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Fabrication and Characterization of Zinc-based Alloy for Bio-implants

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Abstract

Zinc-based alloys are potential biodegradable implants and can replace Magnesium and Titanium-based materials via engineering their biomechanical properties with Zinc. Implants, due to their intricate geometries are difficult to fabricate by conventional manufacturing techniques. Therefore, they are usually fabricated by Additive manufacturing and Powder metallurgy techniques. In the present work, Zinc, Magnesium, manganese and titanium are alloyed using powder metallurgy technique. Vickers micro-hardness, UTM, Zeiss microscopy and Corrosion testing using body fluid are used for evaluation of mechanical properties. MWCNT dip coating is used to enhance the biomechanical characteristics of the prepared samples. The compressive strength of Zinc alloy is increased significantly when Magnesium, titanium and manganese are added in to the Zn based matrix. The micro-hardness of prepared samples are in the acceptable range for bio-implant application. Dip coating of MWCNT on Zn-1Mg-.8Mn shows better deposition and body fluid corrosion results marks considered as potential engineered alloy for bio implants fabrication in place of Ti-based implants.

Keywords: Zinc-based Alloys, Biomechanical characterization, ball milling powder metallurgy, nanotube dip-coating

INTRODUCTION

Zinc has recently been recognised as a potentially useful alloying element in the area of biomedicine, notably for the purpose of improving the corrosion resistance and mechanical qualities of magnesium alloys [1–3]. Zinc is a material that is regarded as a good option in the search for biodegradable substances because of its high quality and its ability to coexist with living organisms. Even though several zinc-based materials [4] with improved mechanical and corrosion characteristics have been produced and investigated, zinc in its purest form has been shown to demonstrate considerably greater cytotoxicity than magnesium, even when administered in amounts that are considered to be safe. The amount of zinc that should be consumed daily by humans is considered to be somewhere around 15 milligrammes.

Magnesium (Mg) dosages of up to 700 milligrammes (mg) per day are generally well tolerated by the body. Zinc is an important mineral that has been shown to provide nutritional benefits. It is most often found in bones and muscles, which together account for around 85 percent of the body's total zinc level. It is an essential component of macromolecules as well as enzymes, and it takes part in a diverse array of chemical processes that are catalysed by enzymes.

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There is no major danger of systemic toxicity from metallic zinc based on the fact that zinc ions are transported quickly throughout living tissue. In most cases, zinc consumption at levels more than 100 milligrammes per day is not considered harmful. Nevertheless, complete cytotoxicity testing is still required for the purpose of future assessment.

At contrast, magnesium at larger levels may activate many enzymes, control protein synthesis and muscle contraction, and stabilise DNA and RNA. Magnesium has also been shown to have anti-inflammatory effects. As a result of the beneficial effects zinc has on the human body, it shows promise as a candidate for use as a basis material in the creation of biodegradable substances. In spite of the fact that they have a lot of promise, there have only been a handful of research that look at the usage of zinc alloys in biomedical applications. There is just one paper that, to the author's knowledge, expressly studies the use of pure zinc as a bioabsorbable cardiac stent material. This is the only article that the author is aware of. In order to investigate the biocorrosion activity of zinc, this research [5] used a set of four-wire samples that were implanted in the abdominal aorta of Sprague-Dawley rats for varying amounts of time (1.5, 3, 4.5, and 6 months). According to the findings of a number of studies [6, 7], the rate of corrosion of zinc is much lower than that of magnesium and the alloys of magnesium. After around four months, zinc implants are still in good shape for the most part, but after that, the corrosion process speeds up.

The information that has been given focuses on different mechanical qualities as well as the alloying impacts that zinc-based materials have. The most important points are as follows:

1. Corrosive substances may be produced from zinc, such as zinc oxide and zinc carbonate. 1. Zinc exhibits some corrosive qualities. However, in comparison to magnesium, it corrodes at a much slower pace, which means that throughout the process of corrosion, hydrogen may be filtered out more effectively [8, 9].
2. Mechanically, zinc in its purest form has properties that are inferior to those of other metals. Because its compressive stress-strain curve does not have a section that is perfectly proportional, it is difficult to calculate the correct compressive modulus of elasticity for the material. On the other hand, estimates put the value at between 70 and 140 GPa. The Vickers hardness value of pure zinc is 30, and its tensile strength ranges from 120–150 MPa for worked zinc to around 25 MPa for cast zinc.
3. In order to increase the mechanical qualities of zinc, researchers have examined the possibility of improving it by alloying it with other elements such as magnesium, aluminium, and silver. The incorporation of aluminium results in the formation of the Al-Zn solid solution as well as lamellar microconstituents, which leads to considerable enhancements in the material's mechanical properties. For instance, increasing the amount of aluminium in a material by 5.5 weight percent may boost its tensile strength to around 308 MPa. However, the yield strength only reaches around 240 MPa at its greatest. The use of zinc-aluminum alloys in the area of medicine is restricted owing to safety concerns over the toxicity of aluminium and a reduced resistance to corrosion as compared to high-purity zinc.
4. The addition of silver to zinc alloys improves both the mechanical characteristics and the biocompatibility of the materials. The ultimate tensile strength of cast zinc alloys may be increased to 287 MPa with the incorporation of 7.0 weight percent silver. In the field of medicine, silver nanoparticles are often used for the treatment of wounds and for inhibiting the adhesion of microbes to implant surfaces. However, micro-galvanic corrosion may happen in Zn-Ag alloys because of secondary phase particles, which results in quicker degradation compared to pure zinc. This is because secondary phase particles are conductive.
5. Magnesium is an alloying element that gets a lot of attention in zinc-based biodegradable polymers because of how important it is. Because of the existence of the hard Mg_2Zn_{11} intermetallic phase, cast alloys that include 3 weight percent magnesium have the potential to produce a hardness of up to 200 HV. Depending on the cooling rate and the microstructure, cast alloys with higher magnesium concentrations (35–45 wt%) may have hardness values that

range between 285 and 300 HV1, respectively. On the other hand, eutectic phases in alloys with magnesium concentrations more than 1% by weight might have a detrimental effect on the tensile strength and lead to a reduction in that property.
6. The mechanical characteristics of zinc-based materials may also be affected by the hot extrusion process. Alloys that have been hot extruded and include 0.8 weight percent magnesium are capable of having a maximum tensile strength of close to 300 MPa and a Vickers hardness of close to 80 HV5. In a similar vein, alloys that include 1.6 weight percent magnesium and are heated via the process of extrusion may achieve an ultimate tensile strength of around 360 megapascals (MPa).

It is crucial to keep in mind that the material that has been given is a collection of a variety of studies and may not cover all of the facets that are associated with zinc-based alloys. To have a complete understanding of the mechanical characteristics and behaviour of these materials, more research and studies that are more detailed in their characterisation are required.

Alloying zinc with other elements including magnesium, aluminium, and silver has been investigated as a potential solution to improve the material's poor mechanical qualities [10–12]. The incorporation of aluminium into zinc alloys results in the formation of an Al-Zn solid solution and lamellar microconstituents, both of which serve as barriers to the motion of dislocations. This leads to a considerable improvement in the zinc alloys' mechanical characteristics. For example, the tensile strength of an alloy that contains 5.5 weight percent aluminium and is formed by hot rolling at 350 degrees Celsius is around 308 MPa, whereas the yield strength of the alloy is approximately 240 MPa. However, owing to safety concerns involving aluminium, the use of zinc-aluminum alloys in the medical field is restricted. Because of intergranular corrosion, the corrosion resistance of Zn-Al alloys is lower than that of high-purity zinc. Furthermore, the volume expansion that is linked with the development of corrosion products may lead to implant cracking and fragmentation.

Silver is another alloying metal that favourably affects the mechanical qualities and biocompatibility of zinc alloys. Silver also contributes to the biocompatibility of zinc alloys. The maximum tensile strength of cast zinc alloy may be improved to 287 MPa by adding 7.0 weight percent of silver to the alloy. The treatment of wounds and the prevention of bacterial adherence to implant surfaces are both medical applications that make use of silver nanoparticles. However, in Zn-Ag alloys, secondary phase particles function as anodes and contribute to micro-galvanic corrosion. This results in a higher deterioration rate when compared to zinc in its pure state.

In zinc-based biodegradable polymers, magnesium is an alloying element that has received a lot of attention recently. The presence of the hard Mg2Zn11 intermetallic phase in an alloy causes its hardness to rise in proportion to the amount of magnesium present in the alloy [13, 14]. A molten alloy with three weight percent of magnesium may, for instance, attain a hardness of two hundred hertz Vickers (HV). Depending on the cooling rate and subsequent microstructure, cast alloys with a higher percentage of magnesium (35–45 wt%) may obtain hardness values of 285–300 HV. In addition to that, the MgZn2 phase is present in the microstructure of the alloy, which helps to further boost its hardness. However, brittle eutectic phases in alloys with magnesium contents over 1 weight percent have a detrimental influence on the ultimate tensile strength, which decreases to 30 MPa for an alloy with 3 weight percent magnesium content. Notably, among Zn-Mg alloys, the one with the maximum elongation is the one with 1 weight percent magnesium content.

Indeed, studies have been carried out to study the effect that hot extrusion processing has on the mechanical properties of zinc alloys in a variety of applications. According to the supplied examples, magnesium-containing alloys that are subjected to hot extrusion have the potential to display higher ultimate tensile strength as well as increased Vickers hardness. For example, an alloy that contains 0.8 percent by weight of magnesium and is put through hot extrusion might potentially attain an ultimate tensile strength of roughly 300 megapascals (MPa) and a Vickers hardness of approximately 80 hertz

(HV). After hot extrusion, an alloy containing 1.6 percent by weight of magnesium may exhibit an ultimate tensile strength of around 360 MPa and a Vickers hardness of approximately 97 HV.

It is important to note, however, that not a lot of study has been done to explore the impact of preparation procedures and their parameters on the mechanical characteristics and corrosion resistance of zinc-based materials that have been created by casting or the mechanical treatment of cast objects. This is something that should be looked into further. There are a number of articles that focus on the influence that alloying elements and the content of those alloying elements have on the mechanical properties and corrosion resistance of zinc-based materials; however, additional research is required to fully understand the relationship between processing methods, parameters, and the resulting mechanical characteristics of zinc alloys.

It has been suggested that carbon nanotubes (also known as CNTs) are suitable nanoparticles to use for reinforcing purposes. However, owing to their intrinsic chemical stability, CNTs need to go through the process of functionalization in order to attain the qualities that are desired. In a study referred to as [15], researchers looked at the effect that ammonia functionalization had on the Young's modulus, bulk modulus, shear modulus, and ultimate tensile characteristics of many distinct CNT architectural configurations. In this work, the thermo-mechanical characteristics of epoxy nanocomposites were investigated. These nanocomposites were created by combining amine-intercalated graphene with multiwalled carbon nanotubes. The tensile strength saw a considerable increase of 52.8 percent with the incorporation of a hybrid filler at a weight fraction of 0.5 percent. When hybrid fillers were added at a concentration of 1.0 weight percent, the results showed that the maximum flexural strength and thermal properties were seen [16]. In addition, atomistic simulations were carried out in order to investigate the effects of ammonia functionalization on the Young's modulus, bulk modulus, shear modulus, energy storage, pull-out energy, and interlaminar strength of a number of different configurations of functionalized CNTs. Comparing unmodified CNTs to those that had been functionalized with up to ten different E-NH2 compounds was the primary focus of this study [17].

The research described centered on the preparation of zinc-based materials by applying powder metallurgy processes, especially cold pressing and cold pressing followed by sintering. A description was also given of the method of MWCNT (Multi-Walled Carbon Nanotubes) ultrasonicated dip coating that was performed on the samples that had been manufactured. It has been brought to our attention that more research needs to be published on manufacturing zinc-based powder metallurgical biomaterials using the procedures above. The purpose of this study was to evaluate the effect that the particle size of the powder materials had on the microstructure and mechanical characteristics of the zinc-based materials that were produced as a consequence. In addition, the research investigated how the preparation method influenced the outcomes of these attributes. The researchers wanted to understand how the various aspects, such as particle size and preparation method, impacted the final properties of the zinc-based materials, so they experimented with different particle sizes and tried various methods. Overall, the study aimed to fill a gap in the current body of literature by contributing to the knowledge of the link between the particle size, preparation procedure, microstructure, and mechanical characteristics of zinc-based powder metallurgy biomaterials.

MATERIALS AND METHODOLOGY

For the purpose of the experiment, spherical particles of pure zinc, magnesium, and manganese were used. The following is an overview of the characteristics of the powder materials:

1. Zinc in its purest form, with a purity of 98%, a particle size of 45 micrometers, and spherical particles as their form.
2. Pure Magnesium: Spherical particles are formed, the purity is 99%, and the particle size is 63 micrometers.
3. Pure Manganese: The purity is 99%. 45 micrometers in diameter, spherical particles make up the particles.

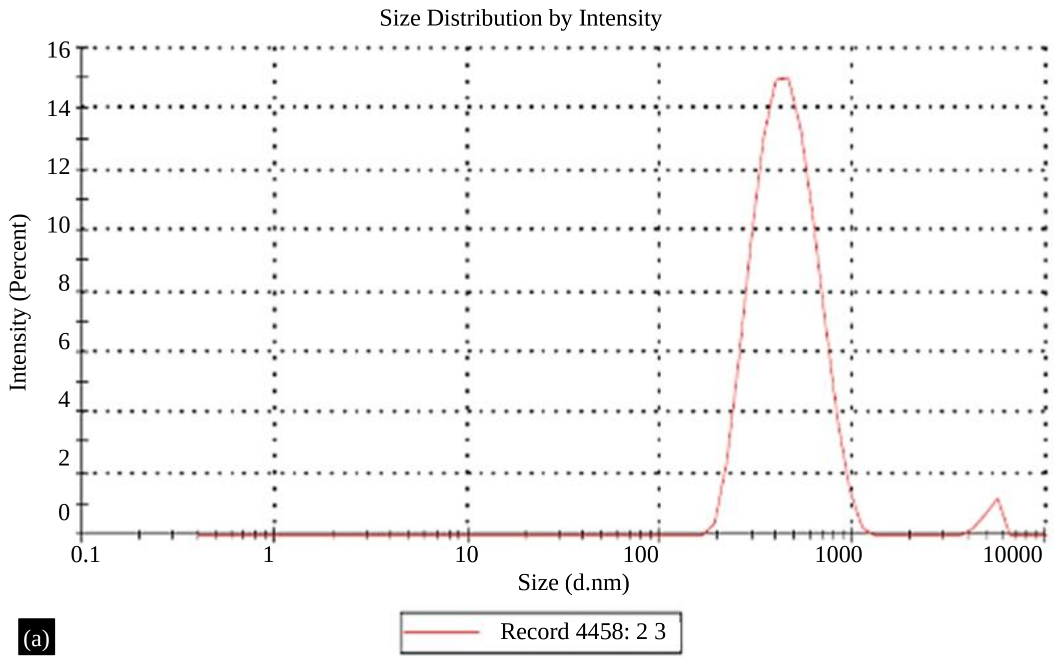
Loba Chemie Pvt Ltd, an organization with headquarters in India, was the supplier of these ingredients.

Sample Preparation

For the preliminary stage of sample preparation, a Zn-1Mg composition consisting of 99 weight percent zinc and 1 weight percent magnesium was used. A planetary ball milling operation was carried out for a period of 16 hours to guarantee that the mixture was thoroughly mixed and homogenised. Acetone was used to remove any contaminants from the balls and containers that would later be used in the milling process [18]. The wet ball milling procedure of Zn-1Mg, which used toluene as the medium, was carried out within a glove box in an environment containing an inert argon gas atmosphere. This was done to avoid oxidation. The rotation speed of the ball milling was set at 350 revolutions per minute, and the ball ratio was 15:1. The procedure included a total of sixteen cycles, each of which lasted one hour and was separated by a break of half an hour. After being subjected to ball milling, the Zn-1Mg powder samples were subjected to a FUMEHOOD machine for a period of twenty-four hours to undergo drying. According to the results of the zeta potential test, the size of the powder particles was cut down by roughly 99%, reaching 397 nanometres. Figure 1(a) shows the smaller size of Zn-1Mg after it has been decreased.

The composition of the second sample consisted of 98 weight percent zinc and 2 weight percent magnesium (Zn-2Mg). It took 16 hours of ball milling at 350 RPM with a 15:1 ball ratio. The process settings remained the same as those utilised for the initial sample preparation. Wet ball milling was carried out in the presence of an inert argon gas environment, just as it was with the preceding sample, in order to avoid oxidation. Following the ball milling operation, the particle size was proven to have been reduced by 99% to 387.9 nm, as determined by the zeta potential test. Figure 1(b) depicts the smaller size of the Zn-1Mg-.8Mn compound after reduction.

The following are the components of the third and fourth samples that were discussed: The third sample (Zn-1Mg-.8Mn) was as follows: Zinc (Zn): 98.2 wt%, Magnesium (Mg): 1 wt%, Manganese (Mn): 0.8 wt%. The composition of the fourth sample was as follows: zinc (Zn): 97.4 weight percent, magnesium (Mg): 1 weight percent, manganese (Mn): 0.8 weight percent, and titanium (Ti): 0.8 weight percent. Taking into consideration that these percentages are derived from the total weight of the ingredients and as illustrated in Table 1.



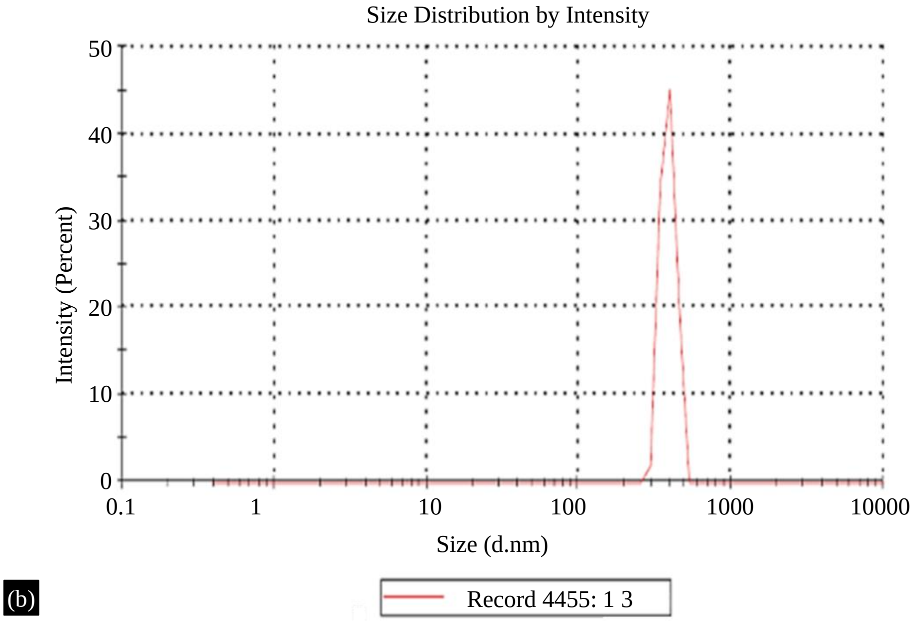


Figure 1. Size distribution of (a) Zn-1Mg and (b) Zn-1Mg-.8Mn by Zeta-potential (16 hr ball milling).

Table 1. Weight (%) composition of alloys in sample preparation

Sample No.	Zinc	Magnesium	Manganese	Titanium
Sample 1	99	1	-	-
Sample 2	98	2	-	-
Sample 3	98.2	1	0.8	-
Sample 4	97.4	1	0.8	0.8
Sample 5	100	-	-	-
Sample 6	-	100	-	-

For the purpose of cold compaction, the experimental samples were treated with the use of manual hydraulic compaction equipment. For the compacting, we employed a steel die that was hollow and cylindrical, with a diameter of 10 millimetres and a height of 70 millimetres. After being cleaned with acetone, the die was let to dry before being loaded with powder. A barrier made of graphite foil was placed between the surface of the die and the sample powder in order to prevent the powder from adhering to the surface of the die. On the hydraulic compaction machine, manual cold compaction was done at a pressure of 400 MPa for a period of five minutes. This pressure was kept constant throughout the process. With the help of this procedure, we got five representative samples out of each mixture.

A furnace found in the laboratory was used for the sintering process. The samples that had been cold compacted were then put into the furnace and heated to 375 degrees Celsius for the sintering process. In order to guarantee that the samples were heated consistently throughout their whole, they were kept at this temperature for a period of two hours.

MWCNT Dip Coating

The materials used for coating MWCNT on zinc alloy components were MWCNT, chloroform, benzene, and tri acetic acid. The MWCNT in row form was dissolved in dilute HCL solution and then subjected to a hot magnetic stirrer for 30 minutes, followed by a strained filter paper. The adhesive media was prepared using 500 mg pure MWCNT, 20 ml Benzene, 20 ml Chloroform, and 1 ml tri acetic acid solution. The solution was then subjected to 2 hours' sonication process for complete mixing. The zinc, magnesium, Zn-1Mg, Zn-1.2Mg, Zn-1Mg-.8Mn and Zn-1Mg-.8Mn-.8Ti samples were then dipped into this MWCNT solution and left in this solution for 4 hours for soaking. After soaking, the samples were kept in an induction furnace for 200 minutes at 95℃.

Density and Porosity Calculation

The density of developed specimens were calculated via Archimedes principle.

To calculate the mas of specimens i.e. W_1= 8.6408 gm is calculated for Zn-1Mg, the water volume initially be taken as V_1= 5.1 ml, and further evaluated volume via Archimedes principle when the sample was inserted into water V_2= 6.7 ml. thus we calculate the volume of the sample, $V_1= V_2-V_1$= 6.7–5.1 = 1.6 ml. Now we calculate the density of the sample i.e. $\mu= W_1/(V_2-V_1)$.

μ= 8.6408/1.62= 5.3775 gm/cm^3. (for Zn-1Mg).

μ= 8.3605/1.6= 5.437 gm/cm^3. (for Zn-1.2Mg).

μ= 8.8203/1.52 = 5.8802 gm/cm^3. (for Zn-1Mg-.8Mn)

μ= 8.9538/1.5 = 5.9692 gm/cm^3. (for Zn-1Mg-.8Mn-.8Ti)

Porosity = (particle density-bulk density)/particle density

Porosity (Zn-1Mg) = (7.07905-5.3775)/7.07905= 24.03%

Porosity (Zn-1.2Mg) = (7.0780-5.225)/7.0780= 24.61%

Porosity (Zn-1Mg-.8Mn) = (7.08032-5.8802)/7.08032= 16.95%

Porosity (Zn-1Mg-.8Mn-.8Ti) = (7.11362-5.9692)/7.11362= 16.08%

Microhardness

Vickers micro harness HV.1 of pure Zn, pure Mg Zn-1Mg, Zn-1.2Mg and Zn-1Mg-.8Mn and Zn-1Mg-.8Mn-.8Ti were measured on cylindrical billet samples having 45 mm height and 10 mm diameter. Cross-sectional planes were perpendicular to the extrusion pressure direction. To measure the micro-hardness UHL VMHT Vickers micro-hardness testing machine was used and the applied load was only 100 grams. For each sample, six readings were taken, i.e., five on the periphery and one at the centre, to get an average result.

Compression Testing

The compression testing on all six compositions has been done using the universal tensile machine with sample dimension according to ASTM C39 standard. The load applied on each sample is 2 kg uniformly throughout the cross-sections and later calculated through its 45º inclination to examine the malleability of the specimen. There were two sample testing calculation taken for each sample material.

Microscopic Investigation

The surface characterization of Zn alloys MWCNT coating is done using Zeiss microscope at 50 µm scale and 100x optical zoom ratio. The images gathered are showing proper surface coating of CNT over the Zn alloy surfaces. Some remarkable changes and surface morphologies are examined while screening the images and are elaborated in the results section.

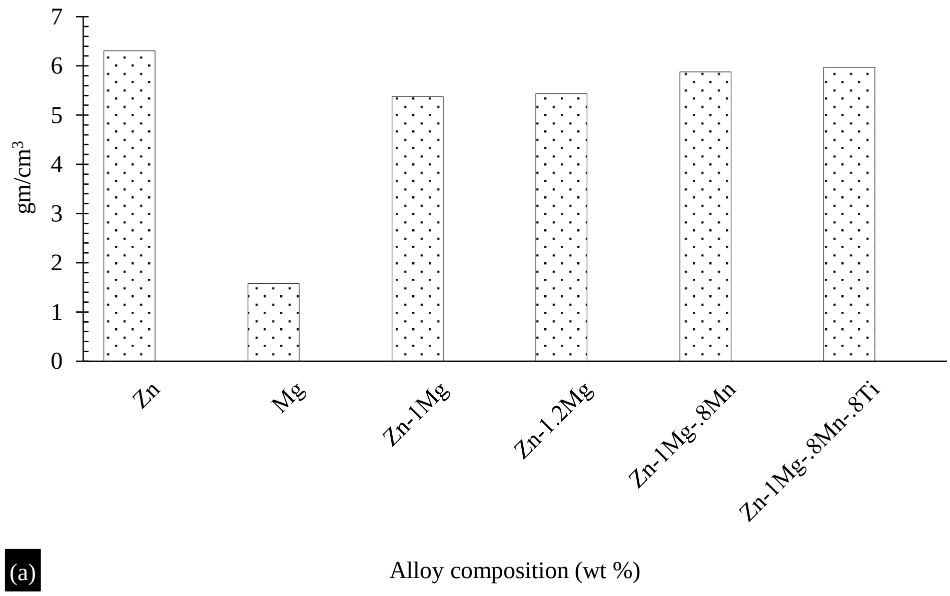
SBF Investigation

The developed samples were placed under simulated body fluid (SBF) media for 72 hours. The fluid consists of 600 ml DI water, 1M HCl, NaCl (8g), -($(CH_2OH)_3CNH_2$)- (7g) and other chemicals mixed at 36.5°C with pH 7.5. The samples were kept at 20°C in closed media to investigate its bio-interactional properties and corrosion of samples to calculate the degradation coefficient.

RESULTS AND DISCUSSION

The powder metallurgy procedure was used in order to manufacture bioresorbable zinc-based alloys that had a porosity that was precisely regulated. Cancellous bone and cortical bone are the two primary components that make up a skeleton's connective tissue. Cancellous bone, which is found in the most interior part of the skeleton, has a spongy structure and a volume porosity that ranges from fifty to ninety percent. On the other hand, the cortical bone has a volume porosity of less than ten percent and is composed of a thick outer layer. Using the powder metallurgy procedure, our goal in this experiment was to generate a Zn-1Mg (1 wt%) alloy with a targeted porosity of 24.03 percent, as well as a Zn-1Mg-.8Mn alloy with a targeted porosity of 16.95 percent.

Figure 2 (a,b) presents a graph that displays the density and porosity values of the sintered zinc-based alloy samples, together with the alloy compositions that are specific to each sample.



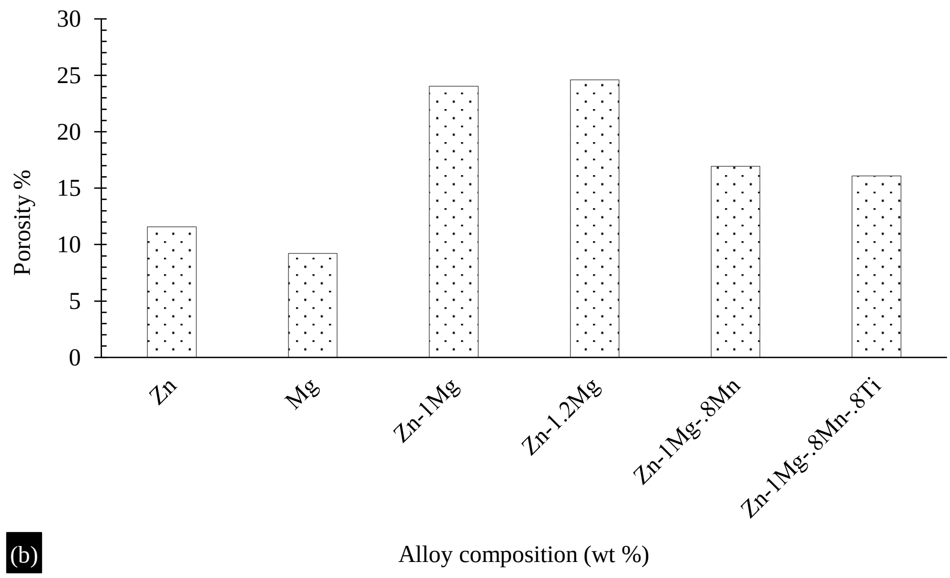


Figure 2. (a) Density and (b) Porosity of prepared samples.

The microhardness required for a cortical bone varies between 35 HV to 45 HV. The following microhardness values: Zn-1Mg has a value of 37 HV.1, Zn-1Mg-.8Mn has a value of 39 HV.1, and Zn-1Mg-.8Mn-.8Ti has a value of 42 HV. These microhardness values are within a range that is considered suitable for use in an orthopaedic implant. In this research, the graph is a comparison of the Vickers microhardness values and the standard error potential may be referred to as Figure 3.

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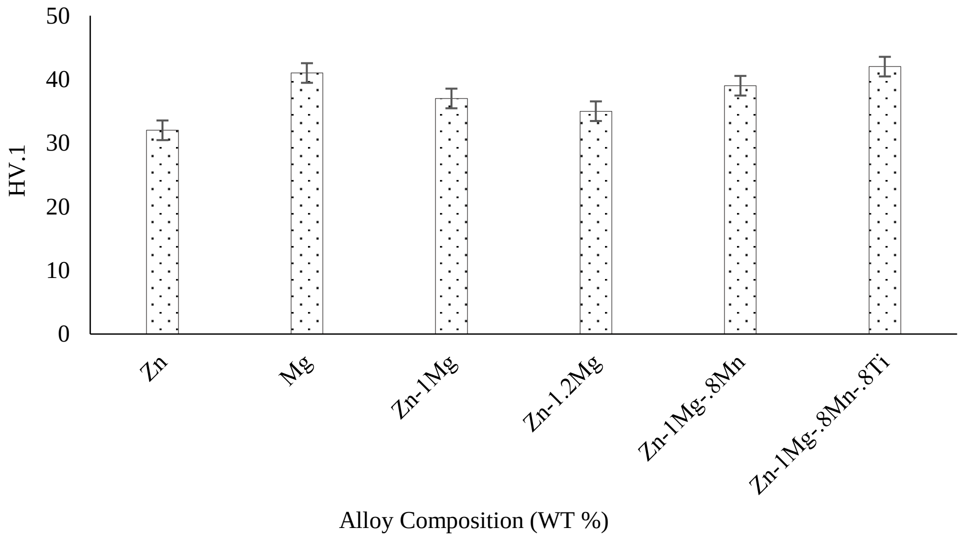


Figure 3. Micro-hardness of prepared alloys.

Cortical bone must have a compressive strength that falls somewhere in the range of 90 to 250 MPa to meet the criteria. According to the findings of compressive testing, sintered pure zinc showed a compressive strength of 96 MPa, whereas pure magnesium showed a compressive strength of 262 MPa. The compressive strength of the zinc-based alloys ranged from 177 to 205 MPa, with Zn-1Mg having a compressive strength of 177 MPa and Zn-1Mg-.8Mn having a compressive strength of 205 MPa. Notably, among the alloys that were tested, manganese (Mn) had the greatest compressive strength.

The compressive strength values of pure zinc and the zinc-based samples are shown in the graph in Figure 4, which takes into consideration an error significance of 5% in the assessment of each sample.

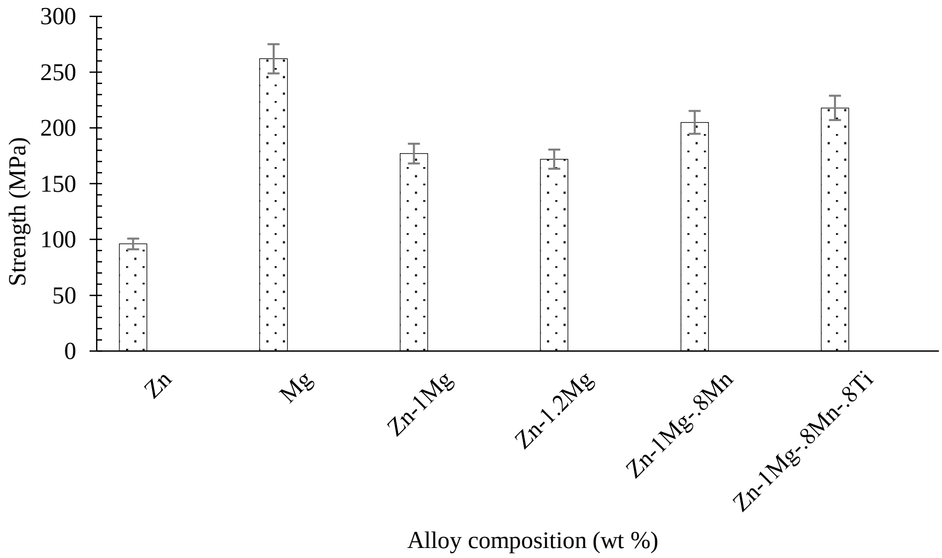


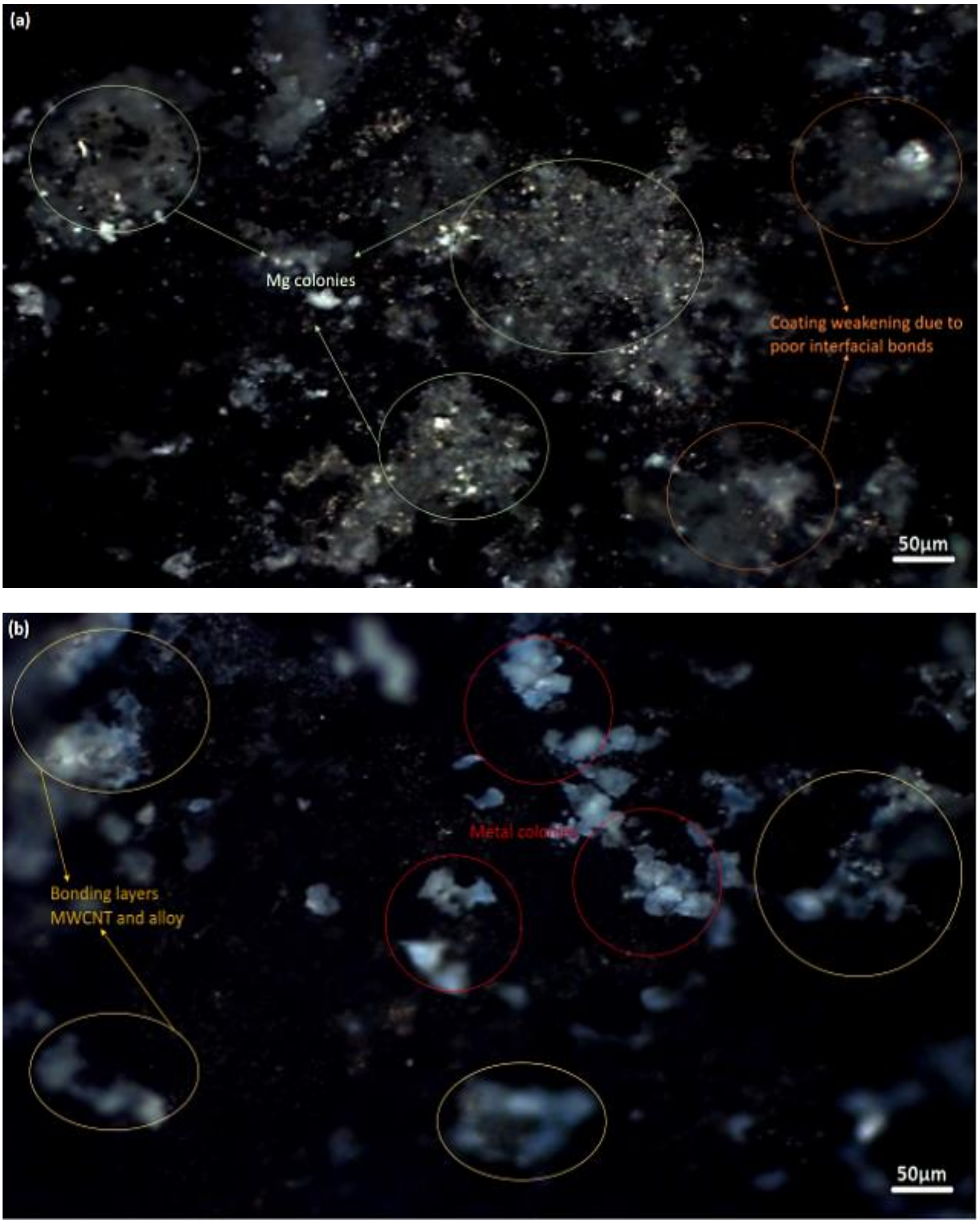
Figure 4. Compressive strength of prepared alloys.

Surface Characterization

In pure Mg samples, the MWCNT doesn't directly react with the metal material. Thus, the colonies of materials formed with intrinsic height and necking after seven weeks of coating, implying that CNT can be submitted alongside and similarly with Mg material while in interaction and doesn't substitute for being a better solution for implant applications, as shown in Figure 5(a).

Similarly, in Zn pure samples, the proper layering of CNT is done in similar conditions and possessing better structure morphologies. Here, the CNT is well meshed with the Zn sample, thus showing proper amalgamation and later possessing good material properties while implantation is accessed later in experiments, as shown in Figure 5(b) and 5 (c).

Later in Zn-1Mg-.8Mn sample represents the through layers of CNT is deposited over the surface of alloy samples and multi-layer deposition has been made. Small colonies of alloy are present, the composition will possess a greater alternative to implant application due to CNT compatibility and Mn strength with proper deposition and filler material characteristics, as shown in Figure 5(d).



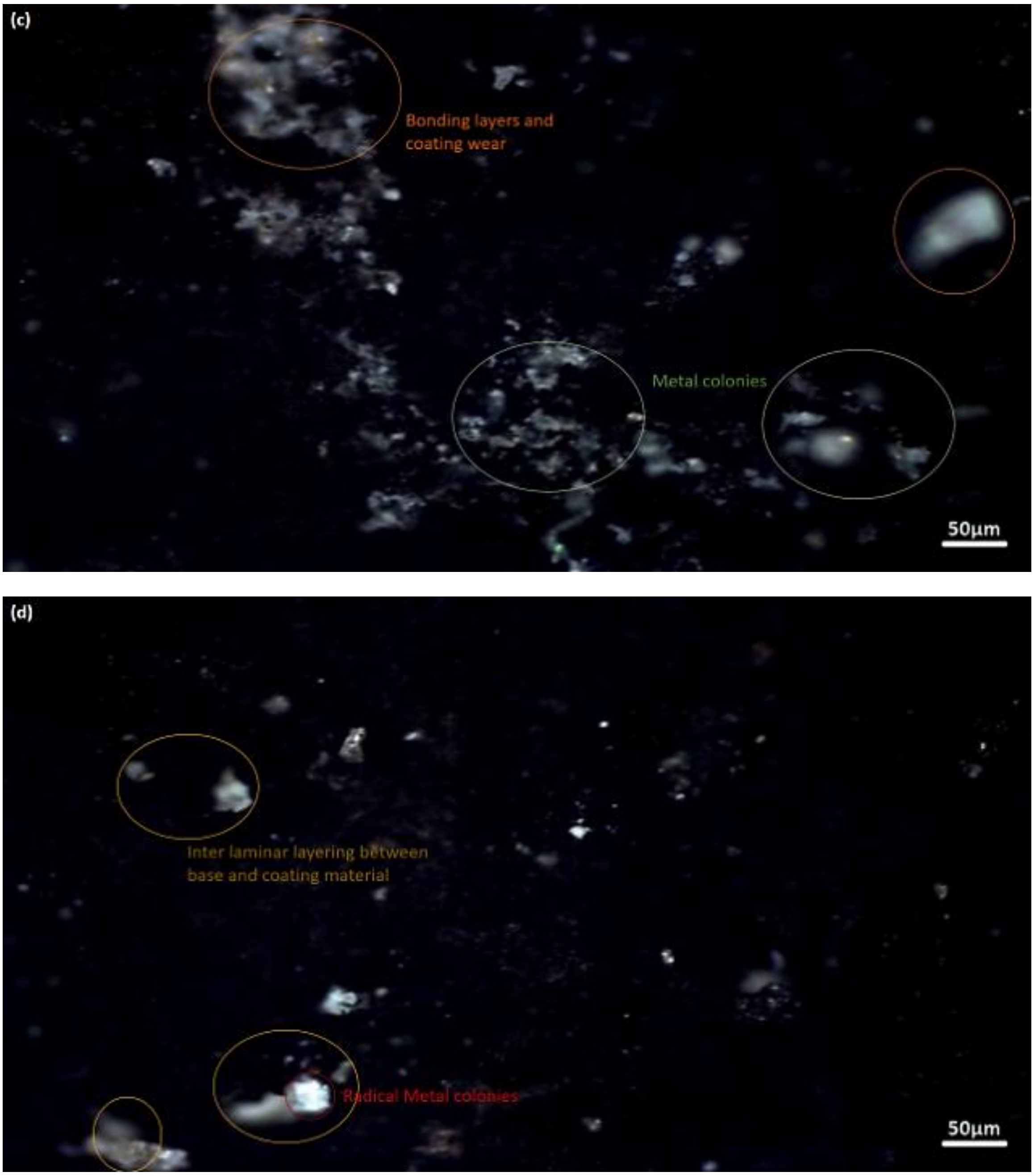


Figure 5. Surface characterization images of MWCNT coated (a) pure magnesium (b) pure zinc (c) Zn-1Mg and (d) Zn-1Mg-.8Mn sample.

Body Fluid Interaction

Immersion of all six types of Zn-CNT coated samples in simulated body fluid (SBF) for 72 hours at constant 20°C at 40 rpm constant fluid rotation speed shows some conclusive remarks for this study. The weight of Zn and Mg sample reduced from 3.32g and 2.68 gm to 3.03 gm and 2.20 gm respectively, whereas Zn-1Mg, Zn-1Mg-.8Mn samples degraded from 3.6 and 4.46 gm to 3.47 and 4.32 gm respectively as shown in Figure 6. The degradability in Zn-1Mg-.8Mn-.8Ti samples are lowered as Titanium prevents degradation and thus effecting in stress staining effects while further modified to implants [19, 20]. The addition of Manganese in samples thus providing programmable degradability and compressibility can be potential alloy material for the implant developments with comparative strength and lowered stress shielding effects.

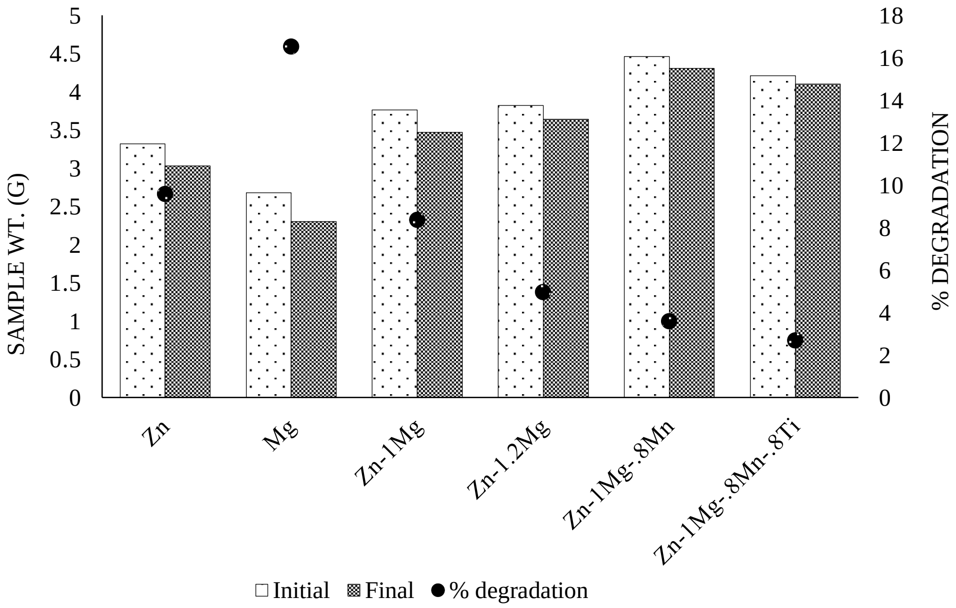


Figure 6. Alloy degradation under SBF in 72 hours.

CONCLUSION

Samples of zinc-based alloys were produced using a powder metallurgy technique, and the weight percentages of the alloying materials were varied during the manufacturing process. The objective of the study was to determine how the qualities of various alloys were affected by a variety of compositions of the alloys. The following points are concluded that:

- The reduction of alloyed powder size to 99%, also increasing the bonding strength between the particle constituents after performing 16 hours ball milling.
- The smaller particle size enhances the shrinkage volume while adding pressure during compaction.
- The enhancement in shrinkage volume and reduction in particle size, enhances the strength of specimens after going through PM techniques.
- Compressive strength of zinc improved to 177 MPa by adding Mg 1wt% and better by adding Mn and Ti (0.8% each) with the composition i.e., 215 and 218 MPa, respectively.
- The desired compressive strength, micro-hardeness required for the cortical bone will be easily achieved using compressive loading and PM techniques.
- Bio-mechanical investigation shows Zn possesses better compatibility with the MWCNT as compared to Mg.
- Zn-1Mg-.8Mn is showing proper deposition characteristics with greater potential in exhibiting mechanical and biocompatible characteristics.

Thus, the Zn based alloys shows better biocompatible results compared to Ti based materials. These combination can be used to develop implant by additive manufacturing techniques using controlled parameters and surrounding environment. The in-vitro and in-vivo analysis can be done after implant development from these materials to better determine the haemolytic properties and bio-cell interaction.

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