An innovative auto-catalytic deposition route to produce calcium–phosphate coatings on polymeric biomaterials

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The aim of this research is to develop a new methodology to obtain bioactive coatings on bioinert and biodegradable polymers that are not intrinsically bioactive. In this study three types of materials were used as substrates: (i) high molecular weight polyethylene (HMWPE) and two different types of starch based blends (ii) starch/ethylene vinyl alcohol blends, SEVA-C and (iii) starch/cellulose acetate blends, SCA. Two types of baths were originally proposed and studied to produce novel auto-catalytic calcium–phosphate (Ca–P) coatings. Then, the coated surfaces were analyzed by scanning electron microscopy and energy dispersive spectroscopy (SEM/EDS), as produced, and after different immersion periods in SBF. The evolution of Ca and P concentrations was determined by induced-coupled plasma emission (ICP) spectroscopy. The crystalline phases present on the films formed on the different material surfaces, after a certain soaking time, were identified by thin-film X-ray diffraction (TF-XRD). The obtained results indicated that it was possible to coat the materials surfaces with a Ca–P layer with only 60 min of immersion in both types of auto-catalytic solutions. Furthermore, it was possible to observe the clear bioactive nature of the Ca–P coatings after different immersion periods in a simulated body fluid (SBF). The results from TF-XRD confirmed the presence of partially amorphous Ca–P films with clearly noticeable hydroxylapatite peaks. These new methodologies allow for the production of an adherent bioactive film on the polymeric surfaces prior to implantation, which may allow for the development of bone-bonding, bioabsorbable implants and fixation devices.

1. Introduction

Hydroxyapatite ($Ca_{10}(PO_4)_6(OH)_2$, HA) is the major constituent of the bone and teeth [1–3], being the calcium phosphate (Ca–P) more widely selected for studies related with the medical field [1, 4]. Due to its good biocompatibility [1–3] and efficacy in promoting biointegration for implants in hard and soft tissue [5, 6], HA has been used in various forms during the last decades. However, one common problem of HA, as well as of all ceramics, is their poor mechanical properties such as low bending strength and relatively inferior values of fracture toughness, which is an obstacle for then to be used in load-bearing clinical applications [7, 8]. Despite of these problems, HA can be effectively used as a coating material on mechanically superior materials [9]. In fact, in the last few years several techniques to coat orthopaedic implants with calcium phosphates have been investigated [5, 9–11]. However, most of the available methods to produce adequate Ca–P coatings, such as the plasma-spraying technique, which is the most common and widely used technique, are difficult to control on what concerns to the Ca–P layer composition, degree of crystallinity and capability to generate strong bonds with the substrates, i.e., to produce coatings with a good adhesion [5, 9, 12, 13]. Furthermore, it has been claimed that Ca–P coatings produced in aqueous medium, such as in biomimetic coatings [14–17], present a stronger bone-bonding ability [14–17].

In recent years, electroless plating has received more attention, mainly due to the fact that this process may be used for metallizing non-conducting materials such as polymers and ceramics to be used in a range of applications [18–20]. However the technique has never been used before to our knowledge to produce bioactive Ca–P coatings on the surface of polymeric biomaterials.

In the present work a new auto-catalytic deposition methodology was developed and used to coat several polymeric biomaterials. The treatment developed may well be used also to coat metal implants. The developed coating methodology uses a deposition route that is totally ''electroless'', i.e., does not require the use of electric current application for its application, being

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TABLE I Auto-catalytic bath compositions and respective operating conditions for electroless production of Ca–P coatings

Bath	Composition (conc. g/L)	pH adjust.	Bath temp.
Alkaline	Calcium chloride ($CaCl_2$) – 25 Sodium pyrophosphate ($NaP_2O_7 \cdot 10H_2O$) – 50 Sodium hypophosphite ($NaH_2PO_2 \cdot H_2O$) – 21 Palladium chloride ($PdCl_2$) – 0.885	NaOH: 9.2 ± 0.1	$60 \pm 2\,^{\circ}C$
Acid	Calcium chloride ($CaCl_2$) – 21 Sodium fluoride (NaF) – 5 Sodium hypophosphite ($NaH_2PO_2 \cdot H_2O$) – 24 Succinic acid ($C_4H_6O_4$) – 7 Palladium chloride ($PdCl_2$) – 0.885	HCl: 5.3 ± 0.1	$80 \pm 2\,^{\circ}C$

based on redox reactions. Therefore, the Ca–P coating is deposited on the materials surfaces by an auto-catalytic chemical reaction. The proposed method has or might have the following advantages over conventional ones, such as plasma spraying: (i) coatings can be produced onto complex-shaped and/or microporous implants, (ii) the wet formed Ca–P layer is expected to be formed under more controlled conditions, (iii) it is a simple and cost-effective way to produce Ca–P coatings and (iv) there is no adverse effect of heat on substrates.

The aim of the present research was to obtain bioactive coatings on both bioinert and biodegradable polymers that are not intrinsically bioactive. The production of auto-catalytic coatings, prior to implantation, on implant materials based on corn starch or on polyethylene, may allow for the development of bone-bonding, biodegradable (bioinert) bone replacement and fixation devices.

2. Materials and methods

2.1. Substrate materials

In this study three types of materials were used as substrates: (i) high molecular weight polyethylene (HMWPE, Hostalen® GM 9255F, Hoescht, Germany) and two different types of starch blends (in both cases with 50 ± 2 wt % starch), (ii) starch/ethylene vinyl alcohol blend – SEVA-C, and (iii) starch/cellulose acetate blend – SCA, both from Novamont, Italy. These polymers were obtained by conventional injection molding in a Klockner–Ferromatik Desma FM20 machine. Small dumb-bell ASTM tensile samples (cross section $2 \times 4\,mm^2$) were produced according to previously reported [21]. Further details on the materials processing and respective mechanical properties can be found elsewhere [21, 22].

2.2. Surface pre-treatments

Prior to the coating process, the HMWPE samples were polished on both surfaces with 3 μm diamond slurry followed by washing with distilled water in an ultrasonic cleaner for 5 min. SEVA-C and SCA samples were irradiated by a UV light for 24 h, in a Hanovia Uvitron irradiation system, with a 100 W high-pressure mercury lamp (wavelength ranging from 254 to 365 nm), in order to activate its surface [23] as a way to improve the adhesion of the films to the polymeric surfaces.

2.3. Coating methodology

In order to produce electroless Ca–P coatings two types of baths were originally proposed and studied: (i) alkaline and (ii) acid bath. The respective compositions and typical operating conditions for each bath are shown in the Table I.

In both baths Palladium chloride was used as the catalyzer and worked as an arm of ionic change between the substrate and the solution. The calcium chloride provides the calcium and the sodium hypophosphite acts as a reducing agent. In the alkaline bath the sodium pyrophosphate provide the phosphorus. In the acid bath the sodium fluoride is an etching agent (specially useful when coating metals) and the succinic acid acts as an accelerator of the reaction.

After the surface treatment the samples were immersed in the baths, which were continuously agitated by a magnetic stirrer for different time periods. Then, the samples were cleaned with distilled water and dried in controlled environmental conditions (23 °C and 55% RH).

2.4. Bioactivity evaluation

Standard *in vitro* bioactivity tests [24] were performed in order to investigate if there was a formation of a Ca–P layer and if there was some dissolution of the coatings. The coated samples were soaked in the SBF, that has ion concentrations similar to those of human plasma, at 37 °C and pH = 7.35 for different periods up to 14 days. After soaking, the specimens were immediately cleaned with distilled water and dried at 23 °C under a constant relative humidity of 55%.

2.5. Surface analysis

The morphology of the coating was analyzed by scanning electron microscopy and energy dispersive spectroscopy (SEM/EDS), as produced, and after different immersion periods in SBF. The correspondent Ca/P values were determined using well-stabilized sub-routines for EDS semi-quantitative analysis. Thin-film X-ray diffraction (TF-XRD) was used to identify the crystalline phases present and to characterize the crystalline/amorphous nature of the formed Ca–P bioactive layers (as produced).

2.6. Solution analysis

The evolution of the Ca and P ion concentrations were determined as function of immersion time in SBF. The

ion concentrations were measured by induced-coupled plasma emission (ICP) spectroscopy.

3. Results and discussion

3.1. Auto-catalytic coating methodology

By using the proposed auto-catalytic deposition methodology it was possible to coat all materials surfaces with a Ca–P layer. The pre-treatments (UV activation and polishing) performed on the surfaces of all materials lead to a better adhesion between the polymer and coating. The effects of UV radiation on starch based polymers may be explained due to the presence of polar groups on the SEVA-C and SCA surface, which seems to facilitate the connection with an apatite layer, i.e., with calcium or hydroxyl ion of the apatite, as reported in others works [23, 25]. Kokubo *et al*. [26] demonstrated that a presence of polar groups on the surface of the polymeric substrates increase the adhesive strength because the respective adhesive strength is assumed to increase with increasing number of points at which the apatite nuclei are attached to the substrates. For the HMWPE, with the polishing treatment, the surface roughness had increased, enhancing the coating adhesion, due to the fact that in general, a rougher surface promotes nucleation (over that of a smooth surface) as a result of the lower free surface energy [27].

After 60 min of immersion, which was the typical coating time for both baths, the surface of the three polymers was completely covered with a calcium phosphate layer, as it is shown in Figs. 1–3.

From the characterization of the coatings generated by the two types of baths, it was found that there were considerable differences in the morphology and crystallinity of these Ca–P films. In fact, it was possible to observe that the acid coating (that typically operates at 80 °C) generates a film with a more pronounced needle like morphology than the alkaline coatings (that typically operates at 60 °C) (Figs. 2(b) and (d)).

These differences are probably related with the fact that operation temperature of the acid bath is usually higher than of the alkaline bath, which help to maintain a higher deposition rate [28, 29]. Furthermore, the major advantages of hypophosphite containing acidic solution include its lower cost, the higher deposition rate, good stability, and better physical properties of the deposits over alkaline solution [19, 20]. Also, the addition of succinic acid in the acid bath helps to increase the speed of the reaction, as it has been reported for other types of auto-catalytic coatings [30].

The XRD patterns of the coatings formed on the alkaline bath on SCA and SEVA-C substrates and their corresponding un-coated substrates are shown in Fig. 4 as a function of immersion time.

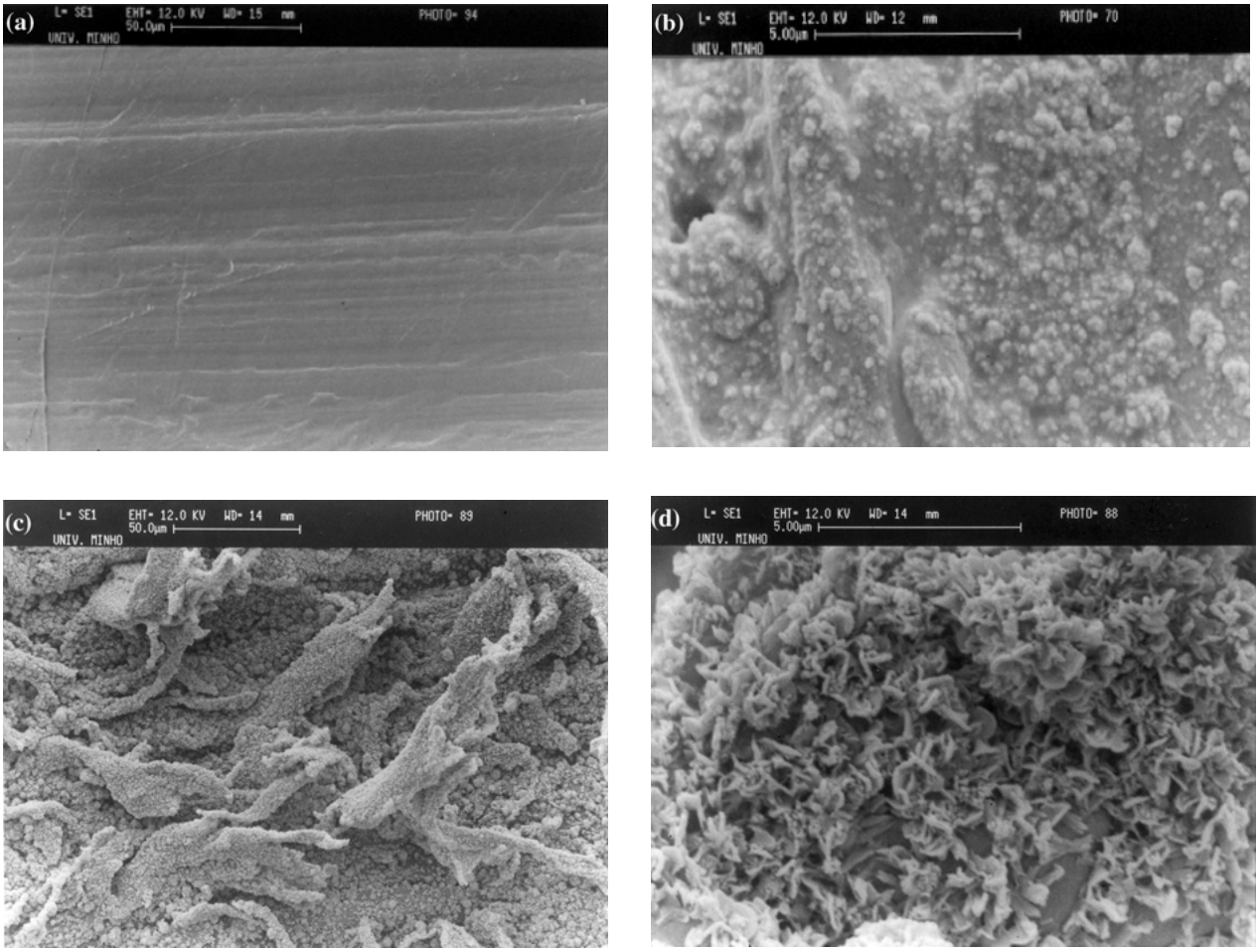

Figure 1 SEM photographs of the Ca–P coatings produced on the surfaces of HMWPE substrate. Sample before coating (a), after 60 min in acid bath (b) and then immersed for (c) 14 days in SBF solution. Magnification (d) showing a detail of the structure presented in (c).

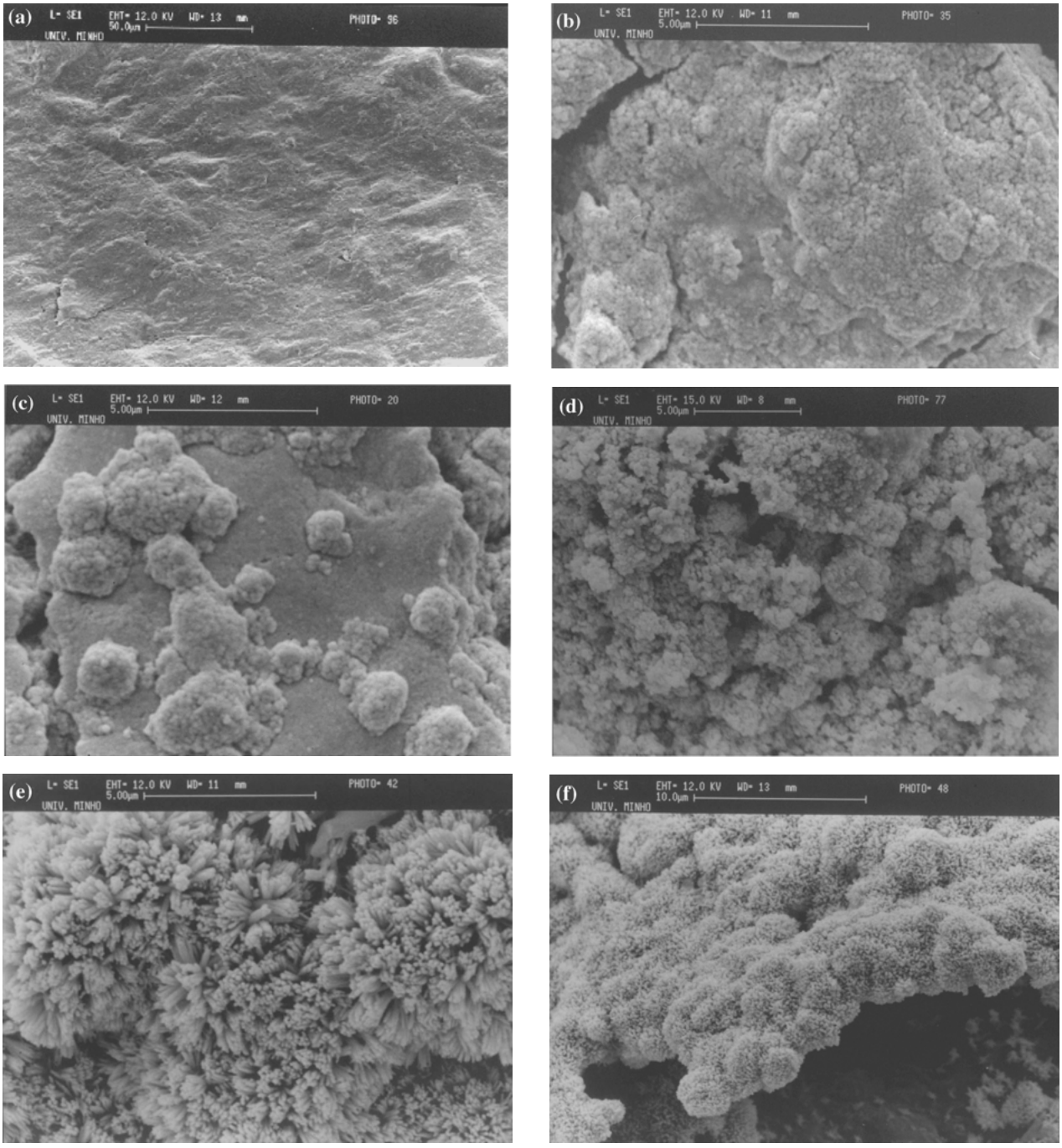

Figure 2 SEM photographs of Ca–P coatings produced on the surfaces SCA substrate. Sample before coating (a), after (b) 60 min in alkaline bath, then immersed for (c) 14 days in SBF solution and after 60 min in acid bath (d) and then immersed for (e) 14 days in SBF solution and (f) the respective cross section.

It may be observed that the Ca–P coatings have a partially crystalline structure with apatite peaks similar to those of bone apatite. This fact was confirmed by the matching of the XRD spectra with the standard pattern of hydroxylapatite (JCPDS 9-432), although the partially amorphous nature (similar to human bone apatite) of this Ca–P film was also evident. These coatings are expected to be more reactive than high crystallinity ones. Also, for longer immersion times in the alkaline bath, it was possible to observe the gradual increase of the intensity of the apatite peaks, which corresponds to the growth of an apatite layer on the substrate. However, the intensity of the apatite peaks for SEVA-C is not so strong as for SCA due to the different water uptake capability and composition of the two materials [31, 32]. It was also possible to observe the intensity of the typical SEVA-C and SCA peaks decreasing as compared to un-coated substrates, which reinforces the existence of a Ca–P layer on the surfaces of these polymers.

As compared to biomimetic coatings, with this technology it was possible to reduce the induction period necessary for the apatite formation due to the fact that with biomimetic coating the first treatment (induction period for apatite nucleation) is at least 24 h for the most common polymers such as poly(ethylene terephthalate) (PET) and polyethylene (PE) [33, 34]. Eventually, that time can be decreased to 6 h if these substrates are previously subjected to glow discharge treatment in O_2 gas for 30 s [12, 34].

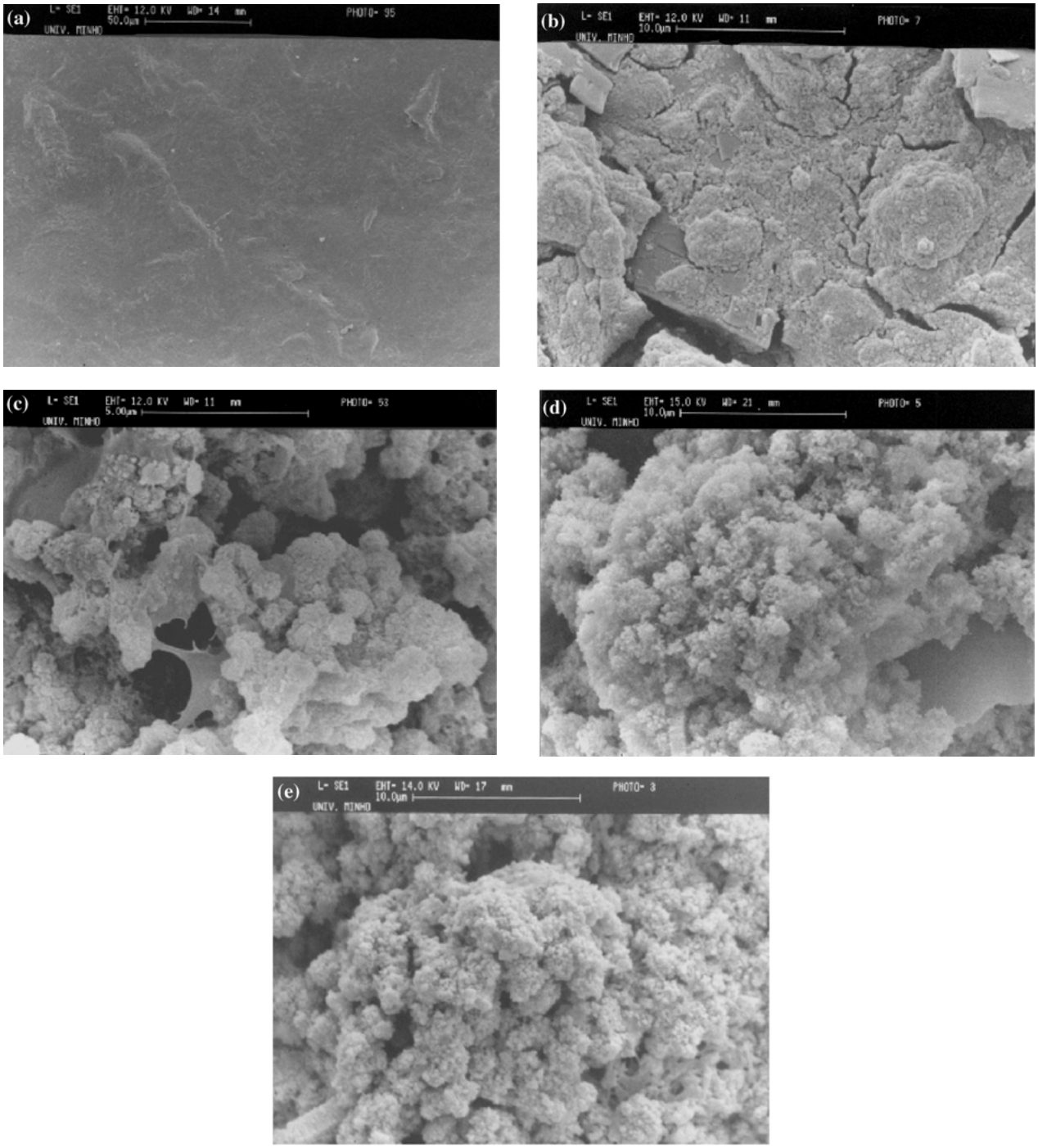

Figure 3 SEM photographs of Ca–P coating produced on the surfaces of SEVA-C substrate. Sample before coating (a), after (b) 60 min in alkaline bath, then immersed for (c) 14 days in SBF solution and after 60 min in acid bath (d) and then immersed for (e) 14 days in SBF solution.

3.2. Bioactivity evaluation

The bioactivity tests showed that after different immersion periods in SBF solution it was clear the bioactive nature of the Ca–P coatings (see Figs. 1–3). The Ca–P film became more compact and dense as they gradually grow. At higher magnifications it was possible to observe the morphology of this film that evidenced a finer structure, where the needle like crystals are agglomerated to produce the so called cauliflower like morphology as it is shown in Figs. 1(d) and 2(e).

For SCA substrates, the thickness of the film obtained with the acid coating is around 3 μm after 14 days in SBF solution, being the needle like morphology also clear (see Fig. 2(f)).

The SEM observations suggest that there are some differences in the morphologies of the Ca–P formed on the surface of the three polymeric substrates as it is shown in Figs. 1–3. For example, for HMWPE substrate only after 14 days of immersion in SBF it was possible to observe the formation of a Ca–P film (Fig. 1(c)). For shorter times only Ca–P nucleus were observed (Fig. 1(b)). Theses differences are associated with the water uptake capability of SEVA-C and SCA, which allowed the material to absorb higher quantities of Ca^{2+} ions from the auto-catalytic bath. This gives raise to finer morphologies and higher adhesion of the Ca–P films to the substrate, as it has been observed before for biomimetic coatings [23, 25]. Also, these results might

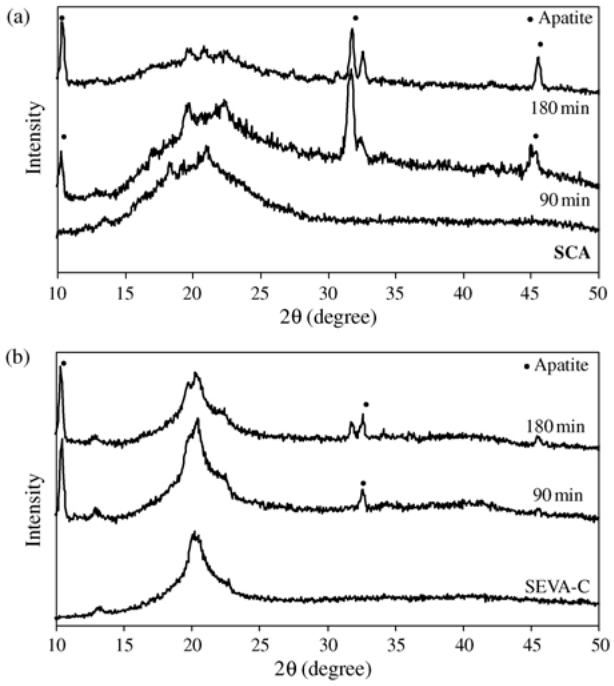

Figure 4 XRD patterns of film formed on (a) SCA and (b) SEVA-C substrate after 0, 90 and 180 min, immersion in alkaline bath.

be attributed to the absence of the polar groups on HMWPE substrate, which decrease the number of points at which the apatite nuclei are attached to the respective substrate.

By EDS analysis it was possible to observe that Ca/P ratios on the Ca–P films formed on HMWPE, SEVA-C and SCA coated substrates were in the 1.5–1.7 range, i.e., between tricalcium phosphate (TCP) and hydroxyapatite. These values were determined using well-established sub-routines for EDS semi-quantitative analysis.

Fig. 5 shows the ICP measurements of Ca and P concentrations as a function of immersion time in SBF solution for un-coated and coated substrates used in this work. The un-coated substrates (not subjected to the auto-catalytic treatment) could not in any case induce Ca–P layer deposition when immersed in SBF, which is an evidence that these materials are non-bioactive and present a bioinert behavior when immersed in SBF solution. This fact has been proved in previous works [23, 25].

As it is shown in Fig. 5, no changes was observed on the Ca and P concentration in the solution for non pre-coated substrates. However, for the auto-catalytic coated substrates, in the first day of immersion in SBF there was a slight increase of Ca and P concentration in the solution, which indicates some dissolution of the coating. Then as the immersion time in SBF solution increased, there was a decrease of the Ca and P concentration in the solution, indicating that these ions had been consumed during precipitation and growth of the bioactive Ca–P layer on the surface of the substrate. These results clearly indicate a bioactive character of the produced coating.

4. Conclusions

A new auto-catalytic deposition methodology to produce Ca–P coatings on the surface of different polymers was successfully developed. By using this new coating route

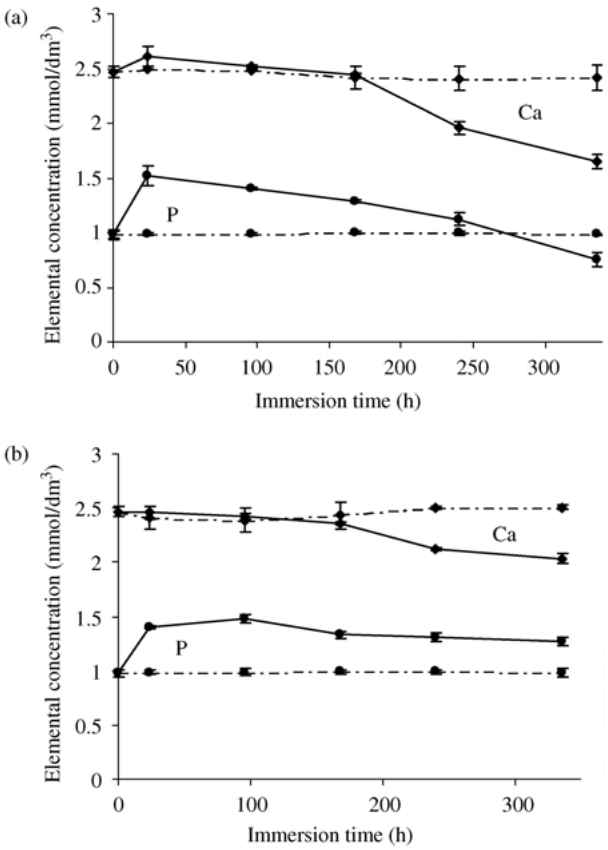

Figure 5 Evolution of Ca and P elemental concentration (ICP results) in the SBF solution as a function of immersion time for un-coated (– · –) and acid coating (—) on (a) HMWPE and (b) SCA surfaces.

it was possible to produce a Ca–P film on the surface of three types of polymers, polyethylene (HMWPE), and two types of starch based polymers (SCA and SEVA-C). Furthermore, one of the main advantages of the proposed technology is the possibility of the reducing the induction period necessary for the apatite formation (for instances as compared to biomimetic coatings) as well as to produce an adherent bioactive film on the surface of both biodegradable and bioinert polymers. The developed route seems to be a very promising and simple methodology for being used as a pre-implantation treatment to be applied to different types of materials, including polymers and eventually metals, previously to their clinical application. The proposed auto-catalytic deposition route is being optimized by studying the ideal bath compositions and the respective operating conditions in an attempt to further control the morphology and microstructure of the coatings, as well as its adhesion to the substrate.

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