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Effect of a plasmaelectrolytic coating on the strength retention of in vivo and in vitro degraded magnesium implants ☆

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ABSTRACT

The strength decrease in magnesium implants was studied in vitro and in vivo, with and without a protective plasmaelectrolytic coating. In vivo, degradation was examined by implanting rectangular plates on top of the nasal bone of miniature pigs. The presence of gas pockets in the soft tissue surrounding the implants was evaluated with intermediate X-rays and computed X-ray tomography scans before euthanasia. After 12 and 24 weeks of in vivo degradation, the large rectangular plates were removed and mechanically tested in three-point bending. In vitro, identical plates were immersed in simulated body fluid for 4, 8 and 12 weeks. In vitro and in vivo results showed that onset of gas release can be delayed by the plasmaelectrolytic coating. Mass loss and strength retention during in vivo degradation is about four times slower than during in vitro degradation for the chosen test conditions. Despite the slow degradation of the investigated WE43 alloy, the occurrence of gas pockets could not be completely avoided. Nevertheless, uniformity of degradation and reliable strength retention make this alloy a prime candidate for the use of magnesium in cranio-maxillofacial surgery.

1. Introduction

Magnesium was used for the first time as a degradable metal implant for osteosynthesis a century ago [1]. After an initial drawback due to an inadequate combination of plates and screws, a number of children and adults were successfully treated by several surgeons [1–3]. These and other historical uses of metallic magnesium implants in medicine were reviewed recently [4].

During the degradation of metallic magnesium, hydrogen gas and magnesium hydroxide are formed by the corrosion reaction [5]. If the amount of gas released surpasses the absorption and diffusion capacity of the surrounding tissue, gas bubbles will form, and these can be seen on X-rays [2,6]. The bare metal surface causes an increased release of gas immediately after implantation. As soon as the surface is covered with degradation products, the gas release rate stabilizes and might be low enough to allow sufficient gas transport. It is hypothesized that the application of a coating may avoid the initial high gas release and the formation of gas bubbles [7,8]. In addition, an adequate coating [9] should be effective in avoiding premature failure of loaded implants due to stress corrosion cracking and corrosion fatigue.

As a coating of a resorbable biomaterial should not completely prevent the degradation process, the corrosion characteristic of the base material is also very important. Rare-earth-containing magnesium alloys are among the prime candidates for the successful application of magnesium in orthopedic and trauma surgery. Relatively high concentrations of rare earth elements like neodymium can be tolerated by various cell types [10]. The use of alloy systems without aluminum has also been recommended [11].

WE43 alloy was shown to exhibit a significantly increased push-out force of bi-cortical pins in growing rats [12,13] compared to Ti and polylactic acid. However, the in vivo corrosion conditions in a small animal model may not simulate the human situation adequately, as the blood flow, water and fat content of the tissue can be quite different [11,14].

A resorbable implant under load should degrade in a controlled manner and present strength retention characteristics suited to the indication. So far, only a few strength retention measurements have been published [15] on small animals, while none are known to have been performed on large animals.

The nasal bone of the miniature pig was identified as an appropriate site for in vivo degradation of large magnesium plates. It presents a straight area more than 6 cm in length, and is covered by sufficient soft tissue to avoid excessive disturbance of the animal. A periostal soft tissue pocket can easily be made to accomodate elongated implants. To our knowledge, this is the first study where long-term in vitro immersion tests (up to 12 weeks)

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are compared to in vivo degraded implants. It is also the first study to use implants in large animals for strength retention measurements.

The following research questions were addressed:

- Is the coating used effective in delaying the initial gas release?
- How fast does the strength of the degrading magnesium implants decrease?
- Can immersion testing in simulated body fluid (SBF) be used to simulate in vivo degradation?

2. Materials and methods

2.1. Alloy and coating

The magnesium alloy used is based on the ASTM B80 standard for WE43 (chemical composition: Mg–Y–Nd–heavy rare earths). The modified alloy version, with a lower impurity level, was developed and manufactured by Magnesium Elektron (Swinton, UK). Implants from a single lot were used for all experiments (lot MI0018B, T5 heat treated, 6.4 × 19 mm extrusion profile). Rectangular plates of 60 × 6.0 × 1.50 mm were machined dry using hard metal tools. All edges were rounded with a radius of 0.5 mm.

A total of 36 plates were tested, half of the plates without a coating and the other half with a plasmaelectrolytic coating from AHC (Kerpen, Germany) [19]. A standard Magoxid™ electrolyte was used and a direct current of 1.4 A dm^{-2} for up to 400 V was applied to generate the coating. Non-coated plates initially weighed 940 ± 5 mg. The Magoxid™ coating had a typical thickness of 10 μm and accounted for around 15 mg of additional mass. The total surface of a plate was 9 cm^2. The plates were cleaned with ultrasound in 98% ethanol, dried in air, packaged in pairs in a double vacuum pouch and γ-sterilized with a dose of 25–30 kGy.

2.2. In vitro immersion testing

Each sample was tested inside a separate immersion unit containing 250 ml of SBF, as described below. An immersion unit consisted of a graduated glass cylinder with 25 mm inner diameter and 240 mm length and a 250 ml plastic bottle (see Fig. 1). The magnesium sample was put inside the glass cylinder, which was then filled with SBF. The plastic bottle then inverted over the glass cylinder. The cylinder/bottle assembly was quickly tilted to prevent the liquid from flowing out and the remaining SBF was poured into the gap between the bottle and the glass cylinder. Finally, the lid of the bottle – which had a 33 mm hole – was slid over the glass cylinder to fix the assembly. The bottles were then placed inside a tempered water bath at 37 °C.

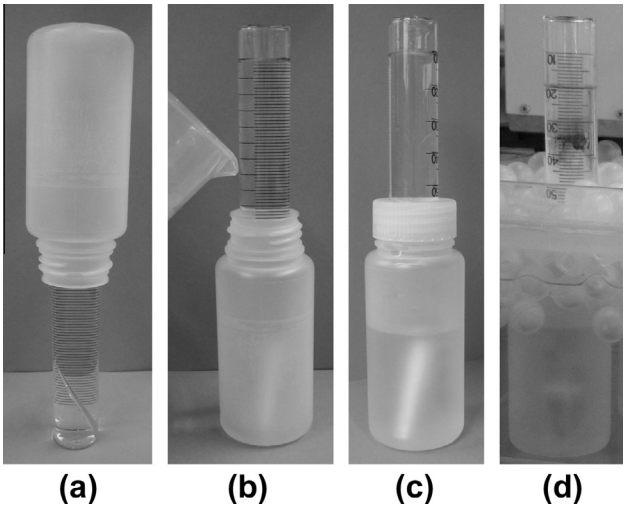

Fig. 1. In vitro testing of strength retention plates. (a) The magnesium plate is put inside a graded glass cylinder, filled with simulated body fluid (SBF) and topped with an empty plastic bottle. (b) The assembly is tilted upside down and the remaining SBF is added. (c) The immersion unit is closed with a lid. (d) The assembly is put inside the water bath at 37 °C.

The SBF was prepared from stock solutions, as described by Müller and Müller [16], with TRIS buffer and an HCO_3^- content of 27 mmol l^{-1}. The addition of NaN_3 was omitted as no bacterial growth was observed and N_2 release into the medium could be avoided. The medium was changed once a week. Identical material lots, coatings and geometries were used for the in vitro and in vivo degradation tests. The samples were immersed for 4, 8 and 12 weeks. The gas release was determined by regular visual inspection of the graded glass cylinders with a precision of about ±1 ml. The average mass loss was determined at the end of the immersion period by brushing off the corrosion products with a common nail brush.

2.3. In vivo degradation

All animal experiments were conducted in accordance with the Swiss animal protection law (Ethics committee permission #50/09). Fourteen skeletally mature miniature pigs, aged 30–36 months and with an average weight of 53 ± 7 kg, were used in this study. The treatment groups and number of animals per time point are shown in Table 1.

The midface of the minipig is approached by a T-type incision. A vertical cut 11–12 cm in length was made in cranial direction starting about 2 cm below the lower eyelids. After exposing the frontal bone, a rasp was used to create a soft tissue pocket that was big enough to accommodate the two rectangular plates and deep enough to profit of the straight portion of the nasal bone. Pre-bending of the plates could therefore be avoided. Two plates per animal were positioned on top of the nasal bone, as shown in Fig. 2a, and fixed with non-resorbable sutures.

In addition to the post-operative X-rays of the head, intermediate radiographs (Philips BV-Pulsera) were taken at 1, 4, 8 and 12 weeks.

The animals were sacrificed after 12 or 24 weeks. After euthanasia, computed X-ray tomography (CT) was performed. A medial incision about 10 cm in length was made along the longitudinal axis of the nose and the implants were removed (Fig. 2b). The pH of the implant bed was determined using a pH-sensitive strip (Merck 1.09557.0003, pH range 6.4–8.0). The strip was moistened with distilled water before use to facilitate wetting and to improve contact with the tissue.

The removed plates were stored in 70% ethanol in a tightly sealed glass bottle. After transportation to the mechanical testing site, the magnesium plates were removed from the glass bottles, dabbed with a paper towel and dried in air.

The weight of the explanted plates was determined after temporary storage in ethanol and drying, and after brushing off the degradation products with a nail brush. Additionally, the plates were immersed in 40% hydrofluoric acid for at least 5 min, as described in Ref. [15], cleaned in distilled water and ethanol, and dried with an air blower. The latter treatment was performed in or-

Table 1
Treatment groups and number of animals per time point.

Time point (week)	Group		
	Titanium	Uncoated magnesium	Coated magnesium
12	1	3	3
24	1	3	3