Corrosion and bioactivity analysis of Mg-based implant developed by SPS Technique

Chander Prakash¹*, Sunpreet Singh¹

¹School of Mechanical Engineering, Lovely Professional University, Phagwara, Punjab- 144411, India

Abstract

In the present research work, corrosion resistance and bioactivity of Mg-based alloys were investigated. The degradable Mg-based alloy for bone fixation devices was manufactured using a mechanical alloy-assisted SPS technique. The effect of MA-SPS produced alloys on the morphology, and mechanical properties was investigated. The impact of Hydroxyapatite, sintering temperature and sintering pressures was evaluated. The surface morphology, corrosion resistance, and bioactivity of SPS fabricated composites were analysed. Hydroxyapatite (HA) was obviously added to the Mg-matrix to modify the Morphology, which has resulted in the observation of coarse porous Mg with HA morphology. In Mg with HA implants, several biocompatible intermetallic phases have been produced, which are beneficial for improved corrosion and bio-activity properties. The HA presence improved the bioactivity of Mg-compact and can be used for biomedical applications.

Keywords: Mg-alloys; SPS; corrosion-resistance, Cell-culture.

1. INTRODUCTION

Due to the growing demand of artificial organ, replacement hard tissues and fixing equipment, new biomaterials have been designed and manufactured using a wide range of materials such as steel, cobalt-chrome, titanium and its alloys / composites due to its proper implant characteristics [1-10]. Nevertheless, its full potential use was not enhanced and hampered by subsequent limitations. It has been demonstrated. For example, the elastic-modulus of these materials is very large than the bone, which is responsible for stress-shielding [10-20]. This bone resorption results in a relaxation and failure of the implant [20-30]. Third, the use of biomaterials as long-lasting instruments or replacements for bone fixation. The implants were removed from the body after bone healing by an additional operation which increased the cost of the medical services and also stressed the patient [31]. Magnesium composite (Mg) gains greater appeal in the current research scenario as a promising substrate materials, as its biodegradation is near the bone, is the main biocompatibility
and is weak in elastic modulus [31-38]. Product alloy is the most effective and feasible method for controlling Mg alloy degradation rate [39-40]. A variety of M-alloys have been produced in response to their biological function in the human body [41-50]. Such alloys are exceptionally integrative in the mechanical-biological antibacterial and corrosion properties of Ca and Zn elements [51-65]. The main inorganic substances Ca and Zn are used to improve the mechanical and biological characteristics of organic insert tissue production [66-75]. Numerous researches have to date been documented in order to control the rate of degradation, improve biomechanical properties and biocompatibility in the developing of Mg-alloy using diverse production processes. In this work, the silicon-hydroxyapatite were applied to the Mg-matrix and creating porous bio-composites using the MA-SPS technique with a special goal of low modulus elastics, increased corrosion resistance and bioactivity for orthopedical bone fixing instruments.

2. MATERIAL AND METHOD

Raw material was procured and prepared using the high purity Mg, Mn, Zn, HA (approximately 99.9%) and mixed using ball-mill. The mixed powder was sintered by the SPS system and tested by various characterization techniques. Fig. 1 shows the image of experimental set-up. The corrosion-resistance was accessed using using Tafel-polarization technique. The bioactivity of the SPS fabricated alloys was assessed by cell culture analysis using MG-63 human osteoblast-like cell lines.

![Experimental set-up](image)

Fig. 1. Experimental set-up

2. RESULTS AND DISCUSSION

Figure 7 shows the SEM-micrograph and elemental-composition of the as-fabricated Mg-alloys fabricated at operating condition of $S_t$ 50 MPa and $S_p$ 450 °C. It can be evidentially seen that the as-fabricated alloy was
The tests from the SEM micrograph demonstrate that the alloys have a particular framework. The Zn alloy element existed in the Mg matrix as the secondary phase with a hexagonal, sealed structure and MgZn2 were observed as a condensed fleck agglomeration. The Mn5Si3 and Mg2Si inner metallics are often polygonal in form and at high magnification can be clearly identified. In the as-produced Mg-with-Si alloy, the related EDS range was reported to be Mg, Zn, Mn and Si components. In comparison, recorded dark and white periods. EDS classified the dark phase as a MgZn2, the gray phase as an Mg2Si. Ben-Hamu and were also found in the comparable microstructure. On the other hand, if HA was used as an alloyside rather than Si, the usual structural change was observed and needlelike MgCaO phases were formed, which eventually spread, as shown in Fig. 7 (c-d). Also, the morphology observed needle-like very interesting structures. The specific findings on HA deposition with Mg alloy to enhance its corrosion resistance have been published. The related EDS-spectrum confirmed the presence, through heating and building of oxides within the system. In the Mg matrix several found dark stages, brown phases and light phases. The dark phases have been classified as Mg series, CaMgSi gray phases and the Mg2Si process as light eutectical like a needle. These elements react with one another during sintering and form different types of oxides.
Figure 3 depicts the alloy polarization graph as produced with 40 MPa sintering pressure and a sintering temperature of 400 K in working conditions. The parameters were included for such corrosion as potential (Ecorr), current corrosion density (Icorr), resistance to polarization (Rp) and the potential for corrosion (Cv) and corrosion rate (CR). It is clear that all specimens had a common corrosion process on anodyne and cathody plopes at the plot of Tafel. The results showed that the Cv for pure-mg was approximately -1.2 mV with the relative Icorr at 19.5 μA / cm²-2 which causes the pure-mg substratum to be active.
As previously reported, the corrosion rate for the pure-Mg alloy was estimated at about 5.47 mm / year (Ref 46). The passive layer developed on pure-Mg alloy was less defensive and precarious. The corrosion capacity of the Mg-Zn-Mn-Si as produced alloy was -1.13 mV and Icorr respectively about 3.5μA / cm–2 which indicated a better resistance to corrosion and pure-Mg alloy for the as-manufactured bio-alloy. For contrast with the pure mg alloy specimens ‘ slope, better resistance to corrosion due to phases in the microstructure. Zhang and. and. al. has stated that the Mg2Si process serves as the barrier to corrosion, regulates the rate of deterioration and enhances the properties of Mg alloys respectively. The snow-covered Mg-Zn-Mn-HA alloy has measured the possible corrosion to be about-1.27 mV with a density equivalent to about 7.7 μA / cm–2. Because of the microstructural phases of CaMg and Mg0.97, the alloy Mg-Zn-Mn-HA is more prone to corrosion. CaMg phases have already been documented to have higher resistance to corrosion. The risk for corrosion in Mg-with HA&Si alloy is calculated to be about –1.17 mV, which is roughly 0.98μA / cm–2 which is more favorable and healthy than the other Mg-alloy.

![Graph showing corrosion test results](image)

**Fig. 3. Corrosion test results**

The cell viability measured by various assays is shown in Figure 12, which includes: cell numbering, MTT checking and DNA quality assessment. The number of cells in the spring alloy specimens are shown in Figure
12 (a). Cell density decreased, as the incubation time for all biocompatible organisms grew, further improving the degree of the surface's cell connection. Figure 12(b) demonstrates the cell proliferation findings for the osteoblast-like cellsMG-63 in the research strain containing samples and samples have a higher rate of cell proliferation. This is due to the high level of porosity in structure and β-TCP content, providing a hydrophilic surface and increasing surface energy and bioactivity, which promotes cell adhesion. The proliferation of cells by surface topography has previously been reported. Fig. 12(c) shows the DNA element for all alloy specimens of Mg-63 cells. The results showed that the DNA composition of all test surfaces was proportionate to the cell proliferation. Lower cell attachment rates and replication resulted in higher DNA material. Fig. 12(d) demonstrates the cell-cultivated APL template behavior on all sprung alloy cases. The ALP activity increases significantly with an increase in culture time. It can be clearly observed.
3. CONCLUSIONS

SPS partial disintegrate HA into Ca and P such as secondary phases, thereby increasing the samples bioactivity. Throughout sintering, which formed biomimetic oxide phases the pore layers, the phase increased the corrosion resistance and bioactivity. For a wide range of applications, more research can concentrate on pore size regulation and the accuracy and creation of custom architectures. In addition, clinical trials are necessary in order to meet all claims for the statistical analysis of in-vivo results.
REFERENCES


