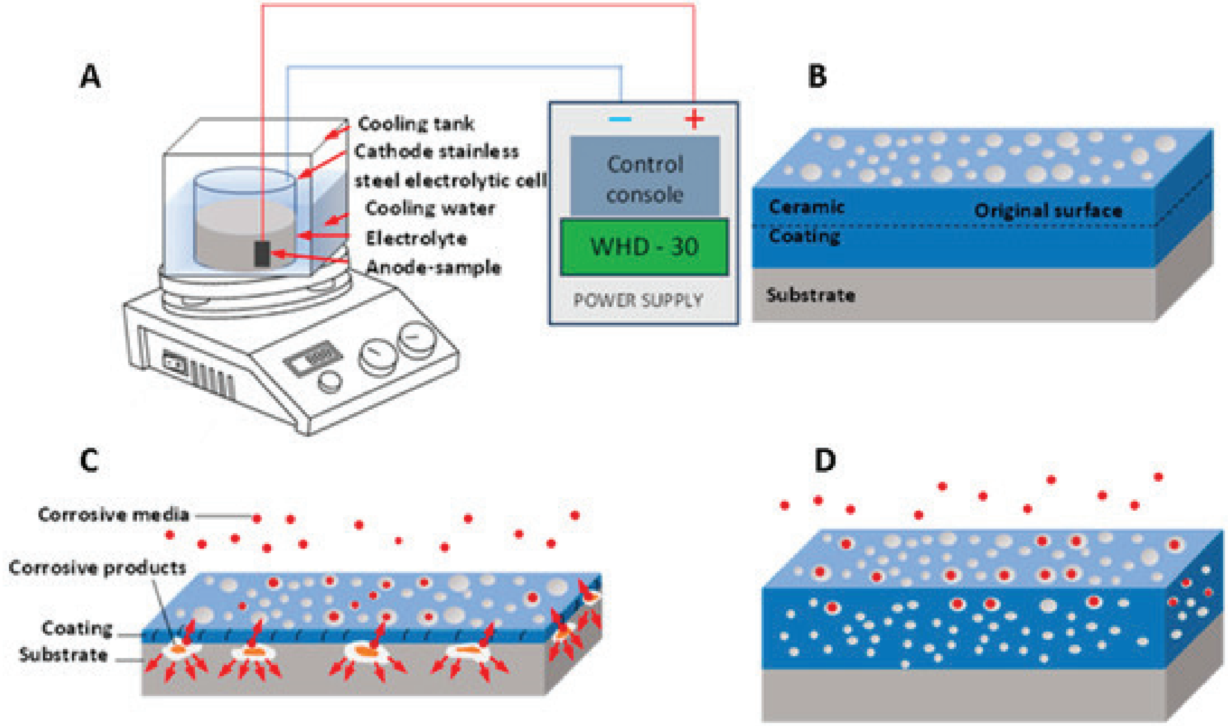

Fig. 5 – *Schematic diagram of the micro-arc oxidation (MAO) coating process: (a) MAO device, (b) MAO coating growth model, (c) thin coating with through-pores, and (d) thicker coating with complex pores*[11]

compared to other magnesium alloy coating techniques, as coating damage can expose the substrate to the corrosive medium attack. Therefore, further research is needed to improve coating stability, broaden the range of depositable materials, and improve bioperformance, thereby facilitating wider adoption of their use in future biomedical and engineering applications.

Micro-arc oxidation coating

One method that uses plasma at high voltages to oxidize metals is micro-arc oxidation (MAO), also known as plasma electro-oxidation (PEO)[27]. The arc discharge used in MAO further enhances and activates the anode reactions, thereby accelerating the formation of a ceramic layer on the metal surface. This process occurs in the presence of an electrolyte. The layer deposited in this way is ceramic, highly corrosion-resistant, and offers strong adhesion, thus providing high mechanical strength, as shown in Fig. 5 [11]. Jia *et al.*[67] conducted a study to develop a rough micro-nanostructured oxide film on Mg-1Ca alloy using a spark electro-oxidation (MAO) technique. The results showed that the corrosion current density of the magnesium alloy in a SBF solution was significantly reduced, resulting in significantly improved corrosion resistance. The benefits of the MAO coating include high corrosion resistance, good abrasion tolerance, and strong substrate adhesion[27,42].

The material's rough structure, characterized by micro- and nano-pores, is effective for drug delivery coatings and promoting cell growth. Despite all these advantages of the layer formed in this way, the corrosion mechanisms of this thin film formed on magnesium alloys have not been comprehensively studied, and therefore require further research. In this context, Lu *et al.*[9] investigated the formation of a thin arc-oxidized layer with properties that enhance corrosion resistance. Their findings demonstrated that corrosion resistance of magnesium alloys can be increased by forming a denser and more adherent film on the surface of the substrate. Compared with other coating techniques, MAO coatings offer superior mechanical strength, excellent adhesion, high corrosion resistance and good wear resistance. Furthermore, the porous structure of MAO coatings can facilitate cell growth and promote drug-delivery capability. However, the high cost and complexity of this technique remain its primary limitations.

Sol-gel coating technology

One of the simplest, most cost-effective, and easily applicable techniques is the sol-gel method. A sol is defined as a colloidal suspension of suitable viscosity[68]. This suspension is used as a layer deposited on the surface of the alloy using various techniques. Sintering is then carried out. During the aging process, chemical transformations occur in the sol layer, resulting in the formation of a gel as shown in Fig. 6 [10,8]. However, this technique has several limitations, including long curing time and poor layer adhesion, which can contribute to increased corrosion rates. Therefore, researchers have investigated various strategies to overcome these drawbacks. Kania *et al.*[68] studied the deposition of a 300 nm thick titanium oxide layer on the surface of a magnesium alloy using two coating methods: magnetic spraying and spin coating. Their results proved that the layer formed by magnetic spraying was better.

Table 2 – *Comparative interpretation of coating methods for Mg alloys*

Coating system	Chemical conversion coatings	Biomimetic deposition	Micro-arc oxidation (MAO)	Sol–gel coating	Ion implantation
Corrosion protection	Moderate	Indirect (bio-driven)	Best	Moderate	Best
Biocompatibility and bioactivity	Good	Best	Good	Moderate	Variable (depends on ions used)
Mechanical strength and adhesion	Moderate	Weak	Best	Weak	Good
Complexity and cost	Low-cost	Moderate	High-cost/high-tech	Low-cost	High-cost/high-tech
Surface structure	Dense	Porous	Porous	Gel-based	No coating layer (modified surface)
Key advantages (best for)	Cost-effective solutions	Biological integration	Durability and corrosion resistance	Cost-effective solutions	Durability and corrosion resistance

Table 3 – *Advantages, disadvantages, and benefits of coating methods for Mg alloys*

Coatings	Advantages	Disadvantages	Benefits
Fluoride	Provide protection for Mg alloy during the initial stage of implantation	Very thin coatings; excessive fluoride may have a negative impact on bone tissue	Enhances biocompatibility
Phosphate	Good corrosion resistance	Porosity and cracks; environmentally unfriendly processing, resulting in higher treatment costs	Stimulates bone growth
Biomimetic	Produce bioactive or biomimetic materials that promote controlled adhesion, osteo-conductivity, and stability of the bone-implant interface	Poor mechanical and chemical stability	Accelerates bone growth
Micro-arc oxidation	Porous coating provides certain corrosion protection for the Mg substrate, and acts as an interlayer to improve adhesion of the composite coating	High cost; requires strict safety measures, may exhibit minor defects such as micro-cracks, micropores, and discharge channels	Accelerates bone growth and good cell biocompatibility
Sol-gel	Good corrosion resistance; applicable to complex shapes and diverse materials	Slow processing; microporosity leads to gas/organic escape and shrinkage; internal stress may cause cracking during heat treatment, reducing corrosion resistance	Provides good conditions for cell adhesion and proliferation
Ion implantation	Improves tribological properties and corrosion resistance; enables selective surface modification without adversely affecting the overall performance; extremely high controllability and repeatability	Possible release of metal ions leading to cytotoxicity	Enhances antibacterial functions

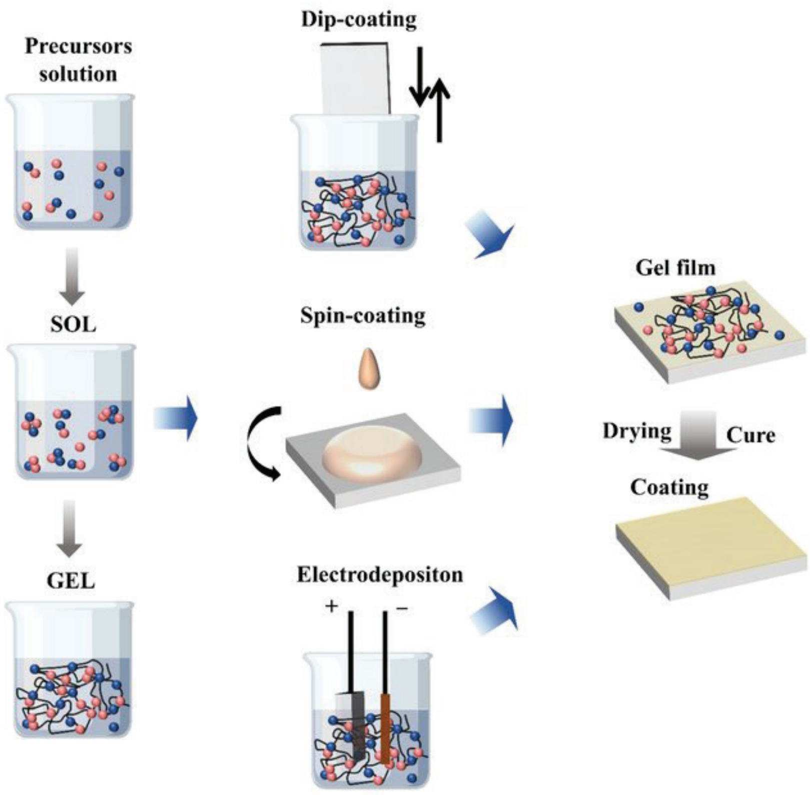

Fig. 6 – *Schematic diagram of sol-gel coating steps*[70]

Ion implantation technology

Ion implantation is an advanced technique used to create a surface layer with a specific composition and structure, and superior chemical and physical properties[69]. The mechanism for creating a layer using ion implantation technology is to bombard the material with an ion beam using high vacuum pressure. In the medical field, this technique is widely used for coating magnesium alloys because it enhances corrosion resistance, increases biocompatibility, and improves the alloy's mechanical performance. The implanted ions used in recent research have included iron ions, zirconium ions, zinc ions, cerium ions, and strontium ions. Non-metallic ions such as carbon, oxygen, and nitrogen ions have also been used as implanted ions[27]. Jia *et al.*[13] improved the performance of magnesium alloys by implanting Sr ions, which resulted in an increase in the modulus of elasticity and toughness, while increasing the corrosion potential of the alloy. In addition, Zhu *et al.*[14] implanted non-metallic ions on the surface of AZ31 alloy, such as N and carbon ions, which led to an improvement in the long-term corrosion resistance of AZ31 alloy. Wang *et al.*[70] showed that implanting metal ions such as Nd in a Mg-Gd-Zn-Zr alloy led to the formation of a composite layer of Nd_2O_3, Gd_2O_3, and Nd, thus improving the corrosion resistance of the alloy because this composite layer contributed to preventing the access of chloride ions to the alloy. Liu *et al.*[71] developed a zinc metal layer on the surface of magnesium alloys using vapor plasma arc deposition. The results showed improved adhesion of the layer to the bone cells, increased proliferation, and increased corrosion resistance. With the continuous development of ion implantation technology, studies are directed towards developing magnesium alloys that combine high mechanical performance, corrosion resistance, biocompatibility, and antibacterial properties through the possibility of implanting multiple ions simultaneously. This opens up vast and unexplored opportunities in the field of medicine. The advantages, disadvantages, and benefits of all the methods discussed are summarized in Tables 2 and 3.

Conclusions and future research

Magnesium alloys possess properties that make them highly suitable for use in the medical field, such as biodegradability and mechanical properties similar to those of natural bone. Recent advances in coating technologies have led to significant improvements in corrosion resistance and bioactivity, resulting in enhanced cell compatibility, blood compatibility, and anti-inflammatory and antimicrobial properties. This review has focused on the corrosion mechanisms affecting magnesium alloys used as implant materials and on strategies for controlling corrosion through surface protection. It has also summarized key surface modification methods used to improve corrosion resistance and extend the functional lifetime of magnesium-based implants. The most prominent methods include chemical conversion, biomimetic, micro-arc oxidation, sol-gel methods, and ion implantation. Although these techniques have proven effective in improving the properties of magnesium alloys, their rapid degradation rate and limited mechanical performance remain major obstacles to their clinical use, especially in applications that require high load-bearing, such as orthopedic implants. Overall, despite these challenges, continued progress in magnesium-based biomaterials offers strong potential for improving the quality of medical implants and contributing to the development of more efficient and sustainable healthcare solutions. To overcome these challenges, future research should focus on several key areas, including the development of advanced multilayer coating systems that combine high corrosion resistance with excellent biocompatibility and simulating biological conditions using computer modeling techniques to gain a deeper understanding of corrosion mechanisms and identify the most influential factors. In addition, advanced future clinical studies should evaluate the performance of magnesium alloys in the body over long periods, facilitating the development of guidelines for their safe use in biomedical applications. Emerging research is also focusing on smart and drug-eluting coatings capable of controlled therapeutic release at the targeted sites.

Future applications may benefit greatly from multilayered hybrid and composite organic/inorganic coatings, where biodegradable polymer outer layers in these coatings serve as drug reservoirs, while the intermediate ceramic layers ensure adhesion between the polymer and the Mg substrate. This approach addresses a key challenge in polymer-based Mg coatings. Therefore, future coating research is expected to increasingly emphasize eco-friendly, cost-effective, hybrid, and multilayered coatings.

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