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Mechanical integrity of magnesium alloys in a physiological environment: Slow strain rate testing based study

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ABSTRACT

When used as implants, magnesium alloys will simultaneously encounter mechanical loading and corrosive physiological environment, which can cause a premature failure due to stress corrosion cracking (SCC). Therefore, it is essential to characterize the SCC behavior of magnesium alloys, before they can be actually used as implants. In the present study, the SCC behavior of a common magnesium alloy, AZ91D, and a biocompatible alloy, Mg–3 wt.% Zn–1 wt.% Ca, was evaluated in the physiological environment using slow strain rate tensile (SSRT) testing. The susceptibility of the alloys to SCC was confirmed by analyzing the fracture surfaces using scanning electron microscope. The slow strain rate tensile testing results, together with fractography, confirmed that both the magnesium alloys were susceptible to SCC in the physiological environment.

1. Introduction

Traditional metallic materials, titanium alloys, stainless steels and cobalt–chromium alloys, have been widely used as medical implant devices due to their excellent mechanical and corrosion properties [1]. However, vast differences in mechanical properties of these alloys with the human bone cause stress shielding problem [2]. Also, degradation products of these alloys may be toxic to the human body. Furthermore, when these implants are used as the temporary devices such as pins and screws, they remain as a foreign body to the human tissues even after completion of the healing process, and are generally removed by a second surgery. The second surgical procedures not only increase health care costs but also patient's morbidity. But, it will be bio-medically and commercially highly attractive if the material of the temporary implant could dissolve into non-toxic products in the physiological environment after accomplishing its functional use. In this context, magnesium is emerging as a potential candidate for its use as the biodegradable implant material. Magnesium is also attractive because its mechanical properties are similar to those of the human bone, besides its primary advantages of biodegradability and biocompatibility [3]. Magnesium ions are the fourth major cations in the human body. Magnesium is also important for many biological functions of the human body [4]. Unlike the traditional implant materials, degradation products of magnesium are not toxic to the human physiology. In fact, the magnesium ions that are produced as a result of degradation are reported to aid growth and healing of the tissues [4]. Any excess magnesium ions are harmlessly excreted through the urinal system.

In spite of the highly favorable attributes, magnesium and its alloys have found very little commercial application as orthopaedic implants. The major limitation arises because of very high corrosion rate of magnesium alloys in the

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Nomenclature

SCC	stress corrosion cracking
SSRT	slow strain rate tensile
m-SBF	*modified*-simulated body fluid
SEM	scanning electron microscopy
UTS	ultimate tensile strength
ε_f	elongation-to-failure
I	stress corrosion cracking susceptibility index

physiological environment [3,5–7]. In addition to metal loss due to corrosion, magnesium alloys also produce hydrogen gas, which can further hinder the healing process [8,9].

Implants are exposed to biochemical and dynamic environment of the human body as well as subjected to both tensile and compressive stresses during the service period. The presence of the mechanical loading along with the corrosive physiological environment may pose further complications of a sudden fracture of an implant due to stress corrosion cracking (SCC). Such sudden failures of implants in the body may necessitate troublesome removal of failed devices, and may cause painful irritation or inflammation of the surrounding tissues. In past, several instances of the fracture due to the stress corrosion cracking of the metallic implants of stainless steels and titanium alloys have been reported [10,11]. Magnesium alloys have also been reported to be susceptible to the stress corrosion cracking in a chloride environment [12–15]. Winzer et al. [16] characterized the stress corrosion cracking of Mg–Al alloys, and reported aluminum containing alloys AZ91, AZ31 and AM30 to be susceptible to the SCC in aqueous environment, and resulting fracture surfaces to have the features of transgranular SCC. Ben-Hamu et al. [12] studied the SCC of Mg–Zn–Mn wrought alloys, and reported all the investigated alloys to be susceptible to the SCC in a chloride environment. Hence, before magnesium alloys are actually used as the temporary biodegradable implant devices (e.g., stents, pins and screws), it is essential to assess the extent of corrosion-assisted degradation in their mechanical integrity.

There are only a few reports available on the SCC behavior of magnesium alloys in the physiological environment [17,18]. These studies are largely on aluminum containing AZ series magnesium alloys. But, the AZ series alloys are unlikely to be used in actual practice since their major alloying element, aluminum, is known to cause various neurological disorders such as dementia and Alzheimer's disease [19]. Accordingly, an Al-free magnesium alloy (Mg–Zn–Ca alloy) is included in this study. While Ca and Zn are both non-toxic to the human body [20], Zn addition also improves strength of magnesium alloys through the solid solution hardening [21,22]. Addition of Ca is also reported to refine the microstructure and improve strength as well as the corrosion resistance of magnesium alloys [6,23].

The present study will investigate the stress corrosion cracking behavior of AZ91D and Mg–3 wt.% Zn–1 wt.% Ca (Mg3Zn1Ca) magnesium alloys in the physiological environment using slow strain rate tensile (SSRT) testing. The alloy, AZ91D, was chosen for the comparison purpose.

2. Experimental

2.1. Test materials and environment

The alloy, AZ91D (Al: 8.8 wt.%, Zn: 0.79 wt.%, Mg: balance) was received in as-cast form, whereas the alloy, Mg–3 wt.% Zn–1 wt.% Ca (Mg3Zn1Ca) was cast by induction melting of high purity Mg, Zn and Ca under an argon atmosphere. The test medium for performing the SCC tests was *modified*-simulated body fluid (*m*-SBF) maintained at the nominal body temperature of 37 °C. The composition of m-SBF is given in Table 1 [24]. The *m*-SBF solution was buffered with 2-(4-(2-hydroxyethyl)-1-piperazinyl) ethanesulfonic acid (HEPES) to maintain a physiological pH of 7.4.

To simulate *in vitro* flow of the body fluid, a pump was used for continuously circulating the *m*-SBF through a vessel in which gauge length of the test specimen was immersed, which also helped in avoiding localized pH increase at the gauge length section.

2.2. Slow strain rate tensile (SSRT) testing

The geometry and dimensions of the tensile specimens are shown in Fig. 1, and the schematic of the experimental set-up of the SSRT rig is shown in Fig. 2. The tensile specimens (gauge diameter: 3 mm and gauge length: 25 mm) were machined from the as-cast billet. The gauge section was ground progressively on SiC papers up to 2500 grit followed by rinsing with acetone and deionised water before the SSRT testing. The specimens were pulled at two different strain rates of 1.2×10^{-7} and 5×10^{-7} s^{-1} until the fracture, while the elongation was measured using a linear variable displacement transducer (LVDT). The exposed area of the test specimen was restricted to the gauge length by using Teflon tapes to wrap the rest of the specimen, thus, maintaining the constant area for exposure to the corrosive solution.