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The effect of tensile and fluid shear stress on the in vitro degradation of magnesium alloy for stent applications

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ABSTRACT

Magnesium alloys have gained great attention as biodegradable materials for stent applications. Cardiovascular stents are continuously exposed to different types of mechanical loadings simultaneously during service, including tensile, compressive and fluid shear stress. In this study, the in vitro degradation of WE43 wires was investigated under combined effect of tensile loading and fluid shear stress and compared with that experienced an individual loading condition. For the individual mechanical loading treatment, the degradation of magnesium wires was more severely affected by tensile loading than fluid shear stress. Under tensile loading, magnesium wires showed faster increment of corrosion rates, loss of mechanical properties and localized corrosion morphology with the increasing tensile loadings. With the combined stress, smaller variation of the corrosion rates as well as the slower strength degeneration was shown with increasing stress levels, in comparison with the individual treatment of tensile loading. This study could help to understand the effect of complex stress condition on the corrosion of magnesium for the optimization of biodegradable magnesium stents.

1. Introduction

Cardiovascular diseases are known as a variety of disorders that involve the heart or blood vessels, which are considered as the leading cause of the mortality worldwide [1]. The implantation of a permanent metal stent is an effective therapeutic procedure to restore the blood flow. However, concerns have been raised about their permanently presence in blood vessels, such as late stent thrombosis, neointimal hyperplasia, and in-stent restenosis. Biodegradable stents may overcome these drawbacks, which can provide sufficient but temporary support scaffolding the blood vessels [2,3]. Magnesium based stents show great potential to be an intriguing alternative to permanent stents due to the adequate mechanical properties, good biocompatibility and biodegradability [4,5]. Actually, a CE certified magnesium based stent from Biotronik Inc. is already clinically available.

More thorough understanding of the degradation of the magnesium alloys is still needed to achieve variable degradation profile for different target applications. The degradation of magnesium alloys is generally related to the alloying elements, microstructure, precipitation distribution [6,7]. The surrounding ion composition also influences the degradation significantly. For instance, chloride ions induce porous pitting corrosion [8,9]; sulphate ions tend to accelerate corrosion to some degree during the initial stages of immersion [9]; while phosphate can increase the density of corrosion product layer and improve its resistance to chloride attack [8,10].

Mechanical loading also plays a key role in the degradation of magnesium alloys owing to their high stress corrosion sensitivity [11–13]. Stents are subject to considerable mechanical loadings during the initial deployment and contractions with heartbeats. Furthermore, the mechanical loadings, i.e. fluid shear stress, are normally higher for patients with cardiovascular diseases with the narrowed lumen [14]. Since mechanical loading is inevitable in blood vessels, recent work has focused on the degradation of magnesium alloys under stress in corrosive media, including tensile, compressive and fluid induced shear stresses. Han et al. [15] introduced the mechanical loading condition using the femoral intracondylar fractured animal model. The in vivo corrosion rate of pure magnesium was nearly 3 times faster than the in vitro condition. In the case of fluid shear stress, Wang et al. [16]

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revealed that flow-induced shear stress (0.07–0.62Pa) accelerated the overall corrosion of MgZnCa and AZ31 alloy. For AZ31 stents, the volume loss ratio (31%) at 0.056Pa was nearly twice that at 0 Pa (17%) after corrosion in DMEM. For tensile and compressive stresses, Zheng et al. [17] reported the enhanced degradation of Mg-Zn alloys under increased tensile (2–10%) and compressive deformation (8–16%) in Hank's solution. The degradation rates increased with increasing tensile or compressive strain. While Denkena et al. [18] found compressive stress (5 kN) led to 3 times enhanced corrosion rates of LAE442 alloy in 0.9% NaCl and tensile stress resulted in reduced corrosion. Our previous studies found that the magnesium corrosion was significantly enhanced under cyclic tensile-compressive loading and the corrosion rates increased with increasing loading level [19,20]. It seems that the effect of different stress types on magnesium corrosion is not consistent in several studies. And the understanding of the corrosion of magnesium alloys in the concurrence of loading and corrosive environment is still not sufficient.

In coronary arteries, due to the large motion of the blood vessel, i.e. bending and stretching, the implanted stents can experience a large variation in fluid shear stress and tensile stress simultaneously in multiple directions. While little is known about the combined effect of different mechanical conditions on degradation and strength decay of magnesium alloys. In the present work, we compared the individual and combined effects of tensile and fluid shear stress on the degradation of magnesium alloy. The corrosion rate, the associated strength degeneration and the corrosion morphologies of magnesium alloys were investigated. In addition, the magnesium wire, 0.2 mm in diameter, with the same order of magnitude as that of the stent struts (0.15 mm [5]) was adopted to give a more appropriate indication of the degradation behaviors of stents.

2. Experimental

2.1. Materials preparation

The as-drawn WE43 wires with 0.2 mm in diameter were purchased from Yangzhou Sanming Medical Supply Co.,Ltd. The chemical compositions of the magnesium wires, measured by inductively coupled plasma optical emission spectrometry (ICP-OES, PerkinElmer, Optima 5300DV). The wires were ultrasonically cleaned in acetone and ethanol for 5 min each to remove the lubricant residues on the wire surface. Specimens were further polished using a solution of 20 ml glycerol, 2 ml hydrochloride acid, 3 ml nitric acid and 5 ml acetic acid [21] for a mirror finish. Afterwards, the wires were treated with 40 wt.% HF solution for 72 h to allow the formation of fluoride conversion layer for better corrosion resistance.

2.2. Degradation

The load providing system is shown in Fig. 1, which consists of one circulating system and a tensile load providing device. The testing platform was described in detail in our previous studies [22,23]. The Dulbecco's modified Eagle's medium (DMEM) was used as the simulated body fluid with an original pH value of 7.4 ± 0.1. In this system, DMEM was driven by a peristaltic pump (Masterflex L/S; Cole-Parmer) and smoothed by liquid capacities and liquid resistances. The WE43 wire was fixed by two stainless steel grips and the right grip was connected with the weights. The values of applied tensile stress can be regulated by changing the weight. The applied fluid shear stress was maintained within the physiological range in large arteries (0.67–3.0Pa) [24] and the constant loadings was in the range of 0–4.9 N. Tests were performed in an incubator at 37 ± 0.5 °C. An average of five measurements was taken for each group. Note that, 100 ml DMEM was used in the dynamic degradation test to achieve steady fluid shear stress in the circulating system. The solution volume was 25 ml for the degradation test under tensile loading.

2.3. Characterizations

After different immersion periods, the specimens were removed from the solution, gently rinsed with distilled water and dried at room temperature. The corrosion morphologies were observed using environmental scanning electron microscopy (ESEM, Quanta 250FEG, FEI). The tensile properties of corroded magnesium wires were examined using a materials testing machine (ElectroPlus™ E10000, INSTRON). A 50 N tension sensor was chosen and measurements of the specimens were carried out at 1 mm/min. To calculate the degradation rate of magnesium, the released magnesium ion was measured by ICP-OES.

2.4. Statistical analysis

Statistical analysis was conducted with SPSS 10.0. Differences between groups were analyzed using an analysis of variance (ANOVA). All experiments were replicated five times (n = 5) and data were expressed as mean ± standard deviation (SD).

3. Results and discussion

3.1. Effect of tensile stress on degradation of magnesium alloys

The chemical compositions of the WE43 wires were shown in Table 1. The as-drawn magnesium wires showed limited ductility. The yield and tensile strength of the magnesium wires are 11.6 ± 0.2 N (369.4 ± 6.4 MPa), and the elongation is 3.1 ± 0.1%. Tensile strain induced by vessel dilation in blood vessels can vary significantly. With the presence of atherosclerosis, the extensibility of the blood vessel could decrease significantly because of the lipid deposition and/or calcification within the blood vessel wall, resulting in reduced tensile strain [25]. Additionally, there is a significant risk of stress corrosion cracking (SCC) for magnesium alloys. The SCC threshold level of magnesium alloys could be estimated to be 40–50% of the yield tensile strength [26]. Thus we applied five different constant loadings (0.196 N-4.9 N) below 50% of the yield tensile strength on the specimens.

Figs. 2–3 show the corrosion behaviors and corrosion morphologies of magnesium wires under tensile loadings in DMEM. With the treatment of 0.098 N tensile loading, the corrosion rates did not significantly increase after 7 h immersion compared to the unloading control. With the applied loadings were higher than 0.196 N, magnesium wires exhibited accelerated degradation with increasing stress levels (Fig. 2a). The corrosion rates for 4.9 N tensile loaded specimens increased by 89.7% after 1 h immersion. And this variation became remarkably larger after 7 h. The corrosion rate was 4.1 times that of control.

After different corrosion intervals, the tensile strength of the corroded specimens were measured, as shown in Fig. 2b. With the applied tensile loading was lower than 0.49 N, there was no significantly difference in the tensile strength of the specimens immersing for 1–3 h compared with that of the uncorroded control. When the applied loadings were higher than 0.98 N, significantly reduction of the tensile strength of the corroded specimens was observed at the 1st hour of corrosion. With the immersion period prolonged to 7 h, 30.0–54.7% strength decay was revealed for different tensile loading levels. In addition, there was no significantly difference in the tensile strength of the specimens with the applied loadings were lower than 0.49 N. It maybe attributed to the relative even corroded surface. The corrosion morphologies were relatively smooth at 0N (Fig. 3a) and small amount of corrosion products were precipitated on the surface at 0.196–0.49 N without out visible corrosion pits (Fig. 3b–c). In the case of 0.98N, some corrosion pits were generated (arrows in Fig. 3d) on the specimen surface. 46.5% reduction of tensile strength was seen, which was significantly lower than that of the unloaded group. Severe localized corrosion which was covered by large amount of corrosion products