



Improvement of corrosion resistance and biocompatibility of biodegradable metallic vascular stent via plasma allylamine polymerized coating

Pengkai Qi, Ying Yang, Sheng Zhao, Juan Wang, Xiangyang Li, Qiufen Tu^{*}, Zhilu Yang^{*}, Nan Huang^{*}

^a Key Lab. of Advanced Technology for Materials of Education Ministry, Southwest Jiaotong University, Chengdu 610031, China

^b The Institute of Biomaterials and Surface Engineering, School of Materials Science and Engineering, Southwest Jiaotong University, Chengdu 610031, China

ARTICLE INFO

Article history:

Received 10 September 2015

Received in revised form 1 February 2016

Accepted 9 February 2016

Available online 10 February 2016

Keywords:

Biodegradable metals

Plasma polymerization

Vascular stent

Corrosion

Biocompatibility

ABSTRACT

Plasma surface modification is one of the most effective ways not only to control the biodegradable behavior but also improve the biocompatibility of the biodegradable stents. In this work, we employed a one-step solvent-free process to construct a multifunctional plasma polymeric allylamine (PPAam) coating on biodegradable stent substrates (MgZnMn alloys and pure Fe). The PPAam surface is well characterized by XPS, GATR-FTIR, and AFM, displaying that the ultra-thin, pinhole-free polymer-like amine-rich layers with a thickness of (~250 nm) covered the substrates completely and uniformly. The results of electrochemical experiments as well as the analysis of corrosion products and surface morphologies confirmed the improved corrosion resistant properties of PPAam coated MgZnMn and Fe compared with their bare substrates. Endothelial cytocompatibility evaluation revealed that the PPAam coated MgZnMn and Fe presented good endothelial cells adhesion, spreading and proliferation properties. Also this protective coating could keep smooth with neither cracks nor webbings before or after the balloon expansion tests. The present study suggests that plasma polymerized allylamine technique might provide a promising potential platform in surface modification on biodegradable stents.

© 2016 Elsevier Ltd. All rights reserved.

1. Introduction

Cardiovascular diseases (CVDs), leading major reasons of death worldwide, are a class of diseases caused by disorders of heart and the blood vessels [1]. According to the statistics published by the world health organization in 2012, the annual number of deaths due to CVDs will increase from 17 millions in 2008–25 millions in 2030 [2]. At present, implantation of vascular stents has been utilized as a minimally invasive treatment for CVDs and becomes the most common percutaneous coronary intervention [3–5]. Advanced techniques have witnessed the development of vascular stents, from traditional bare metal stents (BMS) and drug eluting stents (DES) to endothelial progenitor cell (EPC)-capture stents and biodegradable metallic stents [6–8]. High incidences of in-stent restenosis, late stent thrombosis and neointimal proliferation caused by the uncontrollable differentiation of EPCs are still problems occurring in clinic [9–11].

The pathological evolution surrounding of the implantation site in response to a bare metal coronary artery stent is time-ordered, and it can be described into three overlapping stages: inflammation, granulation and remodeling [12–13]. Once the healing of injured vessels and

the complete endothelialization are obtained, no utility or benefit for stents has been demonstrated. In order to develop transitory stents, researchers have focused on biodegradable vascular stents used in medicine [14]. Biodegradable metals are expected to corrode gradually *in vivo*, with an appropriate host response elicited by released corrosion products, and then dissolve completely upon fulfilling the mission to assist with tissue healing [15]. Unlike polymers based degradable stents, where degraded by-products are assimilated directly into the tissue, the corrodible metallic stents are bioabsorbable that can be metabolized by the human body, and demonstrate appropriate degradation rates and modes in the human body. Taking account of the mechanical factors and degradation properties, the candidates for biodegradable stents substrates can be categorized into pure metals (such as pure iron or zinc) and alloys (Fe-Mn, Mg-Zn-Mn alloys, etc.) [16].

Iron and magnesium both possess low toxicity, and their transport mechanisms for clearance of their degradation products are relatively understood [17]. However, pure Mg does not suit the requirements for stent material from the perspectives of mechanical and corrosion properties. At present, Mg alloys with improved mechanical and corrosion properties are widely designed or optimized via surface modification technique [18–19]. Pure Fe (more than 99.5%) is the major component in biodegradable Fe stents. Because of the very low degradation rate of pure iron in physiological media, such implants are considered to undergo biological reactions similar to those permanent devices material

^{*} Corresponding authors.

E-mail addresses: zhiluyang1029@126.com (Z. Yang), huangnan1956@163.com (N. Huang).

applications, for instance 316L SS [17]. Nevertheless, there were little research focus on the surface modification of pure iron substrates. Meanwhile, there is still not enough evidence that endothelial cells covered Fe surface might accelerate the degradation rate [15]. But the rapid endothelialization of stent will protect the surface from being eroded under blood flow condition. The continuous release of ferrous ions may reduce the proliferation of smooth muscle cells (SMCs), thus may inhibit neointimal hyperplasia [20]. In spite of the excellent mechanical properties, non-toxicity and benign manufacturing performance of biodegradable metals, the major problems which limit their clinical applications are the in vivo degradation characteristics including the unsatisfactory corrosion rate, the loss of mechanical strength, the effect of degradation products on cells and tissues, and the unfriendly endothelial cellular microenvironment [21]. Therefore, numerous surface modification strategies have been carried out to functionalize biodegradable metals since these treatments might not only control corrosion behavior but also improve the biocompatibility as expected. Generally, mechanical methods (such as high-speed dry milling, ball burnishing, laser shock peening), chemical methods (such as chemical conversion coating, anodic oxidation, dip-coating) and physical methods (such as physical vapor deposition, ion implantation, plasma spraying) are the mainly methods to prepare multifunctional surfaces on the biodegradable substrates, among which the plasma surface modification is a one-step dry process without changing the bulk materials [22]. This unique advantage of plasma modification is that the surface properties and biocompatibility could be enhanced selectively while excellent bulk properties remain unchanged [23]. Comparing various common plasma techniques, plasma polymerization could be conducted to produce ultra-thin, pinhole-free polymer-like layers as well as to convert monomer functional groups (especially amine and carboxyl) onto polymerized molecule backbone served as anchoring points or improve cellular adhesive interactions [24].

Plasma polymerized allylamine coatings are widely used to introduce amine groups onto various biomaterials. It has been demonstrated to enhance the adhesion of proteins and cells, inhibit the inflammatory reactions and influence the differentiation of mesenchymal stem cells [25–27]. Also biofunctional molecules such as proteins, antibodies or DNA can be conjugated or electrostatic assembled on to construct smart bioresponsive surfaces [28–29]. Recently, the in vitro *anti*-corrosion properties of PPAam coated biomedical nickel titanium alloys are reported [30]. Therefore, in this study, plasma polymerized allylamine on the biodegradable metals (MgZnMn and pure Fe) is investigated as a platform coating for their corrosion resistant properties. To investigate the potential application of PPAam on biodegradable stents, endothelial cytocompatibility evaluation and balloon expansion tests are also presented.

2. Materials and methods

2.1. Chemicals and reagents.

The melted and extruded to rod shape magnesium alloy with 1 wt.% zinc and 0.2 wt.% manganese (MgZnMn) and commercial extruded pure iron (Fe) (99.9%) were used as substrates with a diameter of 10 mm and a height of 1.5 mm. Allylamine monomer (purity $\geq 99\%$) was purchased from Best-Reagent Co. (Chengdu China). All the other reagents used in the experiments are of the analytical grade.

2.2. Preparation of plasma polymerized films.

Before conducting the deposition of PPAam coating, the MgZnMn and Fe specimens were ground with SiC paper progressively up to 2000 grits and then mechanically polished with 1 μm diamond cloth. Afterwards, they were ultrasonically cleaned with ethanol for 10 min and dried at nitrogen atmosphere. Prior to the plasma polymerization, 15 min of argon (Ar) plasma sputtering was used to clean the vacuum chamber and the sample surface. The plasma polymerization was performed under pulse wave with the duty cycle of 40% ($t_{\text{on}} = 20$ ms,

$t_{\text{off}} = 30$ ms). As organic precursor, the allylamine monomer was allowed to flow through a needle valve to reach the plasma chamber, providing a total pressure of 6 Pa which composed of 2.5 Pa argon as a carrier gas and 3.5 Pa of allylamine vapor. The input power was 30 W, and the negative bias voltage was 80 V. The time of deposition was 4 h, and then the samples were immediately suffered thermal treatment at 120 °C for 1 h in the vacuum environment of 1.0×10^{-4} Pa.

2.3. Surface characterization of PPAam coated MgZnMn and Fe samples

Step profilometer (Ambios XP-2, Ambios, USA) was used to detect the thickness of the coatings. An atom force microscope (AFM, Asylum MFP-3D-BioAFM, Asylum Research, USA) was used to evaluate the surface morphology with the scanning range of $90 \mu\text{m} \times 90 \mu\text{m}$ of the PPAam coatings. AFM images were collected in non-contact mode operation equipped with Si cantilevers with the scanning range of $90 \mu\text{m} \times 90 \mu\text{m}$. The root-mean-square roughness (RMS) was calculated by the Asylum Research software. Infrared absorption spectra of PPAam were obtained using Grazing incidence attenuated total reflection Fourier transform infrared spectroscopy (GATR-FTIR) (Nicolet model 5700) in the range of $4000\text{--}400 \text{ cm}^{-1}$ in order to analyze the chemical structure of the specimens. X-ray photoelectron spectroscopy (XPS, Perkin Elmer 16PC) was applied to determine the element composition of the PPAam coating and unmodified samples. The instrument was equipped with a monochromatic Al K α (1486.6 eV photons) X-ray source operated at 12 kV \times 15 mA at a pressure of 2×10^{-7} Pa.

2.4. Electrochemical tests and analysis of corrosion products.

Electrochemical corrosion tests were conducted on a typical three-electrode cell electrochemical Workstation (IM6, Zahner, Germany), i.e. Specimen acted as a working electrode, a platinum sheet as a counter electrode and a saturated calomel electrode (SCE) as a reference electrode. The testing electrolyte was phosphorus buffer solution (PBS 8.0 g/L NaCl, 0.2 g/L KCl, 1.15 g/L $\text{Na}_2\text{HPO}_4 \cdot 7\text{H}_2\text{O}$, 0.2 g/L KH_2PO_4) in a water bath at 37 ± 0.5 °C. All the samples were connected to a copper wire and then mounted in epoxy resin with an exposed area of 0.785 cm^2 as the working electrode. Potentiodynamic polarization curves were scanned from -2 V to -1 V at a scan rate of $1 \text{ mV} \cdot \text{s}^{-1}$. The natural corrosion current (I_{corr}) and natural corrosion potential (E_{corr}) were determined by the Tafel method. The corrosion rate, CR, in millimeter per year can be determined from Eq. (1)

$$\text{CR} = 3.27 \times 10^{-3} \cdot \frac{i_{\text{corr}} \cdot \text{EW}}{\rho} \quad (1)$$

where EW is the equivalent weight of the corroding species in grams and ρ is the density of the corroding material in g/cm^3 . The electrochemical impedance spectroscopy (EIS) measurement was done in the same set-up, with scanning frequency ranging from 10^5 to 10^{-2} Hz, by using a single AC mode of amplitude of 10 mV. EIS spectra were shown by Nyquist plots, and ZsimDemo software was used to analyze and quantitatively simulate the impedance data using corresponding equivalent circuits. The corroded samples were kept under the vacuum condition until for the analysis of surface morphology and the corrosion products by optical microscope (OP), the scanning electron microscope (SEM, Quanta 200, FEI, Holland), and energy dispersive X-ray (EDX).

2.5. Human umbilical vein endothelial cells (ECs) responses in direct culture with biodegradable coated metals

Human umbilical vein was occluded after filling with 0.1% collagenase II (Invitrogen Corporation). After incubation for 15 min at 37 °C, digestion was stopped by adding F12 medium (Hyclone Company) with 15% fetal bovine serum (FBS, Hyclone Company). ECs were collected by centrifugation of the digested cell suspension. Harvested primary

Download English Version:

<https://daneshyari.com/en/article/7218584>

Download Persian Version:

<https://daneshyari.com/article/7218584>

[Daneshyari.com](https://daneshyari.com)