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An electrodeposition method of calcium phosphate coatings on titanium alloy

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Abstract Calcium phosphates coatings were deposited onto titanium alloy discs via en electrodeposition method. Titanium alloy discs were blasted with calcium phosphate particles, then etched in a mixture of nitric and fluoric acids and rinsed in demineralized water. The titanium alloy disc (cathode) and platinum mesh (anode) were immersed in a supersaturated calcium phosphate electrolyte buffered at pH 7.4 and connected to a current generator. The microstructure, chemical composition and crystallinity of the electrodeposited coatings were studied as function of time 10–120 min, temperature 25–80°C, current density 8–120 mA/cm^2, magnesium and hydrogen carbonate amounts (0.1–1 mM). Uniform calcium phosphate coatings were obtained in 30 min but coating thickness increased with deposition time. Raising the temperature of electrolyte resulted in more uniform coatings as ionic mobility increased. Low current density was preferable due to hydrogen gas evolving at the cathode, which disturbed the deposition of calcium phosphate crystals on titanium. The amounts of magnesium and hydrogen carbonate ions affected both the homogeneity and morphology of the coatings. This study showed that the electrodeposition method is efficient for coating titanium with osteoconductive calcium phosphate layers.

1 Introduction

Titanium and its alloys are amply used for manufacturing orthopedic prostheses and dental implants due to excellent mechanical properties and biocompatibility [1, 2]. All grades of titanium and its alloys exhibit a layer of TiO_2 with a thickness of few nanometers [3–5]. This titanium oxide layer is responsible for giving chemical stability and thus, biocompatibility. Although titanium is biocompatible, it is not bioactive unless chemically treated with alkali [6–8] or coated with calcium phosphate [9, 10]. Titanium implants are therefore not in direct contact with bone and often encapsulated by a thin layer of fibrous tissue. Calcium phosphate (Ca-P) ceramics are bioactive through a dissolution-reprecipitation process forming a direct bond with bone tissue [11]. However, the mechanical properties of Ca-P ceramics are very weak and these materials could not be used under load bearing applications like titanium implants. Several coating methods have therefore been developed for combining the high strength of titanium with the bioactivity and osteoconductivity of calcium phosphate [12–16]. Hydroxyapatite coatings, 40–50 μm thick, have been applied on titanium implants by using plasma-spray and are successfully used in clinical practice for improving the initial healing and long term stability of orthopedic hip prostheses. However, the plasma-spray method takes place at extremely high temperature possibly affecting the structure of titanium and the homogeneity of the hydroxyapatite coating [17–19]. Furthermore, the plasma spray coating method does not allow the incorporation of bioactive molecules within the Ca-P layer [20]. These bioactive molecules like bone morphogenetic proteins or antibiotics could enhance bone healing or prevent local infection around prostheses [21, 22]. Electrodeposition of calcium phosphate coatings on titanium is an alternative process that uses aqueous solutions at low temperatures, which cannot affect the structure of the implant and which can be applied to complex shapes. Ban *et al.* [23] have studied the effect of temperature and deposition time on

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electrodeposited coatings using titanium plates as working electrodes in a simulated body fluid (SBF) and a voltage of –2 V versus a saturated Calomel electrode (SCE). In another work, the same group removed Mg^{2+}, K^+ and HCO_3^- ions from SBF and investigated the microstructure of coatings obtained within a temperature range of 52–92°C using two current densities of 1.3 and 12.9 mA/cm^2 [24]. Higher temperature ranges of 90–200°C have also been used with current densities of 5–25 mA/cm^2 and deposition times of 10–120 min to study the morphological regulation and crystal growth of calcium phosphate coatings on titanium using a SBF solution without the $MgCl_2{\cdot}6H_2O$, KCl and $NaHCO_3$ [25]. Rößler *et al.* [26] performed studies using an electrolyte consisting of $CaCl_2$ and $NH_4H_2PO_4$, at 36°C and current densities ranging from -0.5 to -10 mA/cm^2 for 5 s to 60 min. The effect of pressure on the electrodeposition process has also been studied [27–29]. The aim of this work is to study the effect of a super saturated electrolyte solution at different temperatures, current densities, time of deposition and various Mg^{2+} and HCO^{3-} ion concentrations. The Ca-P coatings on titanium alloy (Ti6Al4V) were analyzed by using scanning electron microscopy (SEM) energy dispersive analysis for X-Ray (EDX) and X-Ray diffraction (XRD).

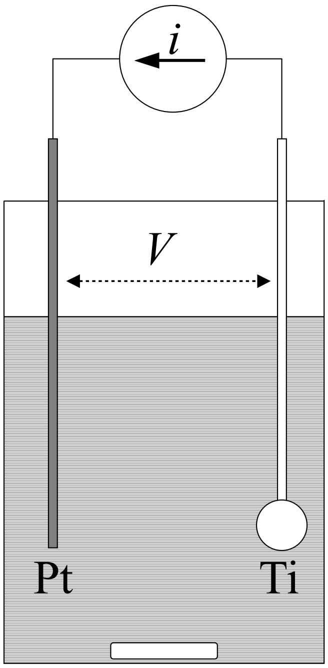


Fig. 1 Electrodeposition setup for calcium phosphate coating on titanium alloy

2 Materials and methods

2.1 Sample preparation

Ti6Al4V titanium discs, 15 mm in diameter × 1 mm in thickness, were used. Before electrodeposition, the titanium alloy discs were grit blasted (Sanduret 3 K, Rosfrei) using biphasic calcium phosphate ceramic particles with a size of 200–400 μm (Biomatlante, France). Samples were then etched in a mixture of acids consisting of nitric acid (HNO_3 50%, 6 ml), fluorydric acid (HF 40%, 3 ml) and water (991 ml). The samples were ultrasonically cleaned in demineralized water and dried. The surface roughness (SJ-201M, Mitutoyo) and the dynamic contact angle (Digidrop, GBX) of the titanium alloy discs were measured after blasting and acid etching.

2.2 Electrodeposition

The experimental set up used for the electrodeposition was a simple two-electrode cell configuration (Fig. 1). The working electrode was the titanium alloy disc and the counter electrode was a platinum mesh (100 × 70 × 0.1 mm, Advent Research Materials Ltd, UK). Both electrodes were immersed in a supersaturated electrolyte (SE) which was prepared by dissolving given amounts of reagent-grade chemicals NaCl, $CaCl_2{\cdot}2H_2O$, $MgCl_2{\cdot}6H_2O$, $NaHCO_3$ and $Na_2HPO_4{\cdot}2H_2O$, in demineralized water. The supersaturated electrolyte was buffered to pH 7.4 using Tris(hydroxymethyl amino methane) and drops of HCl 1M. Different concentrations of Mg^{2+} and HCO_3^- ions ranging from 0.1 to 1 mM were investigated as shown in Table 1.

After dissolution of salts and buffering at pH 7.4, the supersaturated solution was filtered through 0.22 μm top filter bottle (Millipore) and stored at 4°C. For electrodeposition, 250 ml of supersaturated solution was used in each experiment. The electrolyte was thermostated at 25, 37, 50, 65, 70, 75 or 80°C and stirred at 250 rpm using a stirring-heating plate (Heidolph Instruments MR3003, Germany). A direct current power generator (Consort E861) was connected to the titanium alloy and platinum electrodes. Current density of 8, 10, 15, 20, 25, 30, 50, 90 and 120 mA/cm^2 were applied between the electrodes for 10, 20, 30, 60 and 120 min. After electrodeposition, titanium alloy discs were rinsed with demineralized water, dried and stored in multi-well culture dishes at room temperature. A minimum of 2 runs was performed for each experimental condition.

2.3 Characterization of the electrodeposited Ca-P coatings on titanium

Scanning electron microscopy (SEM, Leo 1450VP) was performed on the samples. For comparison purposes, the surfaces were examined at magnifications ×50, ×250, ×500, ×1000, ×2500, ×5000 and ×10000. Energy dispersive X-Ray analysis (EDX, Oxford Instruments) was performed at 15 keV, working distance of 15 mm, magnification ×700 and using a square window. The composition of electrodeposited calcium phosphate crystal was determined using

Table 1 Ion concentrations (mM) of the different saturated solutions with various quantities of Magnesium (M) and Hydrogenocarbonate (HC) used for electrodeposition of calcium phosphate coatings on titanium alloy

	Ion concentration (mM)							
	Na^+	K^+	Mg^{2+}	Ca^{2+}	Cl^-	HCO_3^-	HPO_4^{2-}	SO_4^{2-}
Blood plasma	142.0	5.0	1.5	2.5	103.0	27	1.0	0.5
SE[a]	140.9	—	—	4.0	144.9	—	2.0	—
SE M0.1	140.9	—	0.1	4.0	145.1	—	2.0	—
SE M0.5	140.9	—	0.5	4.0	145.9	—	2.0	—
SE M1.0	140.9	—	1.0	4.0	146.9	—	2.0	—
SE HC0.1	141.0	—	0.1	4.0	144.9	—	2.0	—
SE HC0.5	141.4	—	0.5	4.0	144.9	—	2.0	—
SE HC1.0	141.9	—	1.0	4.0	144.9	—	2.0	—
SE M0.1, HC0.1	141.0	—	0.1	4.0	145.1	0.1	2.0	—
SE M0.1, HC0.5	141.4	—	0.1	4.0	145.1	0.5	2.0	—
SE M0.1, HC1.0	141.9	—	0.1	4.0	145.1	1.0	2.0	—
SE M0.5, HC0.1	141.0	—	0.5	4.0	145.9	0.1	2.0	—
SE M0.5, HC0.5	141.4	—	0.5	4.0	145.9	0.5	2.0	—
SE M0.5, HC1.0	141.9	—	0.5	4.0	145.9	1.0	2.0	—
SE M1.0, HC0.1	141.0	—	1.0	4.0	146.9	0.1	2.0	—
SE M1.0, HC0.5	141.4	—	1.0	4.0	146.9	0.5	2.0	—
SE M1.0, HC1.0	141.9	—	1.0	4.0	146.9	1.0	2.0	—

[a] This solution is our standard supersaturated electrolyte (SE).

atomic Ca/P ratio measured by EDX. The homogeneity of calcium phosphate coating was assessed using the Ca/Ti ratio given by EDX spectra. Samples were analyzed by X-Ray diffraction (XRD, PW1830 Philips) using a Kα copper radiation (40 kV, 30 mA) from 3 to 60 degrees in 2theta with a step of 0.02 degree.

3 Results

After blasting and etching, the titanium alloy discs had an average surface roughness of 1.64 ± 0.14 μm and a contact angle of 99.5 ± 2.29° thus, being slightly hydrophobic. After electrodeposition, the surface roughness remained the same, 1.64 ± 0.10 μm, but the contact angle changed to 13.5 ± 0.17°, hence modifying the surface into a hydrophilic one. In the first set of experiments, different current density and temperature were used. Coatings obtained under different electrodeposition conditions are shown in Fig. 2. At low current density and room temperature, calcium phosphate aggregates, several micrometers wide, were observed on titanium (Fig. 2(a)). Increasing the temperature to 50 or 70°C with the same current density resulted in more homogeneous calcium phosphate layers (Fig. 2(b), and (c)). However, an increase of current density resulted in thinner and less uniform calcium phosphate deposits even if high temperature was used (Fig. 2(d)).

This trend was corroborated by EDX analysis. As shown in Fig. 3(a)–(c), the intensity of calcium and phosphorus peaks increased comparatively to the intensity of titanium peaks while the temperature of the SE was augmented. As previously observed, the calcium phosphate layers were thinner and less homogenous for high current density and high temperature (Fig. 3(d)).

Figure 4(a) exhibited the Ca/Ti ratio for all electrodeposited calcium phosphate coatings on titanium alloy as function of current density and temperature. High Ca/Ti ratio corresponded to thick and homogeneous calcium phosphate coatings. This graph illustrates the optimal experimental conditions for getting thick and uniform calcium phosphate layers. These conditions were high temperature (*e.g.* 50–70°C) and low current density 10–20 mA/cm^2. Figure 4(b) exhibited the Ca/P ratio, which ranged from 1.10 to 2.06 depending on experimental conditions.

It appeared that high Ca/P ratio which are related to calcium-rich phases like hydroxyapatite (HA, Ca/P = 1.67) or AB-type substituted carbonated apatite were preferably obtained using high temperature and low current density. These findings were confirmed by the X-Ray diffraction patterns (Fig. 5). Again, the diffraction lines for HA were relatively higher at low current density and high temperature than in other conditions. In summary, for this first set of experiments, the electrodeposition of calcium phosphate coatings on the titanium surface was favored using low current densities, 10–20 mA/cm^2, and high temperatures, 50–70°C. Using these experimental conditions, coatings were uniform white and their thickness was estimated to be around 20 μm. Thickness of the obtained coatings via the electrodeposition method was estimated by comparing them to a standard obtained by using a biomimetic process described by Barrere *et al.* [30], in which the coating is obtained via a nucleation

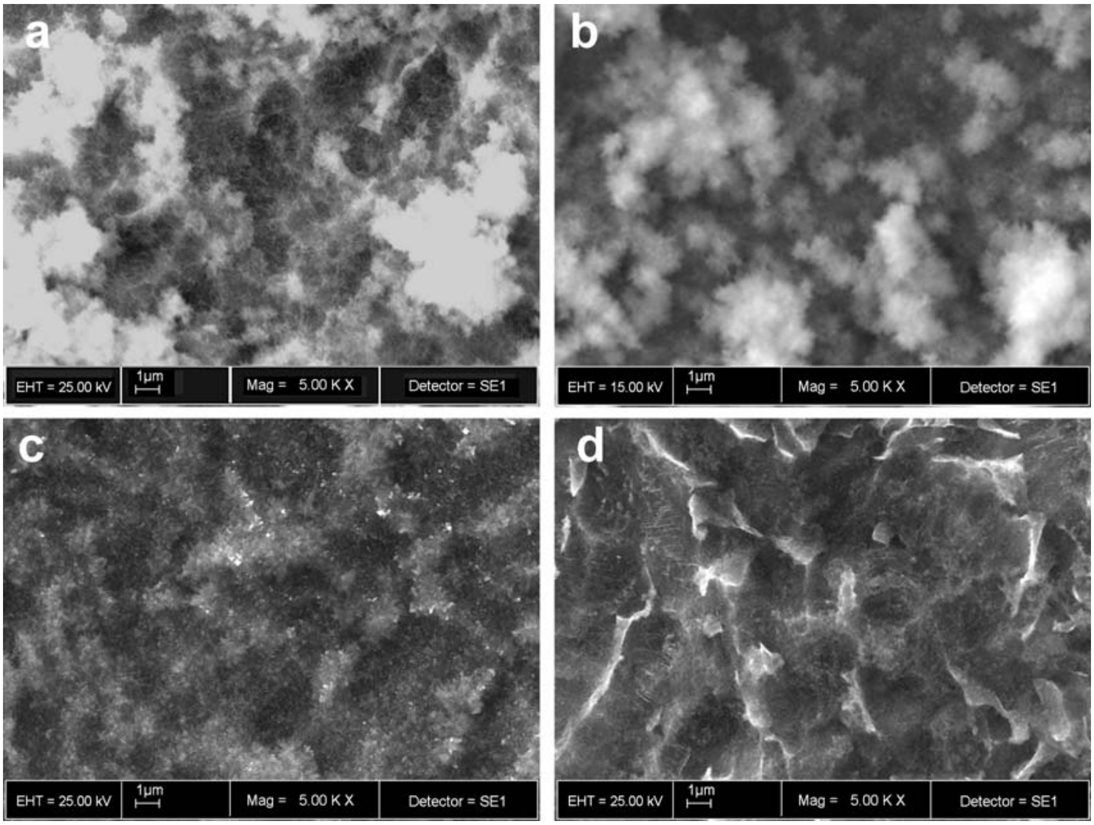


Fig. 2 SEM of electrodeposited calcium phosphate coatings on titanium alloy as function of temperature and current density for 30 min using a SE solution. (a) $T = 25°\mathrm{C}$, $j = 10\ \mathrm{mA/cm^2}$, (b) $T = 50°\mathrm{C}$, $j = 10\ \mathrm{mA/cm^2}$, (c) $T = 70°\mathrm{C}$, $j = 10\ \mathrm{mA/cm^2}$ and (d) $T = 70°\mathrm{C}$, $j = 90\ \mathrm{mA/cm^2}$

and precipitation from a solution whose pH is controlled by the injection of CO_2 to the solution. The colour of the coating and the intensity of the peaks of Ti, Ca and P obtained with EDX were used to evaluate in a qualitative way the thickness.

Table 2 gathers the Ca/Ti and Ca/P ratios obtained with increasing deposition time under the same conditions of temperature of 75°C and current density of 20 mA/cm^2. As the electrodepositing time increased, the Ca/Ti ratio increased suggesting that the coating thickness and homogeneity were greater than after short delays. The chemical composition of the deposited calcium phosphate phase on titanium alloy evolved also with time as evidenced by the Ca/P ratio.

The effect of magnesium and carbonate was also studied in another set of experiments. Ion concentrations for $Mg^{2+}$and HCO_3^- in the supersaturated electrolyte were varied from 0.1 to 1 mM and combinations of those were tested (Table 1). Adding Mg^{2+} or HCO_3^- ions to the supersaturated solutions had an effect on the microstructure of coatings (Fig. 6). Calcium phosphate crystals had a needle-like shape in the presence of magnesium or carbonate.

Figure 7 gathers the EDX spectra of coating prepared with Mg^{2+} or HCO_3^- ions. The coating prepared using high amount of magnesium (SE M1.0) had the best uniformity and thickness as evidenced by the relative intensity of Ca, P and Ti peaks (Fig. 7(b)). The addition of low amount of Mg^{2+} or HCO_3^- ions has led to similar homogeneity and thickness of coatings. These results have been corroborated by systematic measurements of Ca/Ti ratio (Fig. 8(a)).

The more homogeneous and thicker coatings were obtained with the combination of Mg^{2+} and HCO_3^- ions. Nevertheless, coatings were thinner and more heterogeneous using high concentration of Mg^{2+} and low concentration ofHCO_3^-. The best coatings were obtained with a concentration of Mg^{2+} of 0.1 mM. In the case of the HCO_3^-, the higher the concentration, the thinner the coating was (Fig. 8). Regarding composition, Ca/P ratios were comprised between 1.62 and 1.88 for most the concentrations of Mg^{2+} and HCO_3^-. A low Ca/P ratio of 0.74 was solely noted for magnesium content of 0.1 mM.

The XRD patterns indicated that the coatings were composed of an apatitic phase with a low crystallinity (Fig. 9). Peaks of titanium were still visible but those of HA were relatively higher than in the previous set of experiment. In summary, magnesium and carbonate increased the uniformity and thickness of the electrodeposited calcium phosphate layers on titanium alloy.

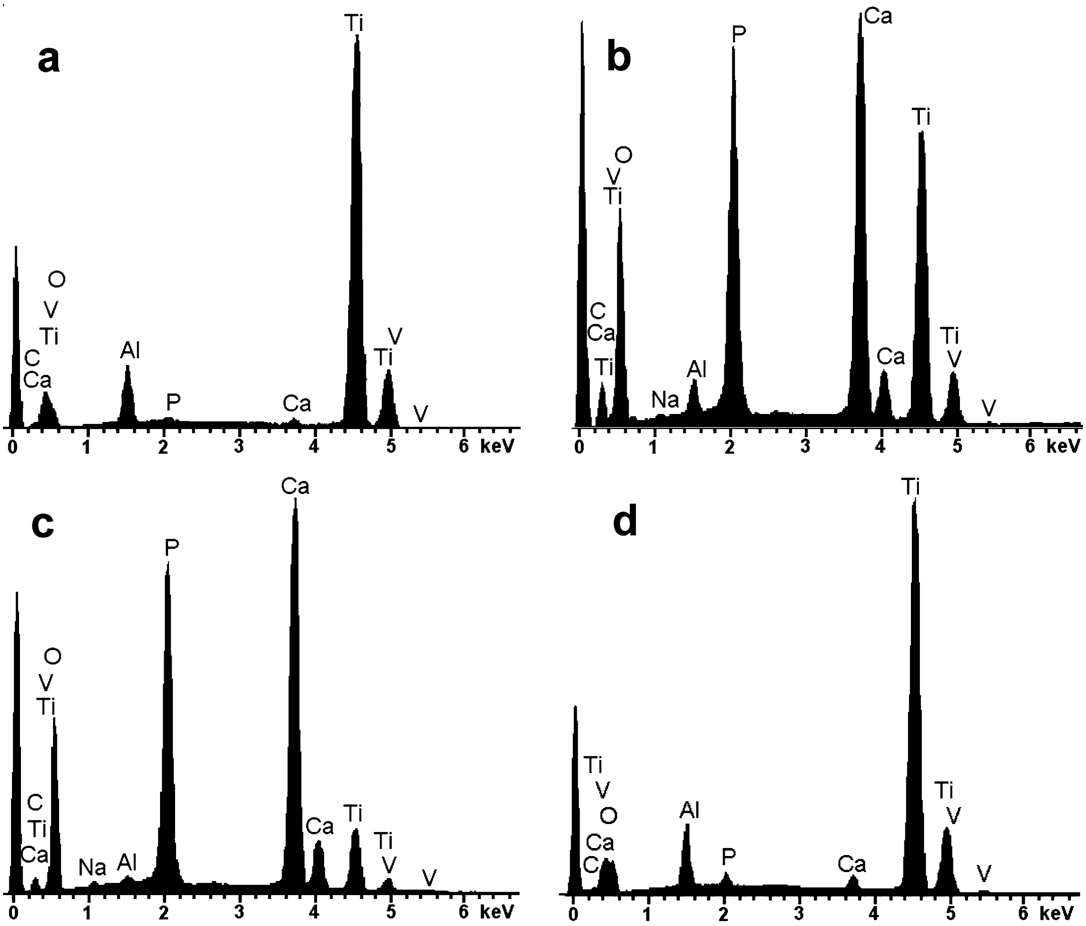


Fig. 3 EDX spectra of electrodeposited calcium phosphate coatings on titanium alloy as function of temperature and current density for 30 min using a SE solution. (a) $T = 25°\text{C}$, $j = 10$ mA/cm^2, (b) $T = 50°\text{C}$, $j = 10$ mA/cm^2, (c) $T = 70°\text{C}$, $j = 10$ mA/cm^2 and (d) $T = 70°\text{C}$, $j = 90$ mA/cm^2

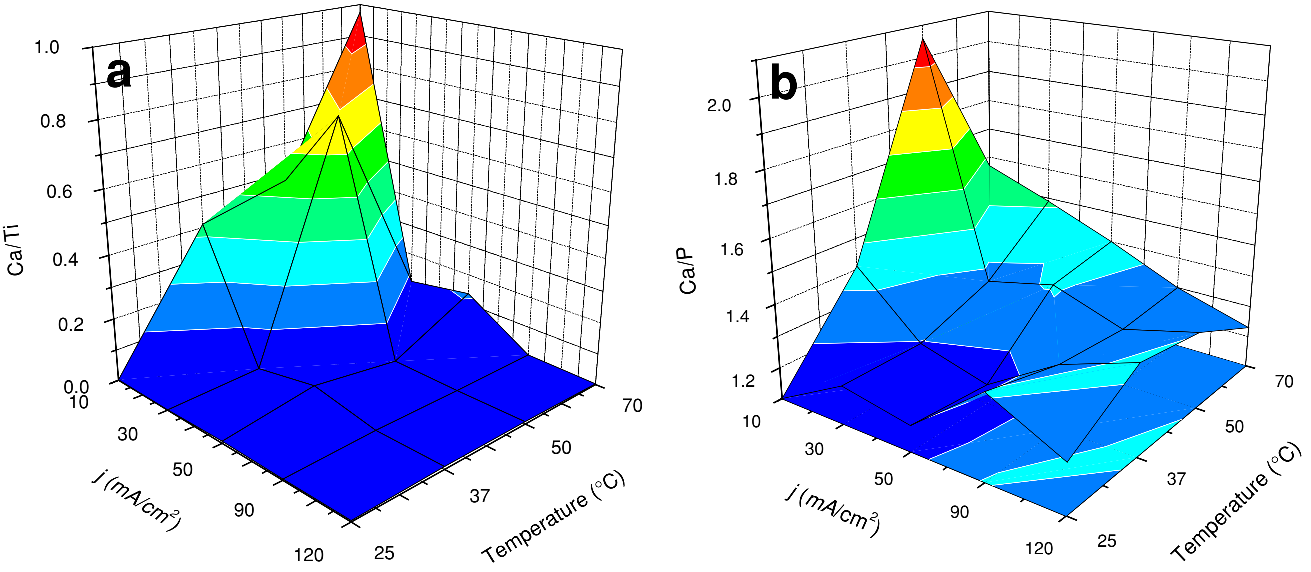


Fig. 4 Ca/Ti and Ca/P ratios measured by EDX for electrodeposited calcium phosphate coatings on titanium alloy as function of temperature and current density for 30 min using a SE solution

Table 2 Ca/Ti and Ca/P ratios of eletrodeposited calcium phosphate coatings on titanium alloy as function of deposition time for a SE[a] solution heated at 75°C and a current density of 20 mA/cm^2

Time (min)	10	20	30	60	120
Ca/Ti	0.025 ± 0.001	0.038 ± 0.001	0.11 ± 0.002	0.68 ± 0.07	2.2 ± 0.12
Ca/P	1.64 ± 0.08	1.73 ± 0.04	1.58 ± 0.02	1.91 ± 0.03	1.96 ± 0.007

[a] See Table 1.

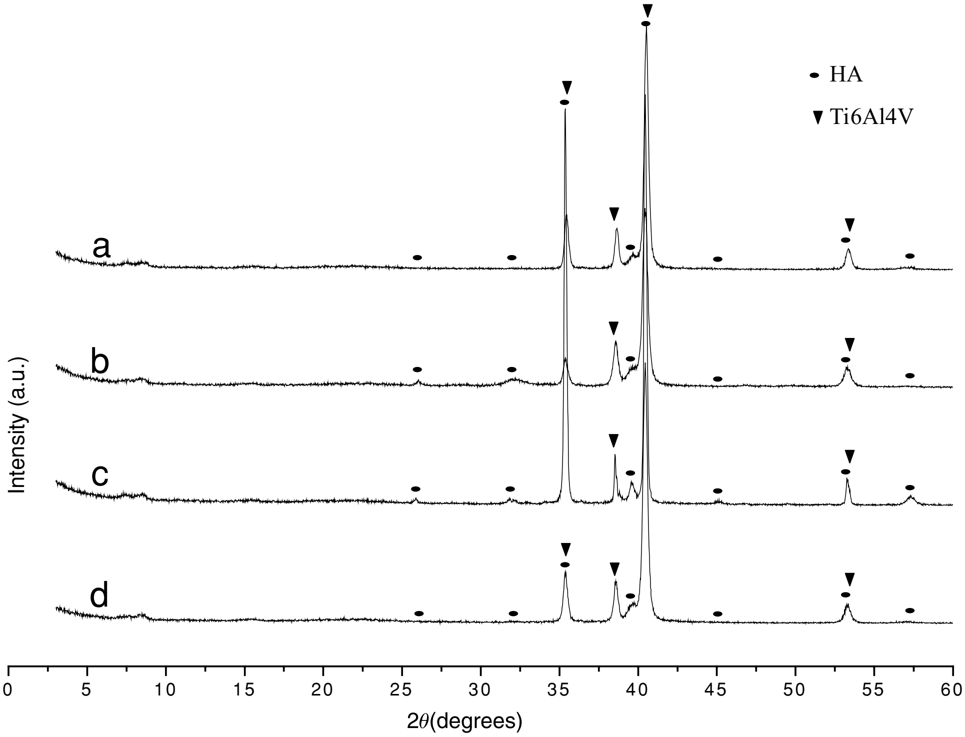


Fig. 5 X-ray diffraction patterns of electrodeposited calcium phosphate coatings on titanium alloy as function of temperature and current density for 30 min using a SE solution. (a) $T = 25°\text{C}$, $j = 10\ \text{mA/cm}^2$, (b) $T = 50°\text{C}$, $j = 10\ \text{mA/cm}^2$, (c) $T = 70°\text{C}$, $j = 10\ \text{mA/cm}^2$ and (d) $T = 70°\text{C}$, $j = 90\ \text{mA/cm}^2$

4 Discussion

Several experimental parameters have been investigated for coating titanium alloy with calcium phosphate by using electrochemistry. Supersaturated electrolytes buffered at pH 7.4 have been used for depositing apatite rather than other calcium phosphate phases. In this electrochemical process, both hydrogen gas and hydroxyl ions were produced on the titanium surface being the cathode. The production of hydroxyl ions at the surface of titanium raised locally the pH of the supersaturated electrolyte leading to the precipitation of calcium phosphate hydroxyapatite as following:

$$2H_2O + 2e^- \rightarrow H_2 \uparrow + 2OH^- \quad (1)$$

$$2H_3O^+ + 2e^- \rightarrow H_2 \uparrow + 2H_2O \quad (2)$$

$$O_2 + H_3O^+ + 4e^- \rightarrow 3OH^- \quad (3)$$

$$10Ca^{2+} + 6PO_4^{3-} + 2OH^- \rightarrow Ca_{10}(PO_4)_6(OH)_2 \downarrow \quad (4)$$

These reactions are considered to be the ones having a major effect on the formation of calcium phosphates because of their active role in controlling the pH and producing hydroxyl ions [31]. The augmentation of pH facilitates the growth of hydroxyapatite [31], which is related to the reduction of water. In this work, homogenous calcium phosphate layers were produced on titanium within a relatively short time, *e.g.* 30 min. As evidenced by XRD analysis (Figs. 5 and 9), the calcium phosphate coating had an apatite structure. Nevertheless, it is well know [23–25, 32, 33] that near the surface of the cathode the reduction of water produces hydrogen gas. The evolving of hydrogen gas at the cathode may disturb the deposition of calcium phosphate crystals on the titanium surface. The electrolyte was stirred at 250 rpm in order to help degassing hydrogen gas from the cathode and to improve the coating uniformity [34]. In this work, we noticed that increasing the temperature affected the uniformity of the coatings. An increase in temperature facilitated both the transport of ions [23] and the crystal growth of hydroxyapatite crystals leading to more uniform and thicker coatings. High current

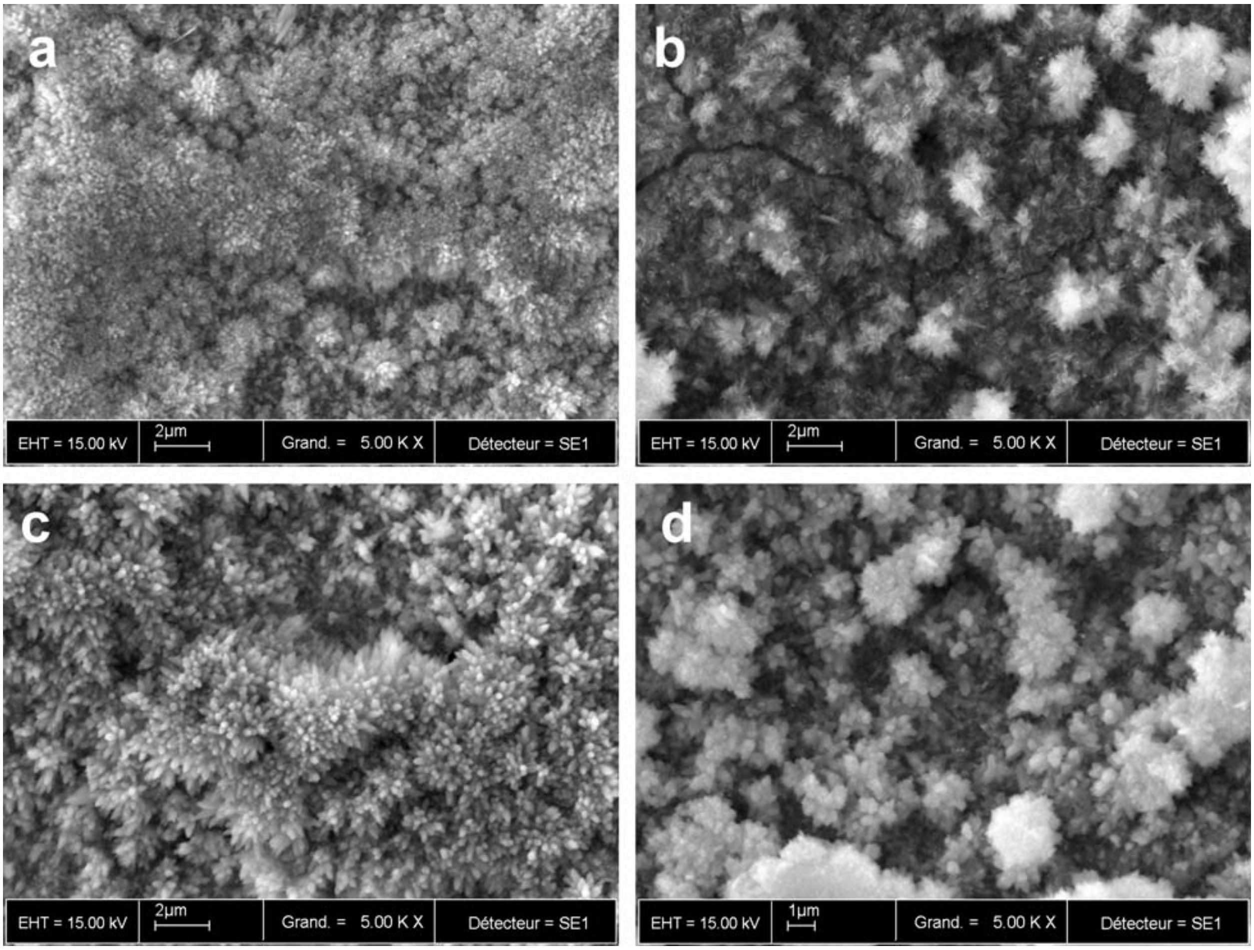
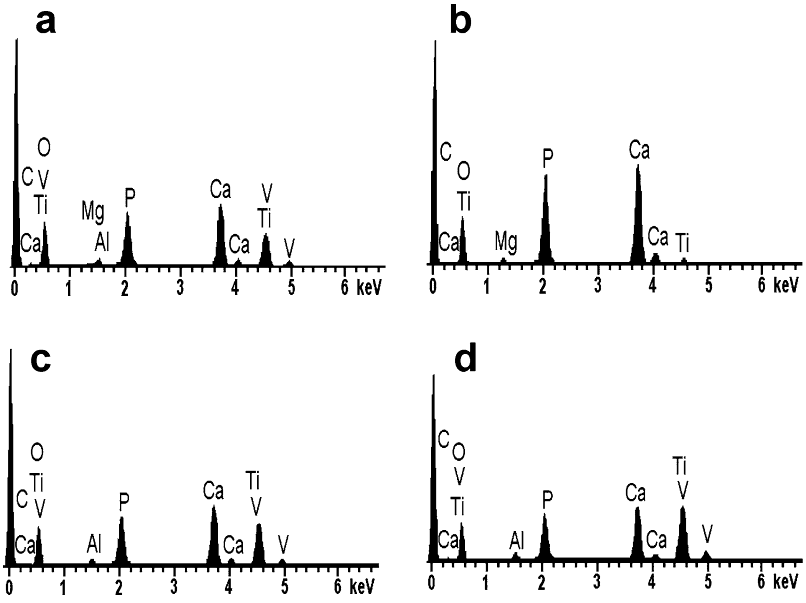


Fig. 6 SEM of electrodeposited calcium phosphate coatings on titanium alloy as function of magnesium and hydrogen carbonate content in the supersaturated electrolyte at 75°C and 20 mA/cm^2 for 30 min. (a) $Mg^{2+} = 0.1$ mM, (b) $Mg^{2+} = 1$ mM, (c) $HCO_3^- = 0.1$ mM and (d) $HCO_3^- = 1$ mM

Fig. 7 EDX spectra of electrodeposited calcium phosphate coatings on titanium alloy as function of magnesium and hydrogen carbonate content in the supersaturated electrolyte at 75°C and 20 mA/cm^2 for 30 min. (a) $Mg^{2+} = 0.1$ mM, (b) $Mg^{2+} = 1$ mM, (c) $HCO_3^- = 0.1$ mM and (d) $HCO_3^- = 1$ mM



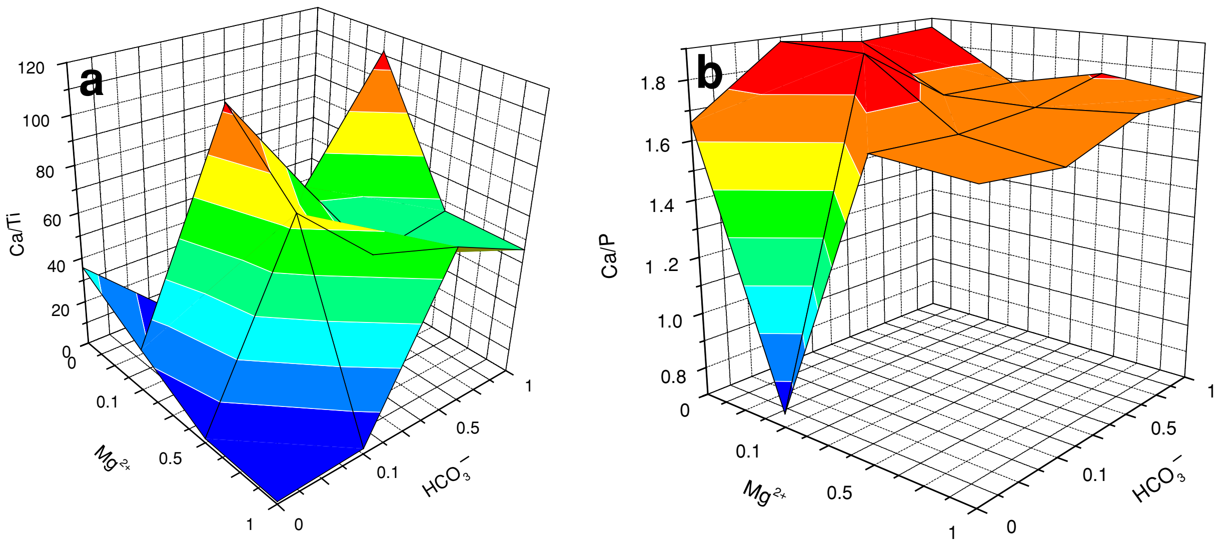


Fig. 8 Ca/Ti and Ca/P ratios measured by EDX for electrodeposited calcium phosphate coatings on titanium alloy as function of magnesium and hydrogen carbonate content ion supersaturated electrolyte for 30 min

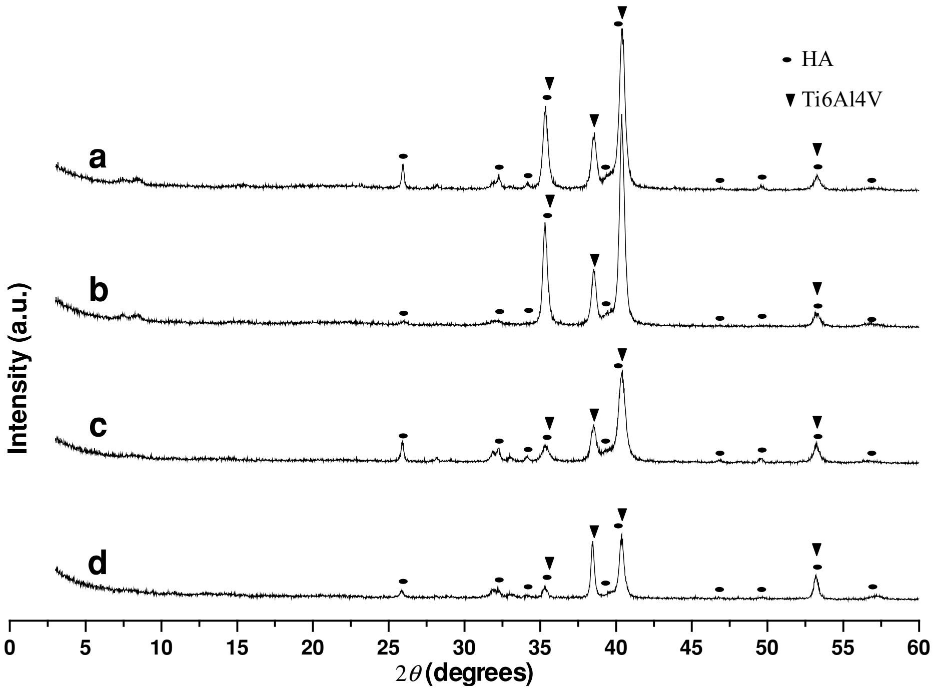


Fig. 9 X-ray diffraction patterns of electrodeposited calcium phosphate coatings on titanium alloy as function of magnesium and hydrogen carbonate content in the supersaturated electrolyte at 75°C and 20 mA/cm^2 for 30 min. (a) $Mg^{2+} = 0.1$ mM, (b) $Mg^{2+} = 1$ mM, (c) $HCO_3^- =$ 0.1 mM and (d) $HCO_3^- = 1$ mM

density enhances ions transport [24] but at the same time produces more hydrogen gas than at low density. In a general way, a sort of equilibrium between ions transport, apatite precipitation and hydrogen gas production was reached compromising current density with temperature. Indeed, the best homogenous calcium phosphate coatings were obtained with low current densities and high temperatures. These conditions favored the transport of ions and reduced the production of hydrogen. However, the current density affects the deposition time as at low current density the deposition of calcium phosphate requires more time. In our experiments, a deposition time of 30 min seems to be good enough to form a homogenous calcium phosphate coating on titanium. Ban *et al.* [25] found that there is a relationship between, principally three parameters: Temperature, current density and time of current loading, and that mainly the temperature affected the precipitation and growth of crystals followed in a slightly dependency on the current density. Their findings are in good agreement with the present results. Furthermore, both temperature and pH affects the solubility of calcium phosphates. For a current density higher than 3 mA/cm^2 it has been reported that pH is higher than 7.4 at the titanium cathode, a value which is appropriate for hydroxyapatite precipitation [35]. Lin *et al.* [33] reported an electrochemical method in which they performed the deposition under a controlled pulse current time to control the movement of calcium and phosphate ions. From our observations, we can establish that the factor that affected more the uniformity of the coating was the hydrogen gas production. The electrochemical pulsed method might be a good approach for getting uniform coatings on titanium. As the solubility of hydrogen gas depends on the temperature and pressure in the cell, Wang *et al.* [27] improved the electrodeposition process by performing it at low pressure. In this way, the hydrogen bubbles produced on titanium were more easily lifted up from the cathode surface. Kuo *et al.* [32] mentioned that at current densities between 10–20 mA/cm^2 OH^- ion concentration is high enough to convert HPO_4^{2-} into PO_4^{3-}. Increasing the concentrations of Mg^{2+} and HCO_3^- ions had an effect on the uniformity and thickness of coatings. It has been noticed that the magnesium and carbonate ions inhibit the crystal growth of hydroxyapatite and contribute to the kinetics formation of a homogeneous calcium phosphate film [30, 36, 37]. In addition, the content of CO_3^{2-} ions favors the precipitation of a carbonated apatite, which is thermodynamically more stable in water than pure hydroxyapatite [38]. Furthermore, it was found that treatments applied to the surface prior electrodeposition created a rough surface with hydrophobic properties. After electrodeposition, the surface of titanium covered by a calcium phosphate layer was hydrophilic and rough. Both roughness and surface energy are two parameters that are important for cell attachment [39]. Osteoblast attachment and spreading are greater on hydrophilic surfaces compared to hydrophobic surfaces [40]. Laurence *et al.* [40] found that reducing the contact angle of a titanium alloy contributes to a better suitability of it as a biomaterial since with these characteristics there is a better response of human fibroblast and osteoblast cells. Aparicio *et al.* [41] tested titanium surfaces with different roughness, obtained with different grit blasting and thermo chemical treatments, by cultivating human osteoblast cells on their samples. They found out that rough and bioactive surfaces with a hydrophilic behaviour stimulate the initial steps of cell adhesion.

5 Conclusion

Calcium phosphates coatings were deposited on a titanium alloy surface by an electrodeposition method within 30 min Temperatures around 75°C, low current densities, low $Mg^{2+}$and HCO_3^- concentrations favored the uniformity and thickness of coatings. These electrodeposited calcium phosphate coatings are expected to enhance bone tissue apposition and healing around titanium implants. Next efforts will be addressed to study the coatings obtained *in vitro* and *in vivo* in a next series of experiments to compare in a same study the performance of this methodology only with cells and in a real organism. In the same way, the incorporation of antibiotics into the coatings could be studied.

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References

1. J. BRETTLE, *Injury* **2**(1) (1970) 26.
2. G. RYAN, A. PANDIT and D. P. APATSIDIS, *Biomaterials* **27**(13) (2006) 2651.
3. U. ROLANDER, L. MATTSSON, J. LAUSMAA and B. KASEMO, *Ultramicroscopy* **19**(4) (1986) 407.
4. C. LARSSON, P. THOMSEN, J. LAUSMAA, M. RODAHL, B. KASEMO and L. E. ERICSON, *Biomaterials* **15**(13) (1994) 1062.
5. J. LAUSMAA, *J. Electron. Spectrosc.* **81**(3) (1996) 343.
6. H. TAKADAMA, H.-M. KIM, T. KOKUBO and T. NAKAMURA, *Sci. Tech. Adv. Mater.* **2**(2) (2001) 389.
7. T. KOKUBO, H.-M. KIM and M. KAWASHITA, *Biomaterials* **24**(13) (2003) 2161.
8. M. TAKEMOTO, S. FUJIBAYASHI, M. NEO, J. SUZUKI, T. MATSUSHITA, T. KOKUBO and T. NAKAMURA, *Biomaterials* **27**(13) (2006) 2682.
9. P. HABIBOVIC, J. LI, C. M. VAN DER VALK, G. MEIJER, P. LAYROLLE, C. A. VAN BLITTERSWIJK and K. DE GROOT, *Biomaterials* **26**(1) (2005) 23.
10. E. GOYENVALLE, E. AGUADO, J.-M. NGUYEN, N. PASSUTI, L. LE GUEHENNEC, P. LAYROLLE and G. DACULSI, *Biomaterials* **27**(7) (2006) 1119.
11. G. DACULSI, O. MALARD and E. GOYENVALLE, *ITBM-RBM* **26**(3) (2005) 218.
12. C. X. WANG, Z. Q. CHEN, L. M. GUAN, M. WANG, Z. Y. LIU and P. L. WANG, *Nucl. Instr. Meth. Phy. Res. Sect. B: Beam Inter. Mater. Atoms* **179**(3) (2001) 364.
13. I.-S. LEE, D.-H. KIM, H.-E. KIM, Y.-C. JUNG and C.-H. HAN, *Biomaterials* **23**(2) (2002) 609.
14. B. FENG, J. Y. CHEN, S. K. QI, L. HE, J. Z. ZHAO and X. D. ZHANG, *Biomaterials* **23**(1) (2002) 173.
15. S. W. K. KWEH, K. A. KHOR and P. CHEANG, *Biomaterials* **23**(3) (2002) 775.
16. L. YAN, Y. LENG and L.-T. WENG, *Biomaterials* **24**(15) (2003) 2585.
17. H. C. GLEDHILL, I. G. TURNER and C. DOYLE, *Biomaterials* **22**(11) (2001) 1233.
18. Y. FU, A. W. BATCHELOR, Y. WANG and K. A. KHOR, *Wear* **217**(1) (1998) 132.
19. Y. FU, A. W. BATCHELOR and K. A. KHOR, *Wear* **230**(1) (1999) 98.
20. M. STIGTER, K. DE GROOT and P. LAYROLLE, *Biomaterials* **23**(20) (2002) 4143.
21. Y. LIU, K. DE GROOT and E. B. HUNZIKER, *Bone* **36**(5) (2005) 745.
22. M. STIGTER, J. BEZEMER, K. DE GROOT and P. LAYROLLE, *J. Control. Release* **99**(1) (2004) 127.
23. S. BAN and S. MARUNO, *Biomaterials* **16**(13) (1995) 977.

24. S. BAN and S. MARUNO, *Biomaterials* **19**(14) (1998) 1245.
25. S. BAN and J. HASEGAWA, *Biomaterials*. **23**(14) (2002) 2965.
26. S. ROSSLER, A. SEWING, M. STOLZEL, R. BORN, D. SCHARNWEBER, M. DARD and H. WORCH, *J. Biomed. Mater. Res.* A **64**(4) (2003) 655.
27. S.-H. WANG, W.-J. SHIH, W.-L. LI, M.-H. HON and M.-C. WANG, *J. Eur. Ceram. Soc.* **25**(14) (2005) 3287.
28. W.-J. SHIH, Y.-H. CHEN, S.-H. WANG, W.-L. LI, M.-H. HON and M.-C. WANG, *J. Cryst. Growth* **285**(4) (2005) 633.
29. S. BAN and S. MARUNO, *J. Biomed. Mater. Res.* **42**(3) (1998) 387.
30. F. BARRERE, P. LAYROLLE, C. A. VAN BLITTERSWIJK and K. DE GROOT, *Bone* **25**(2) (Suppl. 1) (1999) 107S.
31. X. LU and Y. LENG, *Biomaterials* **26**(10) (2005) 1097.
32. M. C. KUO and S. K. YEN, *Mat. Sci. Eng.* C **20** (1–2) (2002) 153.
33. S. LIN, R. Z. LEGEROS and J. P. LEGEROS, *J. Biomed. Mater. Res.* A **66**(4) (2003) 819.
34. X. HOU, X. LIU, J. XU, J. SHEN and X. LIU, *Mater. Lett.* **50** (2–3) (2001) 103.
35. J. M. ZHANG, C. J. LIN, Z. D. FENG and Z. W. TIAN, *J. Electroanal. Chem.* **452**(2) (1998) 235.
36. F. BARRERE, C. A. VAN BLITTERSWIJK, K. DE GROOT and P. LAYROLLE, *Biomaterials* **23**(10) (2002) 2211.
37. F. BARRERE, C. A. VAN BLITTERSWIJK, K. DE GROOT and P. LAYROLLE, *Biomaterials* **23**(9) (2002) 1921.
38. J.-H. PARK, D.-Y. LEE, K.-T. OH, Y.-K. LEE, K.-M. KIM and K.-N. KIM, *Mater. Lett.* In Press, Corrected Proof.
39. M. SVEHLA, P. MORBERG, W. BRUCE, B. ZICAT and W. R. WALSH, *J. Arthroplasty* **17**(3) (2002) 304.
40. J. LAWRENCE, L. HAO and H. R. CHEW, *Surf. Coat. Tech.* **200** (18–19) (2006) 5581.
41. C. APARICIO, F. J. GIL, J. A. PLANELL and E. ENGEL, *J. Mater. Sci.: Mater. Med.* **13**(12) (2002) 1105.