

Evaluation of the Mechanical and Corrosion Resistance of Magnesium Alloys in a Simulated Human Body Environment

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Abstract:

This study focuses on the evaluation of mechanical and corrosion properties of selected magnesium alloys (AZ31, AZ63, AZ91 and AE21) and commercially pure magnesium in a simulated human body environment. The aim of the work was to compare the behaviour of these materials under conditions relevant to biomedical applications. The materials were subjected to mechanical testing and corrosion analysis using electrochemical methods in a simulated body fluid. The results showed that the AZ91 alloy achieved the highest mean hardness (61.1 ± 1.88 HV) and exhibited the most favourable electrochemical corrosion behaviour, as indicated by the lowest corrosion current density (7.041 mA/cm^2) and calculated corrosion rate (0.161 mm/year). These findings suggest that AZ91 is a promising candidate for applications requiring improved mechanical strength and corrosion performance in physiological conditions. The main contribution of this study is the direct experimental comparison of the Vickers hardness and electrochemical corrosion behaviour of commercially pure magnesium and four magnesium alloys (AZ31, AZ63, AZ91, and AE21) under identical experimental conditions, providing a consistent basis for evaluating the influence of alloy composition on their performance in a simulated physiological environment.

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1. INTRODUCTION

Magnesium is an indispensable component of nature and the human body. Due to its high reactivity, it occurs in nature dominantly in the form of compounds. It is present in silicates, carbonates, and chlorides, where it exists in the oxidation state Mg^{2+} . Furthermore, it is a significant component of seawater. As an element, magnesium is also part of the structures of the human body, where it supports enzyme function, proper muscle and nerve activity, and participates in numerous biochemical reactions [1].

Owing to its exceptional properties, such as biocompatibility, low density, and low weight,

magnesium has attracted increasing attention from biomedical engineers and implant manufacturers over the years as a promising material for biodegradable implants intended for internal use [2-4].

However, magnesium has several limitations for medical applications. It exhibits low corrosion resistance, particularly in environments containing chloride ions (Cl^-), and therefore its corrosion rate in body fluids would be excessively high. In addition, corrosion of magnesium leads to the evolution of hydrogen gas, which may adversely affect the surrounding biological environment. Improving the properties of magnesium and its alloys has been the subject of extensive research

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for many years. One of the methods for modifying material properties is alloying, which allows adjustment of both mechanical and corrosion properties through the appropriate selection of alloying elements. Before selecting suitable alloying elements, it is necessary to evaluate the mechanical and corrosion behaviour of magnesium alloys, for example in a simulated body environment [5-7]. Excessively rapid degradation may also lead to premature deterioration of the implant's mechanical integrity before the surrounding tissue has sufficiently healed [3, 4, 8]. Mechanical properties are typically assessed using tensile tests and hardness tests, which show a strong correlation. One of the most used methods is the Vickers hardness test. Corrosion resistance is most often evaluated using potentiodynamic polarization and electrochemical impedance spectroscopy (EIS). These techniques are widely used for simulating material behaviour in the conductive environment of the human body. A suitable combination of tests and measurements provides valuable insight into the corrosion behaviour of materials such as magnesium [2, 9].

Magnesium and its alloys have been investigated for a wide range of biomedical applications, particularly for temporary orthopaedic and cardiovascular implants. In orthopaedics, magnesium-based materials have been considered for fracture-fixation devices such as screws, plates, pins, and other temporary fixation components, where their biodegradability may eliminate the need for subsequent implant-removal surgery. In cardiovascular applications, magnesium alloys have been investigated primarily for biodegradable vascular stents. More recent research has also explored magnesium-based materials for porous scaffolds, tissue engineering, and other regenerative applications [5, 6, 10]. These applications benefit from the ability of magnesium-based materials to provide temporary mechanical support and subsequently degrade as tissue healing progresses.

Potentiodynamic polarization has also been successfully applied to evaluate the corrosion behaviour of AZ31 magnesium alloy and the effect of surface modification on its electrochemical response [11].

Recent studies have demonstrated considerable progress in the development of magnesium alloys for biomedical applications, particularly through alloying, microstructural modification, surface treatments, and advanced manufacturing techniques [3, 4, 6-8, 12-14]. Magnesium alloys

have attracted attention because of their low density, favourable specific mechanical properties, biocompatibility, and biodegradability, which make them promising candidates for temporary biomedical implants [3, 6]. Nevertheless, their broader application remains limited primarily by rapid and insufficiently controlled degradation in physiological environments. Excessive corrosion may result in premature loss of mechanical integrity and hydrogen evolution before the surrounding tissue has sufficiently healed [3, 15]. Consequently, current research is focused on achieving an appropriate balance between mechanical performance and controlled corrosion behaviour through alloy composition, processing, and surface modification. The type and concentration of alloying elements play an important role because they can influence both the mechanical properties and electrochemical behaviour of magnesium-based materials [3, 12, 15].

Magnesium and its alloys have been investigated for various temporary biomedical devices, particularly in orthopaedic and cardiovascular applications [3, 5, 7, 13, 14, 16, 17]. In orthopaedics, magnesium-based materials have been considered for temporary fixation devices, while cardiovascular research has focused particularly on biodegradable vascular stents [13, 17]. Their gradual degradation offers the potential advantage of providing temporary mechanical support followed by progressive degradation after the implant has fulfilled its intended function [5, 10, 13].

This research focuses on commercially pure magnesium and commonly used AZ and AE magnesium alloys. However, these alloys are currently not sufficiently discussed as potential biomaterials. The aim of this study is to evaluate the influence of the type and number of alloying elements on the mechanical and corrosion behaviour in a simulated body environment and to compare their mechanical and electrochemical corrosion behaviour under conditions relevant to biomedical applications. Identifying an optimal combination of magnesium and alloying elements could significantly expand the use of this metal in biomedicine and potentially replace less suitable materials currently used for implants, stents, screws, and similar devices. The main advantages of magnesium alloys for biomedical applications include their low density, favourable strength-to-weight ratio, biocompatibility, biodegradability, and mechanical characteristics that are closer to

those of human bone than conventional permanent metallic biomaterials [3, 5, 6, 16]. Their gradual degradation can potentially eliminate the need for secondary surgery to remove temporary implants, while magnesium ions released during degradation participate in physiological processes [1, 5, 10]. However, magnesium alloys also exhibit important limitations. Their major disadvantages are rapid and difficult-to-control corrosion in physiological environments, hydrogen evolution during degradation, and the associated premature deterioration of mechanical integrity [3, 5, 15, 16, 18]. Therefore, controlling the degradation rate while maintaining sufficient mechanical performance remains one of the principal challenges in the development of magnesium-based biodegradable implants. Previous studies have extensively investigated the mechanical properties, degradation behaviour, and corrosion resistance of magnesium alloys, with particular attention given to alloys such as AZ31 and AZ91. However, differences in alloy composition, testing media, surface preparation, and experimental procedures complicate direct comparison of results reported by different authors. Therefore, a comparative evaluation of several magnesium alloys under identical experimental conditions can provide useful information on how alloy composition influences their behaviour. The main contribution of this study is a direct experimental comparison of commercially pure magnesium and four magnesium alloys (AZ31, AZ63, AZ91, and AE21) under the same experimental conditions. The study specifically compares their Vickers hardness and electrochemical corrosion behaviour in 0.9% NaCl solution. By evaluating materials with different contents of aluminium, zinc, and rare-earth elements within a single experimental framework, the study provides a consistent basis for assessing the relationship between alloy composition, hardness, and electrochemical corrosion behaviour under the investigated conditions.

2. METALLIC MATERIALS IN BIOMEDICAL ENGINEERING

Metals used for medical applications are characterized by their excellent thermal and electrical conductivity. Today, besides fracture fixation, they are used in load-bearing implants such as hip and knee joints. They can also be utilized in artificial heart valves and vascular stents. In all these applications, metals are mainly used thanks to

their corrosion resistance and mechanical properties.

In addition to pure metals, metal alloys are widely used. These contain two or more types of elements. Alloys can be produced in several ways, such as alloying, surface treatments, or casting. The dominant metallic biomaterials are pure titanium (Ti), titanium alloys (Ti–6Al–4V), stainless steels, and cobalt–chromium alloys.

When choosing a suitable biomedical metal, the following factors play a role: appropriate mechanical properties, reasonable cost, biocompatibility, and corrosion resistance [19].

The lifespan of an implant is defined as the period during which the implant performs its function in the body without failure, damage, or the need for replacement. Implants can be divided into the following categories based on durability (degradation time): permanent implants and degradable implants.

Permanent implants are materials designed to last in the patient's body for a long time—up to several years—without significant damage. Examples include joint replacements and dental implants. Permanent implants must have good mechanical properties, such as strength and hardness, and be biocompatible. The expected medium-term lifespan of permanent implants ranges from 10 to 12 years after implantation, unless wear or damage occurs. Factors influencing implant longevity include the patient's diagnosis, age, weight, and the material used.

Degradable materials are those that gradually break down in the patient's body and are replaced by new tissue. They are mainly used as temporary implants in stents, screws, and scaffolds. The main advantage of degradable implants is that there is no need for secondary surgeries to remove or replace the implant [20].

Biodegradable implants are made from materials such as polymers, metals, and ceramics. The three main biodegradable metals (BMs) with potential for implant manufacturing are magnesium, zinc, and iron. Magnesium-based implants gradually degrade in the physiological environment, releasing Mg^{2+} ions into the surrounding tissues [21]. For comparison, the recommended dietary allowance for magnesium in adults ranges from 310 to 420 mg per day, depending on sex and age [22]. The estimated lifespan of degradable implants ranges from several weeks to several months [21].

The main limitation of permanent implants is the difference in elastic modulus compared to human bone. In permanent implants, stress can develop

between the bone and the implant, which may lead to damage to both the bone and the implant.

Implants that fail earlier than their intended lifespan manifest as implant movement or persistent pain for the patient. If an implant outlives its expected lifespan, surgery is required to replace it with a new one. In practice, secondary implants never last as long as the first implant placed. This leads to several complications for the patient: the need for repeated surgeries, inflammation, pain, and immune-related issues [23].

Metallic materials are primarily used in biomedical applications for fracture fixation and joint replacement. The human body environment is relatively aggressive, making metallic implants susceptible to corrosion. One of the most used metals for implants is titanium; however, it is not an ideal material for several reasons. The production and processing of titanium are financially demanding, as it does not occur naturally in its pure form. Another disadvantage of titanium materials is the significant difference in properties between titanium and human bone. Titanium alloys have a modulus of elasticity of approximately 110 GPa, whereas human bone exhibits a modulus in the range of 5–20 GPa [2]. This considerable mismatch results in high stress at the bone–implant interface.

Despite these drawbacks, titanium still offers more advantages than disadvantages. It is widely used for implants due to its excellent corrosion resistance, high strength, biocompatibility, and low weight. In practice, titanium alloys are preferred over pure titanium, as they exhibit improved physical, chemical, and corrosion properties.

In recent years, magnesium has become the focus of biomedical engineering research. Magnesium possesses favourable mechanical properties and, unlike titanium, is biodegradable. In the human body, magnesium reacts with water and gradually degrades, eliminating the need for secondary surgical procedures to remove implants. During degradation, magnesium releases cations that are essential for proper metabolic processes, contribute to enzyme activity, and support bone growth. However, the main limitation of magnesium remains its rapid degradation and pronounced corrosion behaviour [1, 2].

Material hardness is most measured in practice using indentation methods. The principle of hardness testing is based on evaluating the material's resistance to surface deformation. This deformation is induced by the mechanical action of a harder indenter with specified geometry. Less commonly used methods include rebound techniques, which can be classified as dynamic tests. Hardness cannot be considered a

fundamental physical quantity, as it is based on comparative measurements. The evaluation of hardness involves parameters such as applied force [N] and indentation area. The most widely used methods include Brinell (STN EN ISO 6506-1), Vickers (STN EN ISO 6507-1), Knoop (ISO 4545-1), and Rockwell (STN EN ISO 6508-1) hardness tests [19, 20].

3. PROCESS OF BIODEGRADATION OF METALS

Biodegradation is a process during which compounds with high molecular weight are transformed into compounds with lower molecular weight. The resulting compounds are less complex than the original ones. Through biodegradation processes, organic compounds are converted into simple inorganic molecules, mainly water and carbon dioxide. Biodegradation is sometimes also referred to as biomineralization. The metabolism of microorganisms that use organic substances as substrates for their growth and for enzyme induction is also classified among biodegradation processes. It is important to note that the presence of a growth substrate or another utilizable compound is essential for maintaining biomass and inducing enzymes or cofactors involved in biodegradation. This process is known as cometabolism [24].

BMs are metals and their alloys that are expected to gradually corrode in an *in vivo* environment, while the corrosion products elicit an appropriate host response. After the immune response, the metals are completely dissolved. An important role of BMs is the promotion of tissue healing. Currently, three metals and their alloys appear to be the most promising: magnesium-, iron-, and zinc-based BMs. The concept of biodegradation has been known in medicine for a long time, with degradable polymer sutures being a typical example. BMs are a relatively new development in this field; however, their potential benefits for patients are becoming increasingly evident. To date, magnesium alloys are the most extensively studied BMs, mainly due to their low toxicity and weak immune response.

In 2005, Zartner et al. [17] reported the successful implantation of a magnesium-based stent into the left pulmonary artery of a premature infant suffering from a severe congenital heart defect. Postoperative examinations showed that complete degradation of the stent occurred within five months without damage to the surrounding tissue [10, 17].

Magnesium and its alloys belong to biocompatible materials. In the human body, however, they also have the potential to be biologically resorbable. Magnesium is currently considered a revolutionary bioresorbable material for temporary orthopaedic and cardiovascular applications. It can be used in fracture-fixation devices such as rods, plates, and screws, as well as in temporary cardiovascular stents.

Magnesium offers many advantages in temporary biomedical applications compared with, for example, stainless steel or titanium. An important factor is that magnesium has a density and elastic modulus similar to human bone. Therefore, excessive stress at the metal–bone interface does not occur. Furthermore, it is known that magnesium and its alloys do not cause toxicity or allergic reactions in the human body.

Since magnesium alloys completely degrade, the need for secondary surgeries to remove temporary implants can also be eliminated. Thanks to magnesium's osteoconductive character, faster bone regeneration is supported. Magnesium is natural to the body, and therefore its release from biodegradable implants positively influences wound healing, fracture repair, and tissue growth. It is recognized as an element involved in more than 300 enzymatic reactions [1].

On the other hand, a few disadvantages of magnesium alloys can also be identified. The main issues remain the excessively rapid degradation of implants, hydrogen release, and the deterioration of mechanical properties in simulated human-body environments [16].

According to some authors, magnesium shows unusual corrosion behaviour compared to materials such as aluminium and iron. In recent years, the idea has emerged that magnesium has a different corrosion mechanism, known as the negative difference effect (NDE). This phenomenon describes a situation where the rate of hydrogen evolution from degrading magnesium alloys increases with increasingly negative anodic polarization of magnesium. In addition to typical electrochemical corrosion, magnesium is also known to undergo fragmentation (the so-called chunk effect). Hydrogen evolution leads to hydrogen embrittlement, which occurs along grain boundaries [18].

Several factors influence the rate of biodegradation. The most well-known include alloying, precipitation hardening, and grain size. Among alloyed systems, Mg–Al–RE alloys (E alloys) show the lowest biodegradation rate; however,

they also do not achieve the required level of mechanical strength. Mg–Al–Zn alloys (AZ alloys) follow. In recent decades, it has been shown that lanthanides support the treatment of bone diseases due to their positive effect on bone tissue regeneration and their potential to replace calcium ions with lanthanide ions in osteoporosis therapy. Zinc, as part of alloyed magnesium implants, favourably influences macrophage activity and thus participates in micro-immunological responses in bone disorders.

Magnesium alloys biodegrade rapidly. However, they have the ability to form a layer of corrosion products that creates a kinetic barrier on the material's surface. This prevents the migration of Mg^{2+} cations from the implant into the body [15]. In the presence of water, magnesium undergoes an anodic reaction, leading to the production of Mg^{2+} cations [15]. At the same time, a cathodic reaction occurs, releasing gaseous H_2 . However, the human body contains more than just water. As is known, blood plasma contains proteins, water, and ions. The most important plasma ions include Cl^- , Na^+ , K^+ , HCO_3^- , Mg^{2+} , and PO_4^{3-} [15].

Chloride ions are quite aggressive and disrupt the surface of magnesium materials. In contrast, PO_4^{3-} ions increase the resistance of implants because they act as a barrier to corrosion [5]. Corrosion kinetics are influenced not only by inorganic ions but also by organic components such as proteins, cells, and biomolecules. The presence and growth of endothelial cells and osteoblasts on the implant surface slow down degradation and corrosion rates [5].

In order to model the behaviour of magnesium alloys in the human body environment, solutions that simulate the behaviour and chemical composition of the organism are required. Such simulated solutions include, for example, Hank's solution, simulated body fluid (SBF), Ringer's solution, or synthetic saliva. A simple and inexpensive option for a simulated solution is a 0.9% NaCl solution [25]. Due to its simplicity, physiological saline mimics the ionic composition of body fluids and is therefore suitable for simulating the degradation of metals in the body [25, 26].

The degradation rate decreases for AZ alloys from AZ31 (2.67 mm/year) to AZ91 (1.27 mm/year) [22]. These results were obtained by immersing AZ alloys in simulated body fluid. To reduce the degradation rate, extensive research has focused on mechanical properties and corrosion behaviour, aiming to improve the biomedical performance of

magnesium alloys through alloying elements, processing techniques, and other strategies [27].

The selected alloying material, like magnesium itself, must be non-toxic and easily degradable. Alloying elements intended for use in the human body can be classified according to their mineral composition and related effects, such as toxicity and allergic reactions, as follows [12]:

- Mineral elements: Ca (calcium), Mn (manganese), Zn (zinc), Sn (tin), Sr (strontium).
- Toxic elements: Cd (cadmium), Be (beryllium), Pb (lead), Ba (barium), Th (thorium).
- Allergenic elements: Al (aluminium), Co (cobalt), V (vanadium), Cr (chromium), Ni (nickel), Ce (cerium), La (lanthanum), Cu (copper).

It has been demonstrated that a magnesium-zinc alloy (Mg-Zn) did not cause damage to red blood cells and promoted cell adhesion, confirming its *in vitro* biocompatibility [12].

The most commonly used resorbable implants today are made from biodegradable polymers and bioceramics. However, polymers are mechanically weak, and ceramics are relatively brittle. These negative properties limit their potential use in prosthetics. Magnesium alloys, due to their favourable mechanical and corrosion properties, could be used not only as orthopaedic implants, such as bone plates, but also as cardiovascular implants [5, 25].

Zinc plays an important role in supporting various biological functions, such as protein metabolism, cell growth, and division. As an inorganic element, it promotes the growth of bone tissue around bioimplants. When used as an alloying element in magnesium alloys, it improves corrosion resistance because its stable, moderate degradation is more favourable than that of pure magnesium.

Furthermore, zinc has the lowest melting point among common alloying elements, allowing casting in an open atmosphere and facilitating the manufacturing process. Due to its low chemical reactivity and good machinability, zinc also contributes to reducing implant production costs [12].

Currently, degradable magnesium alloy implants are already available on the market, protected with coatings such as MAGNEZIX® and Magmaris®. In modern biomedical practice, not only biocompatible but also biodegradable and bioactive implants are coming to the forefront. These implants may include secondary functions, such as post-operative tissue nourishment [6].

4. MATERIALS AND METHODS

The experimental procedure consisted of the following main stages: (1) selection of five investigated materials—commercially pure magnesium Mg 99.9 and magnesium alloys AZ31, AZ63, AZ91, and AE21; (2) preparation of the sample surfaces for hardness testing; (3) Vickers hardness testing at three different locations on each sample; (4) separate surface preparation of the samples for electrochemical testing; (5) stabilization of the open-circuit potential for 10 min in a fresh 0.9% NaCl solution; (6) potentiodynamic polarization testing using a three-electrode electrochemical cell; and (7) evaluation and comparison of the obtained hardness and electrochemical corrosion parameters. The same experimental procedure was applied to all investigated materials to ensure their direct comparison under the investigated conditions. The experimental investigation therefore consisted of two principal parts: mechanical characterization by Vickers hardness testing and electrochemical characterization by potentiodynamic polarization.

In hardness tests, indenters are pressed into the surface of the tested material under a constant load F , producing an indentation with a surface area S . The hardness is then determined as the ratio F/S and is denoted as HBW (Brinell), HV (Vickers), or HK (Knoop).

The material samples, after grinding, are placed on smooth, clean, and rigid supports. The load F must act perpendicular to the surface. The minimum thicknesses of samples for hardness testing are precisely defined by standards for each measurement method and the corresponding loads. The tests differ from one another mainly in the shape of the indenter (e.g., a square-based pyramid, a hard metal ball, etc.).

In the Vickers hardness test, the load values can be selected as $F = 1.961$ (0.2); 2.942 (0.3); 4.903 (0.5); 294.2 (30); 980.7 (100) N, etc. The Vickers hardness is then denoted as HV 0.2; HV 0.3; HV 0.5; HV 30; HV 100. The designation HV is given after the hardness value and is supplemented by an index characterizing the test conditions in the following order: test load (N) and dwell time (s) [16, 17]. In the present study, the mechanical characterization was limited to Vickers hardness testing; tensile strength, yield strength, ductility, and fatigue properties were not experimentally evaluated [28, 29].

A total of five samples were investigated, one sample for each material (Mg 99.9, AZ31, AZ63,

AZ91, and AE21). Three Vickers hardness measurements were performed at different locations on the surface of each sample, resulting in a total of 15 hardness measurements. The hardness values are reported as mean $\pm$ standard deviation (SD) for each material. The standardized chemical composition of the alloys is shown in Table 1.

Table 1. Chemical composition of the investigated magnesium alloys (wt.%), adapted from [30]

Content [wt.%]				
Alloy	Al	Zn	RE	Mn
AZ31	3	1	-	0.3
AZ63	6	3	-	0.3
AZ91	9	1	-	0.3
AE21	2	-	1	0.3

A static Vickers hardness test was performed on the samples. Prior to the hardness measurements, each of the five samples had to be ground using P4000 grit sandpaper, then cleaned with isopropanol and dried with a stream of air. This sample preparation before measurement serves to remove undesirable oxides and corrosion products from the material surface, as well as to ensure that surface irregularities do not affect the measurement results.

The hardness tester used for the measurements was the INNOVATEST FALCON 450. The indenter had the shape of a square-based pyramid with an apex angle of 136° . The samples were placed on the stage of the hardness tester in the given order, and the test load and dwell time (30 s) were selected.

After starting the device, the indenter approaches the surface until contact is made, and then gradually reaches the full test load. After 30 s at full load, the indenter is removed, and the diagonals of the indentation are measured using optical microscopy. The system then automatically calculates the Vickers hardness value. The Vickers hardness measurements were performed at the Department of Materials Engineering, Faculty of Mechanical Engineering, University of Žilina, Slovakia.

The potentiodynamic polarization test range was from -200 mV to $+300$ mV relative to corrosion potential. The potential scan rate was $1 \text{ mV}\cdot\text{s}^{-1}$. Corrosion measurements were performed using a BIOLOGIC SP300 laboratory potentiostat. A three-electrode configuration was employed, with the material sample serving as the working electrode, a platinum mesh as the counter electrode, and a saturated calomel electrode (SCE) as the reference

electrode. Electrochemical techniques were performed in a 0.9% NaCl solution at a laboratory temperature of $22 \pm 2^\circ\text{C}$. The potential stabilization time prior to each measurement was set to 10 min. Potentiodynamic polarization testing was performed once for each investigated material, without independent experimental repetitions. Tafel analysis of the measured curve was used to determine the slopes of the cathodic (β_c) and anodic (β_a) branches. The linear regions of the Tafel curves intersect at the corrosion potential (E_{corr}) and corrosion current density (i_{corr}). The corrosion rate (v_{corr}) was subsequently calculated. Before testing, the samples were ground using P1200-grit abrasive paper, rinsed with water and ethanol, and subsequently dried. Fresh 0.9% NaCl solution (physiological saline) was used for each measurement. The solution was prepared by dissolving 9 g of NaCl in 1 L of distilled water. Potentiodynamic polarization testing was performed once for each investigated material. Therefore, the electrochemical results are presented as individual experimental values and are used for a comparative assessment of the corrosion behaviour under the investigated conditions.

The resulting potentiodynamic curves were analysed using Tafel analysis, as shown in Fig. 1, from which the values of the corrosion potential, corrosion current density, and the coefficients were determined, and the corrosion rate was subsequently calculated. The potentiodynamic polarization measurements were performed at the Research Centre of the University of Žilina, Slovakia.

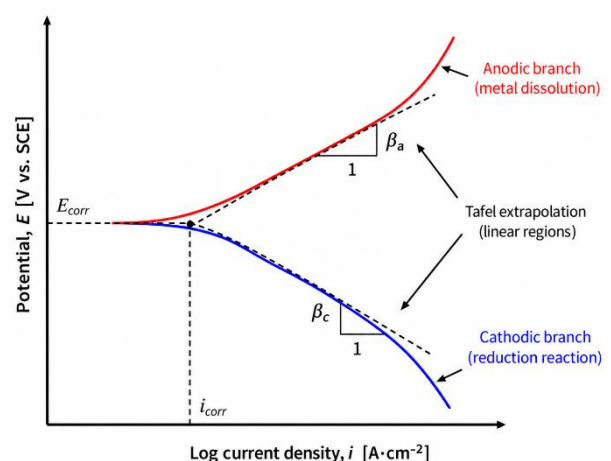


Fig. 1. Schematic representation of Tafel analysis of a potentiodynamic polarization curve, adapted based on [31]

5. RESEARCH RESULTS

Based on the hardness values and their average values presented in Table 2, the AZ91 alloy showed the highest hardness, while the lowest hardness was measured for commercially pure Mg 99.9. The AE21 alloy also achieved relatively high hardness.

Table 2. Measured hardness values with average hardness

Material	Test load/ dwell time [N/s]	Measured hardness values [HV]	Mean hardness $\pm$ SD [HV]
Mg 99.9	49.03/30	22.9 HV	23.8 $\pm$ 2.57 HV
		26.7 HV	
		21.8 HV	
AZ31	49.03/30	42.7 HV	43.7 $\pm$ 1 HV
		43.6 HV	
		44.7 HV	
AZ63	49.03/30	52.9 HV	56.4 $\pm$ 3.50 HV
		59.9 HV	
		56.4 HV	
AZ91	49.03/30	59.1 HV	61.1 $\pm$ 1.88 HV
		61.5 HV	
		62.8 HV	
AE21	49.03/30	56.9 HV	57.0 $\pm$ 0.26 HV
		57.3 HV	
		56.8 HV	

The hardness measurements showed a clear dependence on alloy composition. Table 2 shows that the AZ91 alloy recorded the highest hardness (61.1 HV), while pure magnesium showed the lowest value (23.8 HV). The results indicate that alloys containing higher amounts of aluminium exhibited higher Vickers hardness under the investigated conditions. Fig. 2 presents mean Vickers hardness values of the investigated materials. Error bars represent $\pm$ standard deviation (SD) based on three hardness measurements.

The obtained results in Table 3 indicate that the most positive corrosion potential (E_{corr}) was observed for the AZ63 sample. The corrosion potential characterizes the thermodynamics of corrosion, and more positive values correspond to a higher nobility of the metal.

The lowest value of corrosion current density (i_{corr}) was recorded for the AZ91 alloy.

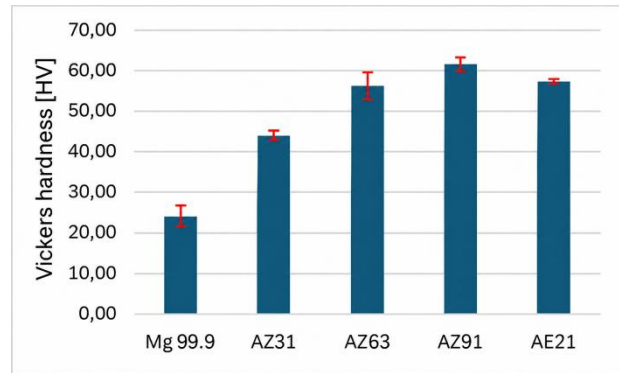


Fig. 2. Mean Vickers hardness values of the investigated materials with error bars

Table 3. Measured results from the potentiodynamic polarization

Material	E_{corr} [mV]	i_{corr} [mA/cm ²]	β_a [mV]	β_c [mV]	v_{corr} [mm/year]
Mg 99.9	-1744	12.924	142	206	0.295
AZ31	-1584	9.034	64	139	0.206
AZ63	-1496	7.6	46	174	0.17
AZ91	-1520	7.041	52	188	0.161
AE21	-1533	20.550	72	139	0.407

For clarity, Fig. 3 also presents a graphical representation of the measured results.

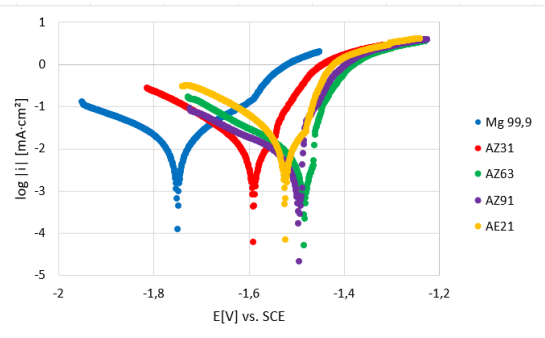


Fig. 3. Potentiodynamic polarization curves of magnesium-based materials

6. DISCUSSION RESULTS

The obtained results demonstrate that the investigated magnesium-based materials exhibited considerable differences in both Vickers hardness and electrochemical corrosion behaviour. These differences can be associated with variations in alloy composition, particularly with the presence and concentration of aluminium, zinc, and rare-earth elements. Among the investigated materials, AZ91 exhibited the highest mean Vickers hardness

and the lowest corrosion current density and calculated corrosion rate, whereas commercially pure Mg 99.9 showed the lowest hardness and AE21 exhibited the highest corrosion current density and corrosion rate. These findings indicate that alloy composition plays an important role in determining both the hardness and electrochemical response of magnesium-based materials under the investigated experimental conditions.

As shown in Table 3, the corrosion current density decreased from 12.924 mA/cm² for Mg 99.9 to 9.034 mA/cm² for AZ31, 7.600 mA/cm² for AZ63, and 7.041 mA/cm² for AZ91. In contrast, AE21 exhibited the highest corrosion current density of 20.550 mA/cm². A similar trend was observed for the calculated corrosion rate, with AZ91 exhibiting the lowest value (0.161 mm/year), while AE21 showed the highest value (0.407 mm/year).

The corrosion current density reflects corrosion kinetics. Furthermore, when evaluating the material's resistance, attention is also focused on the corrosion rate. The lowest corrosion rate was observed for the AZ91 sample.

When assessing the overall corrosion resistance using the potentiodynamic polarization method, it is necessary to consider all the above-mentioned parameters. Based on this, it can be concluded that the AZ91 sample exhibited the highest corrosion resistance. In contrast, the AE21 alloy showed the poorest performance in the potentiodynamic tests.

AZ91 exhibited a corrosion potential of -1520 mV, whereas AE21 showed a slightly more negative value of -1533 mV. The 13 mV difference indicates a slightly nobler corrosion tendency of AZ91 under the investigated conditions. However, corrosion potential alone does not quantify corrosion resistance; therefore, this result must be evaluated together with the corrosion current density.

The results confirm that alloying elements significantly influence both mechanical and corrosion properties. Aluminium and zinc contribute to increased hardness and improved corrosion resistance, while rare earth elements may reduce mechanical strength.

The electrochemical results obtained in the present study are generally consistent with previously published investigations of AZ-series magnesium alloys. Under the investigated conditions, AZ91 exhibited a lower corrosion current density and calculated corrosion rate than AZ31. Previous studies comparing AZ31 and AZ91 in physiological environments reported a similar

trend, with AZ91 generally showing lower degradation and corrosion damage than AZ31. This behaviour has been associated with differences in alloy composition, particularly the higher aluminium content of AZ91, as well as with differences in the formation and stability of corrosion-product layers. Nevertheless, direct comparison of absolute corrosion rates between individual studies should be made with caution because the measured electrochemical response of magnesium alloys is strongly affected by the composition of the testing medium, temperature, exposure time, surface condition, alloy microstructure, and experimental methodology. Regarding hardness, the investigated alloys exhibited substantially higher Vickers hardness values than commercially pure magnesium, with AZ91 showing the highest mean value. This trend is consistent with the strengthening effect of alloying in magnesium. However, direct comparison of absolute hardness values reported in different studies is difficult because hardness is strongly influenced by alloy condition, processing history, microstructure, phase distribution, and testing parameters. Therefore, the present hardness results should primarily be interpreted as a comparative assessment of the investigated materials tested under identical conditions.

7. CONCLUSION

The theoretical part of the paper focused on summarising the current knowledge on magnesium, its alloys, and the methods used to evaluate the mechanical and corrosion resistance of magnesium. The experimental part focused on comparing the Vickers hardness and electrochemical corrosion behaviour of commercially pure magnesium and selected magnesium alloys under the investigated conditions. Potentiodynamic polarization was used to assess the corrosion resistance of the samples. Based on the obtained measurement results and their analysis, the following conclusions were established:

- The presence of aluminium in the AZ91 alloy contributes to achieving higher strength of the alloy.
- The AZ91 alloy exhibited the highest hardness among all samples (61.1 HV).
- Among the investigated materials, AZ91 exhibited the most favourable combination of hardness and electrochemical corrosion resistance. However, further biological and cytotoxicity testing is required before its

suitability for biomedical applications can be confirmed. The addition of rare earth elements (REE) reduces the hardness of magnesium alloys.

- Commercially pure magnesium Mg 99.9 exhibited the lowest Vickers hardness (23.8 HV) among the investigated materials.
- AZ91 exhibited a slightly more positive corrosion potential (−1520 mV) than AE21 (−1533 mV). Together with the other electrochemical parameters, this result indicated better corrosion behaviour of AZ91 under the investigated conditions.
- Although AZ63 exhibited the most positive corrosion potential, AZ91 showed the lowest corrosion current density and corrosion rate. Therefore, the corrosion resistance of the alloys cannot be assessed solely on the basis of E_{corr} . Considering the kinetic parameters, particularly i_{corr} and the calculated corrosion rate, AZ91 exhibited the most favourable corrosion behaviour.
- It should be noted that the potentiodynamic polarization measurements were not performed as independent replicates for each material. Therefore, the electrochemical results presented in this study should be interpreted as a comparative assessment under the investigated experimental conditions rather than as statistically validated differences between the individual materials. Further studies involving repeated independent electrochemical measurements are required to confirm the reproducibility and statistical significance of the observed differences.
- Among the investigated materials, AZ91 showed the most favourable electrochemical corrosion parameters under the applied experimental conditions, particularly the lowest corrosion current density and calculated corrosion rate.

Overall, aluminium- and zinc-containing magnesium alloys showed favourable mechanical and electrochemical corrosion properties under the investigated experimental conditions. In particular, AZ91 showed the most favourable combination of hardness and corrosion behaviour. However, further degradation, hydrogen-evolution, cytocompatibility, and biological studies are required before its suitability for biomedical applications can be established. In contrast, the AE21 alloy exhibited the poorest performance in both experiments.

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CONFLICTS OF INTEREST

The authors declare no conflict of interest.

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