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Corrosion Measurement and Hardness Test of Magnesium-Tri Calcium Phosphate Composite Fabricated by Stir Casting

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Abstract

Magnesium MMC is preferred in applications requiring high-quality materials and has replaced monolithic materials. Bio-implants traditionally used are titanium, steel, and nickel-titanium alloys, however, their removal necessitates surgery, which is costly and painful. While biodegradable 3D-printed implants are effective and eliminate the need for a second surgery, they remain expensive. After several studies to solve this issue, the development of biodegradable implants is considered. Magnesium possesses good biodegradable properties. An investigation is carried out on magnesium alloys, which are more affordable and have better qualities. Although magnesium possesses biodegradable properties it corrodes rapidly in the body's environment. The need of the hour is to have a composite that degrades gradually. A magnesium-based MMC is efficient in securely and steadily degrading, eliminating the necessity for secondary surgery. The present research focuses on magnesium alloy with tricalcium phosphate as reinforcement and a stir-casting technique. The percentage of reinforcements is 0,5,10,15 of wt%. A hardness and corrosion test were performed. The results show that the corrosion rate of AZ31B+5wt% $Ca_3(PO_4)_2$, AZ31B+10wt% $Ca_3(PO_4)_2$, AZ31B+15wt% $Ca_3(PO_4)_2$ was decreased by 48.58%, 39.17% and 16.94% respectively when compared with pure magnesium. The Vickers microhardness test result shows that the hardness values of AZ31B+0wt%$Ca_3(PO_4)_2$, AZ31B+5wt%$Ca_3(PO_4)_2$, AZ31B+10wt% $Ca_3(PO_4)_2$, AZ31B+15wt% $Ca_3(PO_4)_2$ are 96.375HV, 62.725HV, 62.025HV and 66.950HV respectively

Keywords: Monolithic materials, Bio-implants,3D printed implants, biodegradable, AZ31B, tricalcium phosphate, hardness, corrosion

INTRODUCTION

The worldwide rise in cases suffering from mechanization and advancements in technology has resulted in an increased incidence of bone fractures, necessitating surgeries, and implant-based treatments. Alloys based on titanium and steel are utilized in the craniofacial area and have applications in orthopedics because of their resilience to cyclic stress. However, stress shielding is caused by the young's modulus of these materials being incompatible with the bone material. They can wear down, thus resulting in poisonous metallic ions and an immunological reaction. Degenerative and fractured bones and joints might function normally again with the help of biomaterials. Chronic illnesses are now an issue in wealthy nations. The market for orthopedic devices will develop as the number of seniors with the risk for osteoporosis, osteoarthritis, bone injuries, and obesity increases [1]. In addition to the age impact, several factors, such as sports, high-intensity work, and accidents

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that might result in bone fractures, necessitate surgery to restore biological processes using either temporary, permanent, or biodegradable prostheses [2]. Devices are to be implanted in the body environment for fracture fixation or joint replacement to repair or substitute the original ability of natural bone[3].The selection of biomaterials for orthopedic applications needs to withstand cyclic load-bearing processes during implantation[4]. The human body is a complex biochemical system, during implantation several corrosion reactions occur between the device surface and biological fluid[5].

The first generation of bio-implant materials obtained had a combination of properties with high strength and corrosion resistance to avoid the growth of the particles and release of ions from wear and corrosion processes[6]. Figure 1 shows the components of total hip replacement.

Ref: Progress Of Structural Scaffold Biomaterials for Bone Tissue Defect Repair: A Cutting-Edge Review, Composite Structure Vol 349-350 ,1 December 2024,118542

In the study by Yashar Behnamian et al[7] Stir casting was used to create ZK60 alloy matrix composites reinforced with 10-weight percent boron carbide (B4C) and 0–1 weight percent multi-walled carbon nanotubes (MWCNTs), as well as 0.5 weight percent MWCNTs and 5–20 weight percent B_4C. At room temperature, the pin-on-disk setup evaluated the samples' wear behavior under loads of 40 and 80 N at a sliding speed of 0.5 m/s. Energy dispersive X-ray spectroscopy (EDS) and scanning electron microscopy (SEM) were used to examine the produced tribo-layer and worn surface. The findings showed that samples with MWCNTs had a lower coefficient of friction at low loading than samples with varying B_4C concentrations. Increasing the MWCNTs content to 0.5 weight percent reduced the wear rates, whereas increasing them. Sachin Kumar Sharma et al [8] (2022) in his paper explained that Mg-based biocompatible alloys such as AZ31, AZ91, etc., are widely used for biomedical applications such as implant material. powder metallurgy route followed by microwave sintering is used for the fabrication of AZ31 (Mg/3wt.%Al/1wt.%Zn) composites with different reinforcing agents like CNTs (Carbon Nanotubes), Graphene, Hydroxyapatite, and TiO2 because of its high material utilization, less energy consumption, and less machining. Kuldeep Kumar Saxena et al[9] (2022) noticed that AZ31 was reinforced with various weight fractions of MWCNTs (Multi-Walled Carbon Nanotubes), which are made via powder metallurgy, sintering, and hot extrusion and are intended for use in biomedical applications. AZ31 is reinforced using MWCTs in the following weight fractions: 0.66 wt%, 1.5 wt%, and 4 wt%. Each sample's microstructure characterization was examined using X-ray diffraction (XRD) and scanning electron microscopy (SEM) pictures. During the work-study, tensile, microhardness, compression, and wear tests were used to examine the effects on mechanical characteristics, microstructure, load transfer mechanism, and wear parameters..Vinitha .M et al[10] (2024) studied fabrication and characterization of ceramic composite using the stir casting method where AZ91D Magnesium alloy and the reinforcement TCP by varying the weight fractions ranging from 5% to 15%. They studied the impact of different weight percentages of reinforcement on the particulate distribution, matrix particulate interfacial reactions, and mechanical properties, the study utilized SEM and EDS analyses, revealing uniform TCP distribution which indicates successful fabrication. Evaluation of both reinforced and unreinforced composites showcased improved mechanical properties.

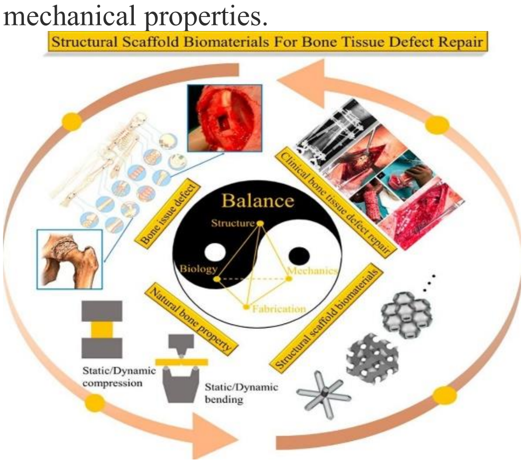

Figure 1. Structural scaffold biomaterials for bone tissue defect repair

Abdul Rahman et al [11] (2024) studied the synthesis of WE43-based alloy with calcium and Zinc. Then they reinforced it with nano reinforcements (Hydroxyapatite,beta-tricalcium phosphate, graphene, bioglass)through stir casting to create four nanocomposites. The microstructural and mechanical properties were examined and results reveal that Ca and Zn alloys and all nanocomposite materials have two additional phases compared to monolithic WE43 alloy. Adding Ca and Zn enhances the mechanical properties like ultimate tensile strength, tensile yield strength, and uniform elongation, improved compressive properties with an ultimate UCS of 370MPa.Graphene reinforcement was considered a potential biodegradable material. Mostafa M.El-sayed et al [12](2023) observed that an increase number of FSP passes caused a refinement in the grain structure of the fabricated composite and the average microhardness values in the stirred zone of the processed sample were higher than that of BM and the best improvement was achieved at four passes Mg-HA composite with about 32% higher than that of the BM and FSP caused an enhancement in the elongation as a result of ultimate tensile strength and the best achieved corrosion behavior from immersion test belong to Mg-HA composite at 4 and 2 passes FSP whose corrosion rates after seven days of immersion in SBF were decreased with about 27% and 19% respectively compared to BM.

Debao Liu et al[13] (2012) introduce new biodegradable magnesium matrix composites formulated with beta-tricalcium phosphate and magnesium matrix composite formulated with beta-tri-calcium phosphate with magnesium and zinc. A unique melt shearing technology in combination with high-pressure die casting was employed to fabricate the composite this innovative method significantly enhances the uniform distribution of beta-tricalcium phosphate particles within the magnesium matrix, reducing agglomerations, and the microstructure was analyzed. Mechanical properties were evaluated which resulted in Vickers hardness of 79, Yield strength of 125.4 MPa, Ultimate tensile strength of 150MPa, Elastic Modulus of 45.3GPa, and Elongation of 2.85% these properties indicate that beta-tricalcium phosphate composite has promising characteristics that make it suitable for the use of biodegradable implant material.

Jayavelu Sundaram et al [14] (2023) observed that reinforcing Mgo with tricalcium phosphate enhances its properties for biomedical use. Mgo composite was produced using the bottom pour stir cast method, optimizing for sample consistency at a temperature of 960^0C. Three sample composites were created with varying TCP ratios, results were analyzed for microstructure and hardness for the sample containing 90 wt% Mgo and 10 wt% TCP reveals superior properties. Magnesium alloys are known for its specific strength, damping capacity, and high strength-to-weight ratio. X-ray diffraction demonstrated that altering the microstructure through varying TCP concentrations significantly impacts mechanical properties. Although Magnesium corrodes is rapidly Mgo and Tcp are stable and are mostly used for cardiovascular stents and orthopedic implants. And it can be used in wound healing and tissue engineering. Bottom pour stir casting is identified as an effective method for fabricating homogenous Mgo-Tcp composites. This method is cost-effective for bulk manufacturing, addressing challenges regarding mixture uniformity. The paper further suggests exploration into compositional variations and microstructural modifications to improve performance. Further studies are recommended to focus on long-term biocompatibility and mechanical stability in clinical settings. Utilization of Mgo can significantly decrease the expenses associated with secondary operations. Lower costs benefit both healthcare providers and patients alike. Mgo is biodegradable, enhancing its appeal for medical applications. This property minimizes adverse bodily reactions often seen with other metals. Using Mgo in medical procedures reduces the risk associated with traditional metals. Its biocompatibility contributes to safer patient outcomes during recovery. Mgo's porous nature fosters better integration with biological tissues. This structural characteristic enhances mechanical resistance compared to other materials.

Katarzyna et al [15] (2016) reveal that magnesium alloys are considered biocompatible material suitable for orthopedic implants. They exhibit biodegradability within biological environments which is significant for medical applications. The study discusses various corrosion mechanisms affecting magnesium alloys. Understanding these mechanisms is crucial for improving their durability in biomedical applications. There is a high demand for designing magnesium alloys with controlled in vivo corrosion rates.

DESIGN CHALLENGES AND THE APPLICATION OF MG-BASED ALLOYS IN ENGINEERING AND MEDICINE

Designing alloys with the necessary mechanical properties remains a key challenge in their application. The development of metal matrix composites with magnesium alloys and bioceramic particles is explored in their work. These composites aim to enhance mechanical properties and biodegradation behaviors for bone implants and special focus is given to the integration of calcium phosphates like hydroxyapatite and tricalcium phosphate. These bioceramics serve as second phases to improve the overall performance of magnesium alloy implants. It highlights the advancements in magnesium alloys specifically tailored for biomedical use and the innovations in alloy compositions and treatments that are continuously enhancing their viability as implants. The research emphasizes the relationship between microstructure and mechanical properties of magnesium alloys. This interplay is critical for their application in load-bearing bone implants. The findings pave the way for the potential use of magnesium alloys in various medical applications and continuous research is expected to further refine the properties of these materials for improved clinical outcomes. The Corrosion of Magnesium Alloys research highlights the critical review of current methodologies for assessing corrosion in biodegradable magnesium implants. The global rise in trauma cases driven by mechanization and technological advancements has resulted in an increased incidence of bone fractures, necessitating surgeries, and implant-based treatments.

P.Kanchana et al [16] (2012) observed that magnesium-TCP (Tri Calcium Phosphate) composites are gaining attention in medicine and engineering due to their unique properties that combine magnesium's light weight and strength with the bioactivity of TCP. Magnesium-TCP applications in different fields like biomedicine in bone implants and scaffolds, dental applications, and drug delivery systems (Mg-Tcp) can be engineered to act as carriers for controlled drug release, especially in localized delivery for bone-related diseases. In the aerospace and automotive engineering field lightweight and high-strength characteristics are the integral parts achieved by Mg-TCP composites. Mg-TCP is a coating material to improve corrosion resistance in harsh environments. It is also used in renewable energy systems, such as cells or batteries where Mg-TCP acts as a corrosion-resistant component. The major advantages of Mg and TCP are biocompatibility, biodegradability, Mechanical properties, and corrosion resistance. Future research should focus more on control of degradation rate, Manufacturing Technique, and cost and scalability.

CORROSION

Major Parameter That Needs to Be Evaluated for Bio-Implants

Corrosion is a critical parameter for magnesium composites and they need to be corrosion-resistant to have a better shelf life. The corrosion behavior of magnesium-based composites is highly influenced by environmental factors such as pH, temperature, and specific exposure environment. These factors affect the degradation rate, corrosion mechanism, and overall performance of the materials. Hu H et al [17] (2014) in his research found that the role of pH has a major effect on magnesium composite as it tends to form a passivating layer of magnesium hydroxide ($Mg(OH)_2$) in an alkaline environment because of the formation of this layer the corrosion process degrades and material durability is enhanced. However, in biomedical applications, this passivation limits biodegradability. In physiological environments (for example blood plasma), magnesium corrodes moderately due to chloride ions that can break down the magnesium hydroxide layer. TCP can help mitigate this effect by buffering localized pH changes and encouraging tissue integration. Magnesium corrosion accelerates in an acidic environment due to the solubility of ($Mg(OH)_2$) and the release of hydrogen gas. This is particularly relevant in inflammation or infection sites in biomedical applications, where localized acidity can undermine the material's integrity.

Temperatures increase the corrosion rate of magnesium composites due to enhanced electrochemical activity. High temperatures may destabilize protective layers in industrial or harsh environmental conditions, increasing material degradation. Corrosion rates typically decrease at lower temperatures due to reduced reaction kinetics[18]. However, low temperatures may lead to changes in the microstructure of the composite, affecting mechanical properties and corrosion resistance. In biomedical applications, the corrosion rate of magnesium is slightly accelerated at physiological temperature compared to room temperature. Properly designed magnesium-TCP composites can mitigate this by forming bioactive layers that protect the material while supporting tissue growth.

Mg-TCP functions in various environments, including dry, alkaline, acidic, and salty. It has been noted that in saline solutions like seawater or simulated body fluids, corrosion occurs at an increased rate due to the presence of chloride ions, which penetrate and destabilizes the $Mg(OH)_2$ layer, increasing the corrosion rate. The presence of TCP will help in the reduction of corrosion by acting as a physical barrier and releasing calcium and phosphate ions that promote the formation of protective bioactive layers[19]. These protective layers can effectively inhibit the access of aggressive ions, thereby enhancing the overall durability and longevity of materials exposed to such harsh environments. Furthermore, the ability of TCP to facilitate the biomineralization process may lead to improved integration with biological tissues, making it particularly advantageous for biomedical applications. Whereas in an acidic solution $Mg(OH)_2$ layers dissolve rapidly with an increase in hydrogen gas evolution but this can be reduced by the incorporation of TCP which maintains the material's structural integrity. In alkaline and dry conditions, we have minimum corrosion. To avoid corrosion we need to have protective coatings, Alloying elements, composite optimization, and environmental control

MATERIALS AND METHODS

Magnesium AZ31B alloy is used as matrix material and Tricalcium phosphate as reinforcement, fabricated by stir-casting process. First, a steel crucible is introduced into the furnace. Then the main switch is turned ON and the furnace is operated at high power. As magnesium is highly reactive, argon gas is introduced into the furnace to surround the crucible. The Mg-AZ31B alloy is introduced into the furnace and heated until a uniform temperature of 700ºC. The melt was then cooled to 600°C to a semi-solid state. Now, 5wt% of $Ca_3(PO_4)_2$ reinforcement is added to the pre-heater and heated to a temperature of 200°C. Due to the induction principle the molten metal rotates at 600Hz frequency as shown in *Fig. 2* While the molten metal is rotating $Ca_3(PO_4)_2$ powder is introduced into the melt and uniformly mixed for about 20 minutes. After proper mixing of Mg-AZ31B alloy with $Ca_3(PO_4)_2$ powder, the crucible is taken out and the molten metal is poured into the mould as shown in *Fig. 3* After solidification the casted samples are removed as shown in *Fig. 4* The same procedure is to be followed for 10wt%, 15wt% and 0wt% of $Ca_3(PO_4)_2$ reinforcement.

After casting, the samples were machined and then samples are cut into pieces of 15mm height each to perform hardness and corrosion tests as shown in Figure5

CHARACTERIZATION

Hardness is a fundamental method for evaluating magnesium-based composites' mechanical performance and wear resistance. Hardness testing is vital in assessing magnesium-based composites' mechanical performance and wear resistance. It provides essential data that guide the development and optimization of these materials for various engineering applications. Hardness tests provide insights into its strength, durability, and ability to withstand wear by measuring a material's resistance to localized plastic deformation. samples were tested for hardness by the Vickers hardness testing method as shown in the Figure 6.

CORROSION MEASUREMENT

Optimizing the corrosion resistance and hardness of magnesium (Mg)–β-tricalcium phosphate (β-TCP) composites is crucial for their effective application in biomedical implants and structural components. Several strategies have been explored to enhance these properties for example alloying with biocompatible elements like zinc, calcium, and manganese into magnesium matrices can significantly improve corrosion resistance and mechanical properties. These elements contribute to grain refinement and the formation of stable secondary phases, enhancing the overall performance of the composite.

Figure 2. Stirring of molten metal.

Figure 3. Pouring of molten metal.

Figure 4. Samples after casting.

Figure 5. Samples after machining.

Figure 6. Vickers Hardness testing machine.

Figure 7. Samples immersed in simulated Body fluid

Figure 8. Measurement of weight loss

After the hardness test, corrosion measurements are done—the volume of solutions is calculated according to ASTM G31-72. Four beakers of 250ml capacity filled with 165ml of SBF, as shown in Figure 7, are taken. Sample weights are measured as shown in Figure 8 and tabulated in Table 2, and weight loss is noted in Table The test was carried out for 6 days, the weight loss was tabulated, and the corrosion rate was calculated.

Calculation of corrosion rate:

The corrosion rate of samples is calculated by using the formulae given below:

$$\text{Corrosion rate} = \frac{K \times \Delta M}{A \times T \times \rho}$$

Where $K = 8.76 \times 10^4$ is the conversion factor

ΔM = weight loss in grams

T = Immersion time in hours

A = Surface area of the specimen (cm^2)

ρ = Density of specimen (g/cm^3)

The density of the specimen is calculated by the formulae given below:

$$\rho(mixture) = \frac{Total\ mass}{Total\ volume}$$

Where, Total mass=mass of Mg-AZ31B + mass of reinforcement ($Ca_3(PO_4)_2$)

$$\text{Total volume} = \frac{mass\ of\ \text{Mg−AZ31B}}{density\ of\ \text{Mg−AZ31B}} + \frac{mass\ of\ \text{Ca3(PO4)2}}{density\ of\ \text{Ca3(PO4)2}}$$

$$\rho(\text{AZ31B+5wt\%}Ca_3(PO_4)_2) = \frac{1000}{\frac{950}{1.77}+\frac{50}{3.14}} = 1.8\ gm/cm^3$$

$$\rho(\text{AZ31B+10wt\%}Ca_3(PO_4)_2) = \frac{1000}{\frac{900}{1.77}+\frac{100}{3.14}} = 1.85\ gm/cm^3$$

$$\rho(\text{AZ31B+15wt\%}Ca_3(PO_4)_2) = \frac{1000}{\frac{850}{1.77}+\frac{150}{3.14}} = 1.89\ gm/cm^3$$

Sample Calculation for Corrosion Rate:

Corrosion rate of pure magnesium for 24 hours:

$$\text{Corrosion Rate} = \frac{K \times \Delta M}{A \times T \times \rho}$$

$$= \frac{8.76 \times 104 \times 0.039}{8.24 \times 24 \times 1.77}$$

$= 9.76$ mm/year

Similarly, the corrosion rate for all composites was calculated and tabulated in the results and discussions.

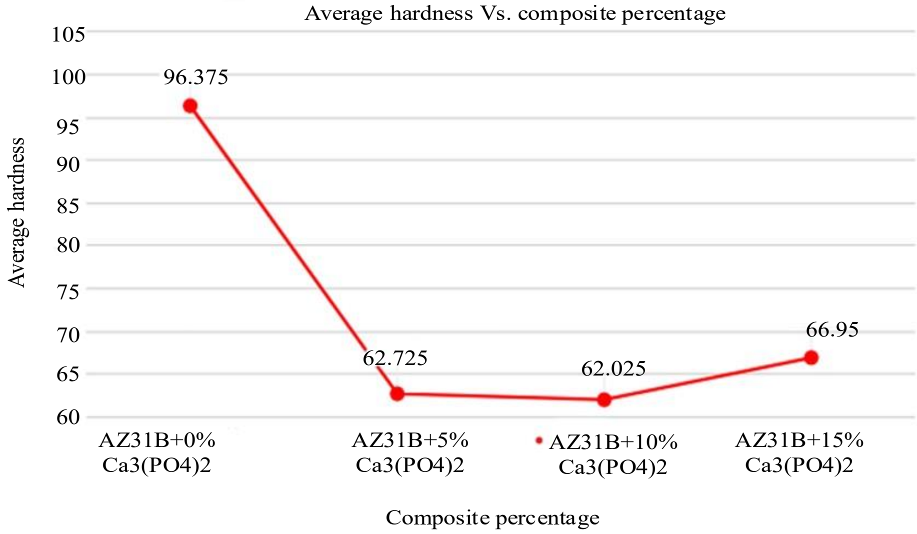

Figure 9. Average hardness Vs composite percentage.

RESULTS AND DISCUSSION

This work used the stir casting method to fabricate an Mg alloy composite reinforced with Tricalcium phosphate, four different samples with varying weight percentages of tricalcium phosphate as 0%,5%,10%, and 10% were fabricated and the modified alloy was analyzed for their microstructure and mechanical characteristics and compared to the base alloy. The key results are summed up as follows.

Hardness Test

The hardness values of AZ31B with 0%,5%,10% and 15% Tricalcium Phosphate are 96.375, 62.725,62.025,66.95 respectively as shown in Figure 9. As the graph shows the vickers hardness value is maximum at 0% magnesium and then values increases subsequently and shows an increasing trend at 15% of Tricalcium phosphate

The hardness value with 15%tri calcium phosphate was observed as highest its hardness value has increased due to the addition of a ceramic phase and it depends on many factors such as the volume fraction of the reinforcing material TCP in the composite, distribution, and bonding of TCP particles within the magnesium matrix, Processing method (powder metallurgy, casting, extrusion etc).The mechanical properties of the composite increase when the weight percentage of Tricalcium phosphate is increased.

IMMERSION CORROSION TEST

In the Immersion test, the initial weight of the composite is weighted and it is tabulated in Table 1 for different weight percentages of Tricalcium phosphate the weight loss in grams after the immersion test is tabulated in Table 2, there is a significant loss in the weight after 144 hours which is 8.86 gms. The immersion tests reveal that the weight loss was highest in 15 weight percent of Tricalcium phosphate

Table 1. The initial weight of the composite in grams.

Composite	**AZ31B+0wt % $Ca_3(PO_4)_2$**	**AZ31B+5wt% $Ca_3(PO_4)_2$**	**AZ31B+10wt % $Ca_3(PO_4)_2$**	**AZ31B+15wt% $Ca_3(PO_4)_2$**
Initial weight Ingrams	3.066	3.143	2.909	2.966

Table 2. Rate of corrosion of composites in mm/year taken at different periods.

	AZ31B+0wt% $Ca_3(PO_4)_2$(Corrosion rate in mm/year)	AZ31B+5wt% $Ca_3(PO_4)_2$ (Corrosion rate in mm/year)	AZ31B+10wt% $Ca_3(PO_4)_2$(Corrosion rate in mm/year)	AZ31B+15wt% $Ca_3(PO_4)_2$(Corrosion rate in mm/year)`
24	9.76	1.23	2.25	4.68
48	10.26	2.54	4.90	7.03
72	10.51	3.47	5.90	8.74
96	10.57	5.22	6.22	8.84
120	10.61	5.56	6.51	8.85
144	10.88	5.61	6.66	8.86

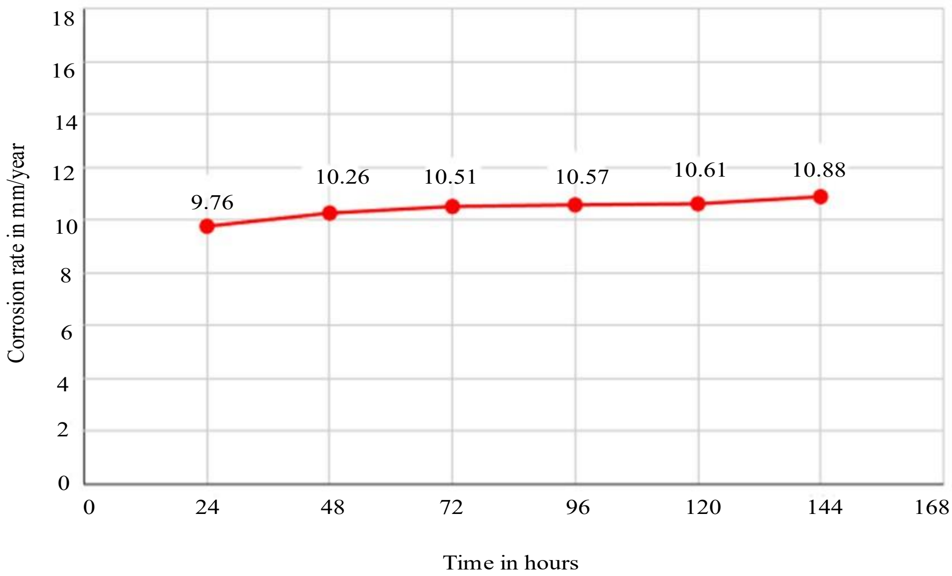

Figure 10. Corrosion rate v/s Time for Pure Magnesium (AZ31B+0wt% $Ca_3(PO_4)$.

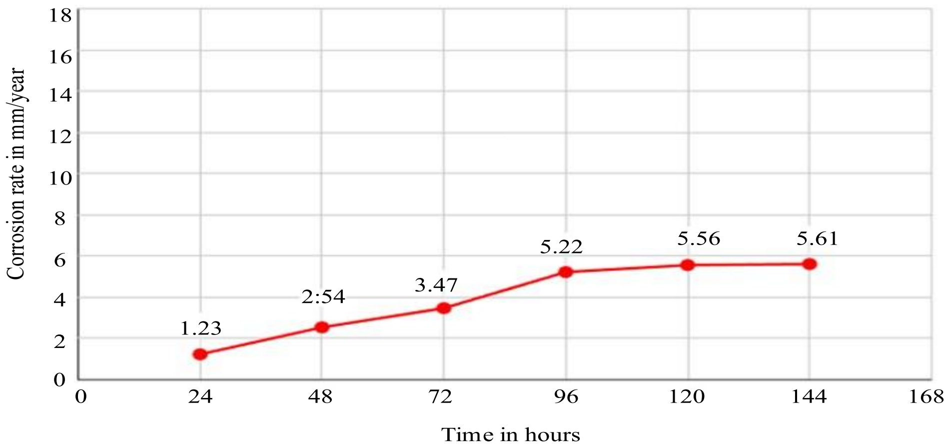

Figure 11. Corrosion rate v/s Time for AZ31B+5wt% $Ca_3(PO_4)_2$

The above graph shown in Figure 10, variation of corrosion rate of Pure Magnesium (AZ31B+0wt% $Ca_3(PO_4)_2$) with time in hours. From above graph, we can see that the corrosion rate initially increased up to 72 hours then almost remained same. Finally, the results depict that the average corrosion rate was 10.62mm/year.

The graph above shown in Figure 11 illustrates how AZ31B+5wt% $Ca_3(PO_4)_2$ corrosion rates change over time in hours. As seen in the graph above, the corrosion rate initially increased for the first 96 hours then almost remained same. The data show an average corrosion rate of 5.46mm/year.

The above graph shown in Figure 12, variation of corrosion rate of Pure Magnesium (AZ31B+10wt% $Ca_3(PO_4)_2$) with time in hours. From above graph, we can see that the corrosion rate initially increased up to 96 hours then almost remained same. Finally, the results depict that the average corrosion rate was 6.46mm/year.

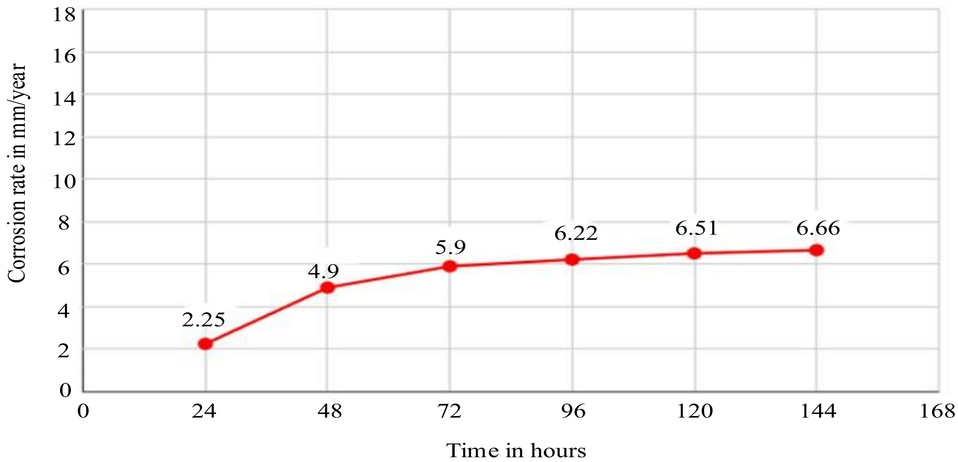

Figure 12. Corrosion rate v/s Time for AZ31B+10wt% $Ca_3(PO_4)_2$.

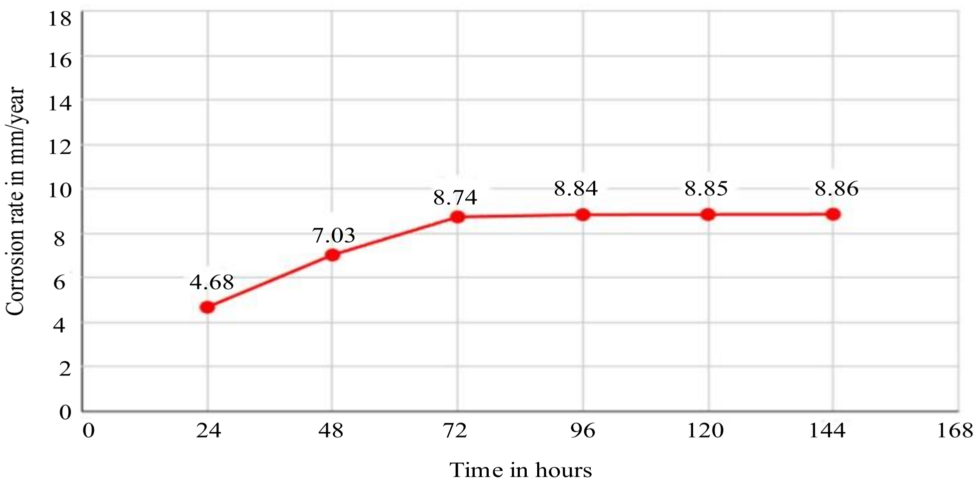

Figure 13. Corrosion rate v/s Time for AZ31B+15wt% $Ca_3(PO_4)_2$

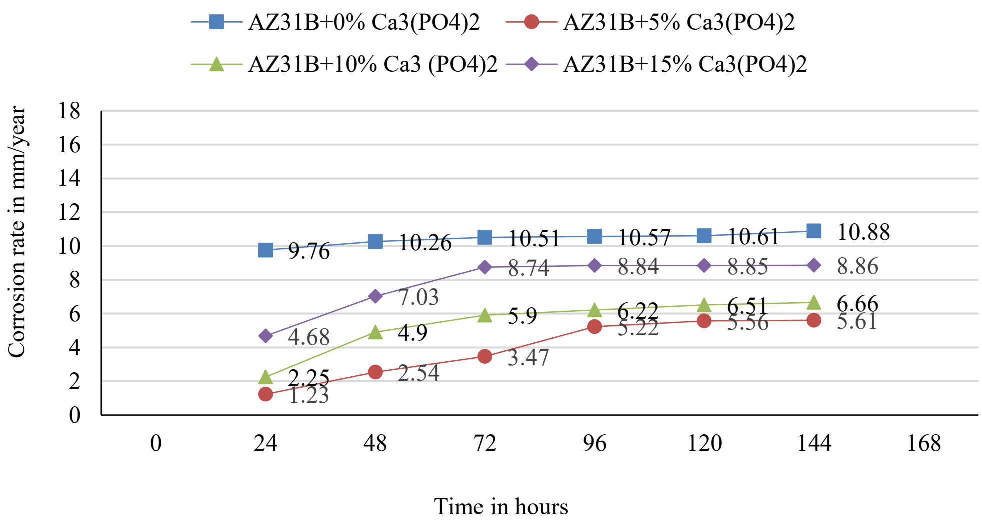

Figure 14. Corrosion rate v/s Time for AZ31B with different weight percentages (0wt%,5wt%,10wt%,15wt%) of $Ca_3(PO_4)_2$

The graph above shown in Figure 13 illustrates how AZ31B+15wt% $Ca_3(PO_4)_2$ corrosion rates change over time in hours. As seen in the graph, the corrosion rate initially increased for the first 72 hours and then almost remained the same. The data show an average corrosion rate of 8.82mm/year.

From the data Corrosion rate in mm/year v/s Time in hours graphs for different weight percentages is plotted

The above graph shown in Fig 10, portrays corrosion rate of Pure Magnesium,5,10 and 15wt percent of $Ca_3(PO_4)_2$ with time in hours. It increases up to 72 hours and then remains the same.

CONCLUSIONS

The corrosion test and hardness results show that the corrosion rate of AZ31B+5wt% $Ca_3(PO_4)_2$, AZ31B+10wt% $Ca_3(PO_4)_2$, AZ31B+15wt% $Ca_3(PO_4)_2$ was decreased by 48.58%, 39.17% and 16.94% respectively when compared with pure magnesium.It was observed that as the weight percentage of reinforcement ($Ca_3(PO_4)_2$) increased, the corrosion rate also increased. The results from the vicker's microhardness test show that the hardness value of AZ31B+0wt%$Ca_3(PO_4)_2$,AZ31B+5wt%$Ca_3(PO_4)_2$, AZ31B+10wt% $Ca_3(PO_4)_2$, AZ31B+15wt% $Ca_3(PO_4)_2$ are 96.375HV, 62.725HV, 62.025HV and 66.950HV respectively. It was observed that with the addition of the percentage of Tricalcium phosphate ($Ca_3(PO_4)_2$) to Mg-AZ31B alloy, there was a significant drop in Vicker's hardness value up to 5wt%$Ca_3(PO_4)_2$ and it remained almost the same until 10wt%$Ca_3(PO_4)_2$. At 15wt%$Ca_3(PO_4)_2$, Vicker's hardness value slightly rose from 62.025 to 66.95.The volume fraction of TCP particles within Mg matrices influences hardness and corrosion behavior. This is done by grain refinement and impeding corrosion by acting as a physical barrier. The potential future research areas to improve the corrosion resistance of magnesium-based composites are by advanced surface modification in which surfaces are treated with a protective layer to reduce corrosion, alloying with rare earth elements is one of the methods that can be implemented to incorporate rare earth elements into magnesium matrices which can enhance corrosion resistance by refining the microstructure and forming stable intermetallic compounds. Nano-scale reinforcements are a method in which nano-sized particles, such as graphene nanoplatelets or carbon nanotubes can be introduced into magnesium composite to impede corrosion. With technological advancements, we can use machine learning and computational methods that will help us discover new alloy compositions and surface treatments for enhancing corrosion resistance and improving hardness .

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