Bio-mechanical property analysis of Mg-based implant developed by SPS Technique

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Abstract

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, elemental, phase composition, mechanical properties of SPS fabricated composites were characterized in detail. 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 such as CaMg, Mg-Zn, Mn-CaO, Mn-P, Ca-Mn-O and ZnO2 have been produced, which are beneficial for improved corrosion and bio-activity properties. The elastic 32 GPa modulus is manufactured by SPS compounds Mg-with-HA Alloy.

Keywords: Mg-alloys; SPS; Mechanical; Properties.

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-
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. The microstructure, morphology, elementary and phase composition of samples. Nanoindentation technique was employed to develop the mechanical properties of Mg-alloys. Fig. shows the image of experimental set-up.

![Fig. 1. Experimental set-up](image)

2. RESULTS AND DISCUSSION

Figure 2 shows plots and micrograms indicating architecture and mechanical properties. As noted, the porosity decreased when the application force per unit area increased. In addition, the sintering pressure was found to
make an important contribution to compact densification. High sintering temperatures, on the other hand, helped the powder particles coalesce to reduce and densify the porosity completely and subsequently. The change in the percentage of porosity and density of as-fabricated Mg-Zn-Mn-Si alloy at various levels of sintering temperature and applied pressure. It was evidentially observed that at the as-fabricated compact has average porosity of 42.45 % and density 0.80 g/cm³, respectively, at sintering pressure and temperature of 30 MPa and 350 K. The porosity decreased from 42.45 % to 21.95 %, when the applied pressure increased from to 40 MPa. The observation of high level of porosity is due to the partial sintering of compact at low level of sintering pressure and temperature. If applied pressure increased further up to 50 MPa, the porosity decreased to 12%.

Correspondingly, the density of the as-fabricated alloy increased from 0.89 g/cm³ to 1.62 g/cm³, as the sintering pressure increases from 30 to 50 MPa. The average level of porosities and density was increasing proportionally.

**Fig. 2. Effect of process parameters on Mechanical properties and architecture**
with increasing the sintering pressure. When the Mg-compact was consolidated at 50 MPa and 450 K, and a solid full densified compact was obtained. The respective density and porosity were 1.76 g/cm³ and 5.55 %. The variation in the percentage of porosity of as-fabricated Mg-with-HA alloy at various level of parameters. It was observed that the level of porosity in the compact is 60.45 % and density 0.79 g/cm³, respectively, at sintering pressure and temperature of 30 MPa and 350 K. The porosity decreased from 60.45 % to 35.56 %, when the applied pressure increased from to 40 MPa. If applied pressure increased further up to 50MPa, the porosity decreased to 20%. Correspondingly, the density of the as-fabricated alloy increased from 0.79 g/cm³ to 1.59g/cm³. When the Mg-compact was consolidated at 50 MPa and 450 K, and a solid full densified compact was obtained. The respective density and porosity was 1.79 g/cm³ and 9.25 %. While, in the case in the case of Mg-compact fabricated at 350 ºC and 30 MPa, the level of porosity is 50% and density is 0.89 g/cm³. The trend of increasing the porosity and density was similar to the previous ones. When the Mg-compact was consolidated at 50 MPa and 450 K, respectively, a full densified solid compact was obtained with density and porosity 1.79 g/cm³ and 9.25 %.

The plot loading and unloading and indenting of nanoindentation specimens are shown in Figure 3. Mg-specimens with HA discharge curve have a smaller overall penetration and slope; the value of hardness and elastic modulus are thus weak. The Mg-Zn-Mn-HA check is clear. For Mg with HA, for minimum elastic module hardness, 30 GPa (~55 HV) and 0.54 GPA (~55 HV) were measured. The Mg with Si specimen, on the other hand, has the lowest penetration and slope of the unloading curve for the Si&HA sample, and therefore has a high hardness and elastic module reliability. For Mg with Si specimens, 45 GPa and 0.72 GPa were tested in the elastic module and the maximum hardness. Once HA and Si were alloyed together into the Mg-matrix, there was less penetration and slope of the discharge curve than Mg with Si and a high penetrating and slope of the discharge curve than Mg with HA, which was produced by the Alloy Specimen. The results demonstrated the bioalloy Mg-with HA.
3. CONCLUSIONS

Three separate porous and bio-degradable structures were successfully developed by the SPS technique for bone fixing devices. The average porosity of 30% – 60%, ranging from > 50μm, was observed to be attainable at 10wt. HA and Si content percent. In addition, developed alloy structures were achieved in close proximité to nature of the bone properties, with reasonable elastic modules (29-45 GPa) and hardness (86-200HV). In the background of XRD plots, partial decomposition of HA and Si was observed during MA-SPS processing and the development of Ca and P, such as secondary stages, increased.

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