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Electrophoretic deposition of polyetheretherketone (PEEK) and PEEK/Bioglass® coatings on NiTi shape memory alloy wires

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Abstract Polyetheretherketone (PEEK) and PEEK/Bioglass® coatings were produced on shape memory alloy (NiTi, Nitinol®) wires using electrophoretic deposition (EPD). Best results were achieved with suspensions of PEEK powders in ethanol in the range (1–6 wt%), using a deposition time of 5 minutes and applied voltage of 20 Volts. EPD using these parameters led to high quality PEEK coatings with a homogeneous microstructure along the wire length and a uniform thickness of up to 15 μm without development of cracks or the presence of large voids. Suspensions of PEEK powders in ethanol with addition of Bioglass® particles (0.5–2 wt%) (size < 5 μm) were used to produce PEEK/Bioglass® coatings. Sintering was carried out as a post EPD process in order to densify the coatings and to improve the adhesion of the coatings to the substrate. The sintering temperature was 340 °C, sintering time 20 min and heating rate 300 °C/h. Sintering led to uniform and dense PEEK and PEEK/Bioglass® coatings without any cracks. The bioactive behaviour of PEEK/Bioglass® composite coatings was investigated by immersion in acellular simulated body fluid (SBF) for up to two weeks. As expected, hydroxyapatite crystals formed on the surface of the coated wires after 1 week in SBF, confirming the bioactive character of the coatings. The results have demonstrated for the first time that EPD is a very convenient method to obtain homogeneous and uniform bioactive PEEK and PEEK/Bioglass® coatings on Nitinol® wires for biomedical applications.

Introduction

Although considerable research efforts are dedicated to examine the biocompatibility of materials used as implants or prosthesis, there is still some concern regarding the use metallic biomaterials in the body, i.e. due to possible corrosion or leakage of ions or corrosion products [1]. It is the challenge for materials scientists to develop tailored, optimised biomaterials in terms of resistance to corrosion, high biocompatibility and in some applications, high bioactivity, i.e. materials exhibiting strong bonding to human tissues [2]. One solution is to combine the good mechanical properties of metals with the biocompatible properties of a non-metallic (organic or inorganic) coating, which in some cases can be also bioactive [3]. Non-toxic, biocompatible and stable polymers are available, which combine outstanding chemical and hydrolysis resistance, high strength and excellent tribological properties with proven biocompatibility. These polymers are hence ideally suited for medical device applications [4]. Polyetheretherketone polymer (PEEK) is one of the available polymers exhibiting high biocompatibility. It has been increasingly considered as a biomaterial in a wide variety of biomedical applications including dental, orthopaedic and cardiovascular devices [5, 6].

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PEEK is essentially a bioinert material, eliciting neither a positive or negative response in the body. Thus surface activation of PEEK has been proposed to enhance its bioactivity for wider use in the fields of orthopaedics and dentistry [7]. Surface modification studies have included the addition of topographical features or coatings to the PEEK surface to increase cellular response. Another possibility to enhance the bioactivity of polymers is to incorporate bioactive inorganic particles, for example hydroxyapatite or bioactive glass particles [8], however this approach has not been extensively investigated previously for PEEK [5].

In the group of inorganic bioactive materials, bioactive glasses, e.g. commercially available 45S5 Bioglass® [2], are the materials of choice for several applications where a strong bonding between the implant and the surrounding tissue is required. These applications include bioactive coatings on a variety of metallic substrates for orthopaedic applications [9].

Nickel–titanium shape memory alloys (SMA) were introduced to medical applications in the 1970’s, because of their outstanding properties like the shape memory effect, superelasticity, good resistance to corrosion and high biocompatibility [10]. Currently shape memory alloys are used in many different medical applications, including dental implants, cardiovascular devices, surgical instruments and orthopaedic implants [11, 12]. The unique properties of SMA are due to a reversible solid–solid phase transformation between a high symmetry austenitic crystal structure and a lower symmetry martensitic structure. Various techniques have been introduced for surface treatment and corrosion resistance enhancement of NiTi implants with different degree of success, including chemical passivation, anodic oxidation, electropolishing, thermal oxidation, laser surface melting, nitriding, plasma ion implantation and sol-gel hydrothermal coating [11–13]. One desirable feature for a coating or surface treatment is to involve a low processing temperature, which should be ideally <400 °C to minimise any microstructural changes in the base alloy.

An effective, low temperature and low cost procedure to achieve high-quality coatings on metallic substrates is electrophoretic deposition (EPD) [14, 15]. Electrophoretic deposition consists of two steps, electrophoresis and deposition. Electrophoresis is the motion of charged metallic, ceramic or polymeric particles suspended in a liquid. Deposition is the coagulation of the particles to a dense mass at one of the electrodes. Usually a heat treatment as a post EPD processing step is required for the densification of the deposited material. The heat-treatment involves lower temperatures if a polymer rather than a ceramic is the deposited material.

EPD is a highly flexible method for many applications, and it has been extensively used to deposit inorganic bioactive coatings, e.g. hydroxyapatite, Bioglass®, on a variety of metallic substrates for orthopaedic applications, including stainless steel, titanium alloys and shape memory alloys [16–19]. EPD offers many advantages as a coating technique, like simplicity of equipment, cost effectiveness and environmentally friendly processing, especially when aqueous suspensions are used, as discussed elsewhere [15]. There has been however only limited work on the use of EPD for the deposition of biopolymer coatings on conductive substrates. To the authors’ knowledge, only the previous investigation by Wang et al. [20] has explored the possibility of using EPD for deposition of PEEK particles onto planar conductive substrates. They successfully deposited PEEK coatings of adequate mechanical properties and uniformity on carbon substrates from ethanol suspensions. Electrophoretic deposition of PEEK on metals relevant for biomedical applications, however, has not been reported. Moreover the electrophoretic co-deposition of PEEK particles and an inorganic particulate inclusion phase for production of composite coatings has not been investigated to date. In fact a literature search has indicated that there has been no previous attempt to fabricate composites combining PEEK and bioactive glasses.

The main focus of this research was therefore to combine the outstanding properties of nickel-titanium shape memory alloys with the ones of PEEK and Bioglass® to produce protective and bioactive coatings on Nitinol® wires. EPD was the coating method of choice because of its simplicity and cost effectiveness. After deposition, PEEK and PEEK/Bioglass® coated wires were subjected to heat-treatment to fuse the polymer on the wires and to increase the polymer/metal adhesion strength.

Experimental

Materials

Nitinol® wires were obtained from Memory Metalle GmbH (Weil am Rhein, Germany) and had a nominal diameter of 0.27 mm. This material is a standard alloy for body temperature actuator applications. The nominal austenite finish temperature was 28 °C and the surface was oxidised. This wire was composed of 50.2–50.4% nickel (49.6–49.8% titanium). The mechanical and physical properties [21] are illustrated in Table 1.

Table 1 Mechanical and physical properties of Nitinol® wire alloy used [21]

Density [g/cm^3]	Ultimate tensile strength (fully annealed) [MPa]	Young's Modulus (austenite) [GPa]	Thermal expansion coefficient (austenite) [10^{-6}/°K]
6.45	Approx. 900	Approx. 70–80	10–11

PEEK is a semi-crystalline thermoplastic polymer for high performance and biomedical applications [5, 6]. The polymer used in this study (Victrex® PEEK powder 150XF) was obtained from Victrex® Manufacturing Ltd. (South Yorkshire, UK) and it was delivered in the form of a fine powder with a mean particle size of 20 μm. Thermal, mechanical and physical properties of Victrex® PEEK powder [22] are given in Tables 2 and 3.

For preparation of composite coatings, bioactive glass powder of type 45S5 Bioglass® was used. The composition of the glass in wt.% is: 45% SiO_2, 24.5% Na_2O, 24.5% CaO and 6% P_2O_5, which is the first bioactive glass composition synthesised in 1971 by Hench et al. [2]. The mean particle size of the glass powder used was <5 μm but particles in the range 1–50 μm were present. This glass was chosen due to its long history of successful biomedical use in dentistry, orthopaedic applications, as bone filler material [2] and, most recently, in bone regeneration and tissue engineering strategies [8, 23]. To the authors' knowledge, however, this is the first time this bioactive glass has been used in combination with PEEK to produce bioactive coatings.

Table 2 Thermal properties of Victex® PEEK powder 150XF [22]

Glass Transition Temp [°C]	Melt Point [°C]	Thermal expansion [10^{-6}/°C]	Thermal expansion [10^{-6}/°C]
143	343	10.8 > TG	4.7 < TG

Table 3 Mechanical and physical properties of Victrex® PEEK powder 150XF [22]

Density [g/cm^3]	Tensile strength [MPa]	Tensile strength [MPa]	Youngs's modulus [GPa]	Youngs's modulus [GPa]
1.26–1.32	100 (at 23 °C)	12 (at 250 °C)	4 (at 23 °C)	0.3 (at 250 °C)

Experimental Setup and EPD

PEEK powder suspensions (1–6 wt% solids loading) were prepared using ethanol as solvent. Suspensions were prepared adjusting the pH at 5.5, because at this pH the zeta potential of the PEEK particles acquires a high value [20]. Hydrochloric acid (0.1 M) was used as dispersion agent at a concentration of 0.5 wt%. Small amounts of sodium hydroxide solution were used to adjust the pH. Glass jars with a diameter of 55 mm containing the suspensions were placed into an ultrasonic bath (Sonomatic®, Langford Ultrasonics, UK) for 30 min in order to disperse the particles and to dissolve agglomerates within the suspension. Moreover, the suspension was stirred magnetically for 5 min before each EPD experiment to disperse the particles and to avoid sedimentation of PEEK particles ensuring homogeneous particle dispersion.

All EPD experiments were carried out at constant ambient temperature. After approximately 10 experiments using the same suspension, the pH-value of the suspension was measured using a Jenway 3510 pH-meter (Barloworld Scientific Ltd. T/As Jenway, Essex UK). All EPD experiments were carried out under constant voltage conditions. The Nitinol® wires cut to a length of 20 mm were attached with adhesive tape to a stainless steel plate which served as deposition electrode. The counter electrode was made up of a stainless steel band, which was formed to a tube-like geometry with a diameter of 45 mm. The electrodes were cleaned with acetone before each experiment to enhance adhesion of the coating. The Nitinol® wires and the tube-like counter-electrode were positioned centrically in the glass jar, to produce a symmetrical electric field, as described elsewhere [19]. A TTi EL 561 equipment (Thurlby Thunder Instruments, UK) was used as a DC power supply and the current was measured with a TTi 1906 computing multimeter (Thurlby Thunder Instruments).

The applied voltage, deposition time and suspension concentration were the parameters chosen to be varied in order to optimize the coating quality by a trail an error approach. The parameters were varied initially in a wide range to gain knowledge about the influence of each parameter on the reproducibility of the EPD process and to optimize the parameters in a second step with the aim to improve the coating quality. The applied voltage was varied between 3 Volts and 50 Volts. The distance between the electrodes remained constant during all experiments at 20 mm. Deposition time was varied between 2 min and 20 min. Suspensions were prepared with PEEK concentrations in the range 1–6 wt% PEEK powder. After optimisation of EPD parameters for PEEK suspensions, experiments using diphasic suspensions, e.g. containing PEEK and Bioglass® particles in ethanol were carried out. The concentration of Bioglass® in a base suspension of 6 wt% PEEK was varied between 0.5 and 2 wt%. The pH of the suspensions was adjusted to 5.5 by addition of sodium hydroxide solution, as mentioned above. After electrophoretic deposition the samples were extracted carefully and left to dry in a desiccator at room temperature.

Coating sintering

Sintering was carried out in order to densify the PEEK and PEEK/Bioglass® deposits and to improve the adhesion of the coatings to the NiTi wires. The samples were heat-treated in a tube furnace (Carbolite, Hope Valley, UK) at 340 °C for 20 min using heating and cooling rates of 300 °C/h. In order to reduce possible oxidation of the wire, sintering was carried out in argon atmosphere.

Assessment of bioactivity in simulated body fluid (SBF)

Selected PEEK/Bioglass® coated NiTi wires were immersed for periods of up to 14 days in simulated body fluid [24]. This is the usual technique employed in the biomaterials field to assess bioactivity of materials, i.e. the ability of the material to nucleate and deposit hydroxyapatite crystals on their surfaces is examined [2, 8, 24]. The SBF composition and method of preparation were those described by Kokubo et al. [24]. Briefly, the SBF which has inorganic ion concentrations similar to those of human extracellular fluid, was prepared by dissolving respective amounts of reagent chemicals (all purchased from Sigma, Steinheim, Germany) of NaCl, $NaHCO_3$, KCl, $K_2HPO_4·3H_2O$, $MgCl_2·6H_2O$, $CaCl_2·2H_2O$, and Na_2SO_4 into distilled water. The SBF was adjusted to physiological pH (pH 7.25) by HCl and buffered by tris(hydroxyl-methyl) aminomethane. Individual coated wires (length 20 mm) were immersed in 50 mL of SBF in clean conical flasks, which had been previously rinsed with hydrochloric acid and deionised water. Lids were placed on the flasks to form an airtight seal, preventing contamination. The flasks containing the specimens were placed inside a chamber which was maintained at a temperature of 37 °C. The wires were left in immersion in SBF for varying time periods of 3, 7 and 14 days. The SBF was changed every three days as the cations' concentration decreases during the in vitro studies due to chemical change of the samples, as discussed below. After immersion in SBF the samples were gently rinsed with deionised water prior to drying for 48 h in a desiccator that maintained a temperature of 37 °C and controlled relative humidity of ~60%. Samples were subsequently placed in desiccator jars awaiting characterisation. Non coated wires (as-received) were subjected to the same treatment in SBF for comparison purposes.

Characterisation

Scanning electron microscopy (SEM) (JEOL 5610 SEM and LEO 1525 field emission gun, FEG-SEM) was used to analyse the microstructure of the coatings in as deposited condition, after sintering and after immersion in SBF for 7 and 14 days. The samples were coated with carbon or gold before the examination. SEM was mainly used to characterize the microstructure of the coating surface in terms of uniformity and to discover the presence of microcracks or pores. The analysis of cross sections gave information on the thickness as well as on the microstructural homogeneity of the coatings. After sintering, the surfaces of the coatings were observed to discover possible microcracks and to assess changes in the surface topography due to the sintering process. Characterisation of samples after immersion in SBF was carried out by SEM observations and EDX analysis. The goal of this characterisation was to assess the formation of (bone-like) hydroxypapatite on the surface of the wires at different immersion times in SBF, which serves as a qualitative indication of the in-vitro bioactivity of the materials. For SEM, samples were gold-coated and observed at an accelerating voltage of 15–25 kV.

Results and discussion

Microstructural characterisation of PEEK and PEEK/Bioglass® coatings

In total 50 experiments were carried out for optimisation of the EPD parameters. The production of PEEK coatings on Nitinol® wires by EPD was straightforward. Most applied parameters for EPD led to good quality coatings. Variations of the EPD parameters resulted in changes in the quality of the coatings, in terms of macroscopic and microscopic uniformity and homogeneity, only if the parameter variations were rather large.

The uncomplicated electrophoretic coating process can be ascribed to the high stability of the PEEK suspensions in ethanol prepared for this investigation, which exhibited very little sedimentation of the PEEK particles during EPD. In agreement with the previous investigation by Wang et al. [20], working at pH = 5.5 has ensured a high zeta potential of PEEK particles in ethanol and thus high colloidal stability of the suspension. At the working pH the particles had a negative charge, hence they migrated to the anode upon application of the external electric field. Moreover addition of small amounts of hydrochloric acid as a dispersion agent increased the conductivity of the PEEK suspension leading to high rate of deposition.

It was observed that changes in the applied voltage needed to be large in order to have an effect on the EPD process. It was found that the quality of the PEEK coating was appropriate in terms of macroscopic homogeneity and uniformity, if the applied voltage was about or above 20 Volts. Under these conditions, regardless the PEEK concentration in ethanol (in the range 1–6 wt%), the wire was covered entirely and uniformly with PEEK particles. This can be related to the high electric field in the suspension using these relatively high voltages, which is possible if non-aqueous suspensions, like ethanol based suspensions, are used.

If voltages below 15 Volts were applied, the quality of the coating was insufficient. The PEEK coating did not cover the wire entirely and the homogeneity of the surface was strongly reduced, even if long deposition times were used. Moreover, it could be observed that the coating thickness increased with increasing applied voltage, which is thought to be related to the higher deposition rate achieved at higher voltages. Figure 1 shows typical SEM images of PEEK coated Nitinol® wires obtained at 20 V at given suspension concentration and deposition times.

At the magnifications shown in Fig. 1(a, b) there are no apparent differences in the microstructure (topography) of the PEEK coatings obtained with suspensions of different PEEK concentration. Both samples exhibit a fairly homogenous microstructure of the coatings which were uniform throughout the complete length of the wire. Therefore a homogeneous and controlled deposition of PEEK particles on the wire can be confirmed at 20 Volts and 5 min deposition time. Observation of the coating structure at higher magnifications demonstrated that the suspension with higher PEEK concentration (6 wt %) led to a more uniform topography of the coating surface, as shown in Fig. 2. This is thought to be due to the high colloidal stability of the PEEK suspension which allows both small and large particles to be deposited and thus a more uniform coating is achieved. At a given voltage and suspension concentration, the thickness of the coatings was always found to be in the range of 5–15 μm, depending on the deposition time, which was varied between 1 and 5 min. It was found that very short deposition times (1 min) led to thin coatings, clearly due to insufficient time available to achieve an appropriate thickness. A deposition time of 5 min was found to be appropriate in order to achieve a suitable thickness of 15 μm.

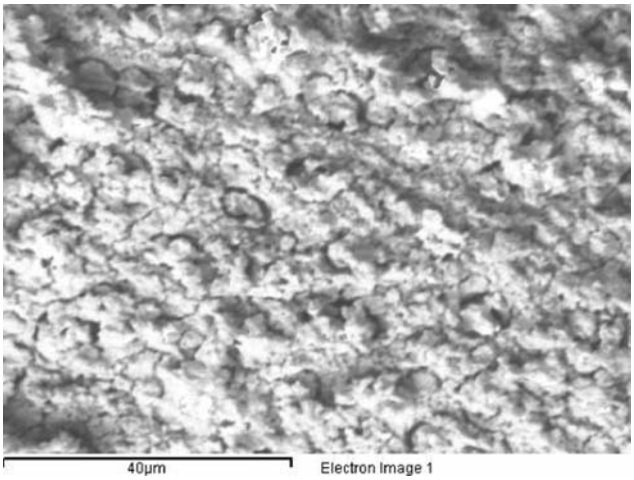

Fig. 2 SEM image of a PEEK coated Nitinol® wire, produced using optimised EPD parameters: Voltage = 20 Volts, time = 5 min, PEEK concentration in ethanol suspension = 6 wt%

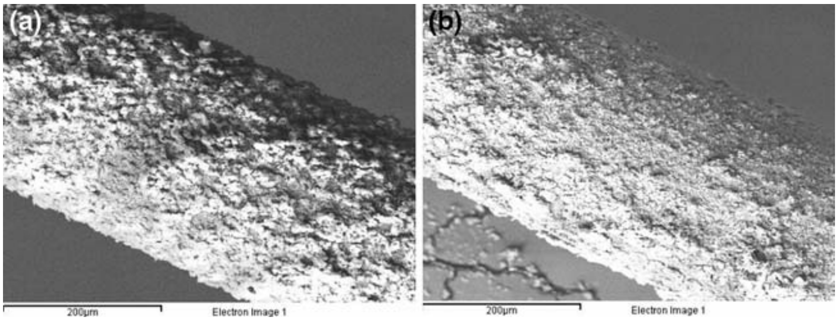

Fig. 1 SEM images of PEEK coated Nitinol® wires obtained by electrophoretic deposition at 20 V, deposition time 5 min and different PEEK concentrations in ethanol: (**a**) 2.5 wt% and (**b**) 6wt%

It is apparent in Figs. 1 and 2 that the surface of the coatings is smooth and uniform. It can be seen that the PEEK coating completely covers the Nitinol® wire surface. The continuity of the coating was confirmed by energy dispersive X-ray (EDX) analysis. Although the surface structure is homogeneous in general, some little gaps or areas of the coating could be found, where the deposited PEEK material was thinner compared to the coating in the surrounding area. However it was assumed that due to viscous flow of the polymer during the following sintering step, the gaps or insufficiently covered parts of the wire would be filled, as discussed below. Furthermore, there are no large cracks or pores visible, indicating that the drying process in air at room temperature was adequate.

A related important result is the fact that no cracking or flaking of the coating was observed, even if the coated wire was severely bent after the coating deposition. Figure 3 shows a SEM image of a bent unsintered PEEK coated Nitinol® wire. It is apparent that in the region where the stress is at a maximum level and therefore cracking could be expected, the coating is still homogeneous and covering the wire completely, confirming the high adhesion strength between Nitinol® and PEEK, even before heat-treatment (sintering). This result is significant because it confirms various possible uses of PEEK coated wires in a variety of applications where large deformation of the shape memory alloy wire is needed.

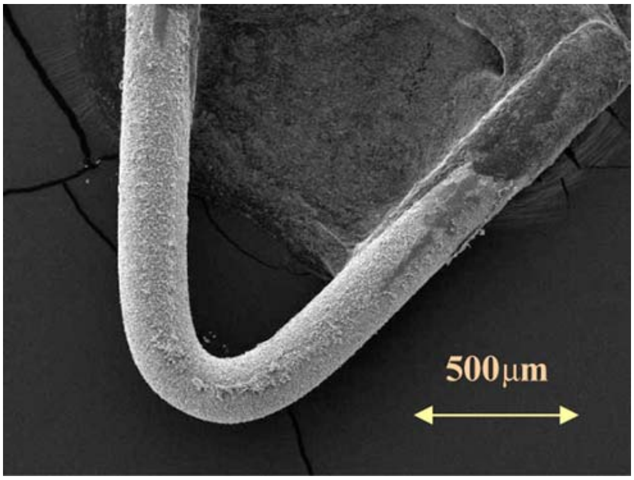

Fig. 3 SEM image of a strongly bent, unsintered PEEK coated Nitinol® wire demonstrating the structural integrity of the coating. (EPD Parameters: Voltage = 20 Volts, time = 5 min, PEEK concentration = 6 wt%)

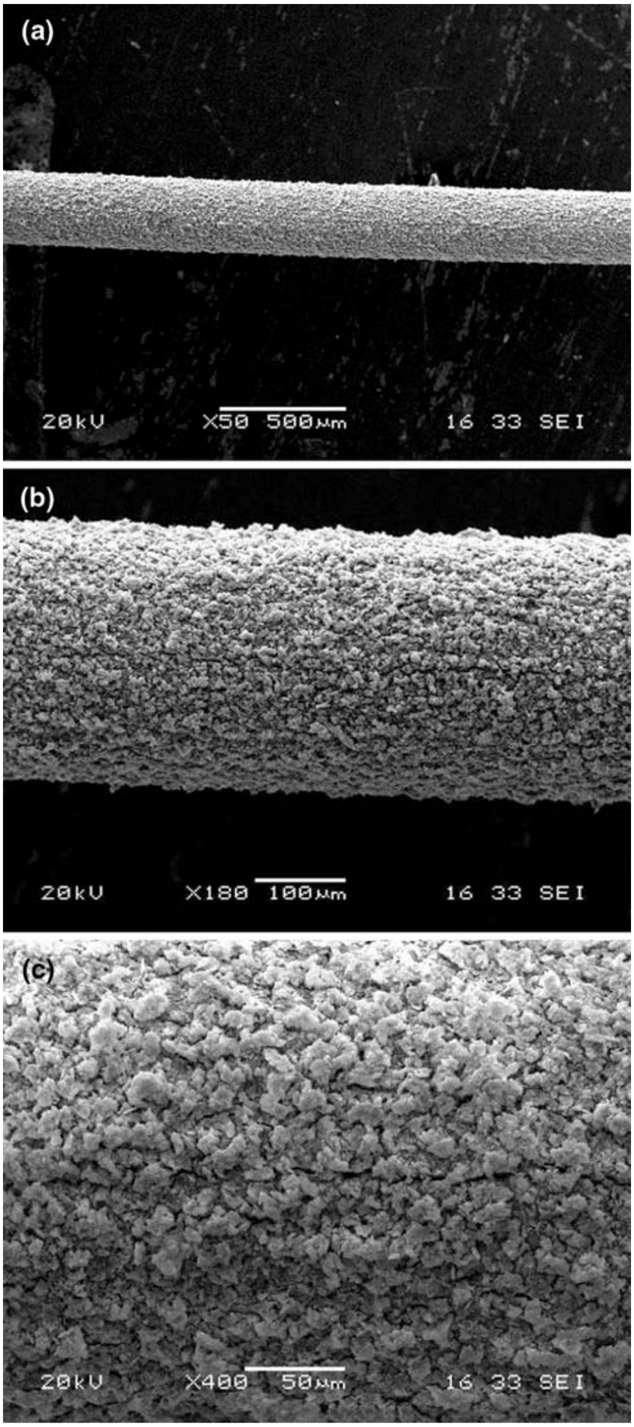

Fig. 4 SEM images of PEEK/Bioglass® coated NiTi wires at (**a**) low, (**b**) medium, (**c**) high magnification showing the high uniformity of the coating structure achieved. (EPD Parameters: Voltage = 20 Volts, time = 5 min, PEEK concentration = 6 wt%, Bioglass® concentration = 1wt%)

Figure 4 shows typical images of PEEK/Bioglass® coated NiTi wires obtained at 20 V and 5 min deposition time at different magnifications. The concentration of Bioglass® particles was 1 wt% in a base suspension containing 6 wt% PEEK. These images show that the EPD process leads to the formation of a very homogeneous PEEK/Bioglass® coating covering the wire uniformly. The nominal Bioglass® content of the composite coating, as determined from the suspension concentration, is ~15 wt% (51 vol.%), however the exact concentration of Bioglass® in the actual deposited coatings was not determined. It is possible that due to some sedimentation of Bioglass® particles during EPD, the actual Bioglass® concentration in the composite coatings is <15 wt%. The different glass particle sizes of the starting powder are visible in Fig. 4, this has led to a close packing of the Bioglass® and PEEK particles deposited by EPD. Although the coating exhibits a rough surface, there is no apparent formation of microcracks or porosity. It is expected that viscous flow of the PEEK matrix will lead to the formation of a smooth surface upon sintering. Certainly the use of Bioglass® particles of smaller size would lead to coatings of smoother surface.

Coatings were sintered at 340 °C for 20 min using heating and cooling rates of 300 °C/h. These parameters were chosen following the results of Wang et al. [20]. Figure 5 shows SEM images of a PEEK coated and sintered Nitinol® wire. It is apparent that the surface structure of the sintered coating is homogeneous and fairly uniform as a result of the viscous flow of the polymer at 340 °C. The sintered PEEK coating covers the wire completely and no microcracks or large pores are visible. It can be concluded that the sintering parameters were properly chosen. Moreover, the sintered coating retained a high flexibility and hence the wires could be bent without flaking or cracking of the PEEK coating, as shown in Fig. 6. This behaviour was also confirmed for PEEK/Bioglass® coated wires. This behaviour indicates that the coatings developed here will not negatively affect the shape memory effect and superelastic strains of the Nitinol® wires.

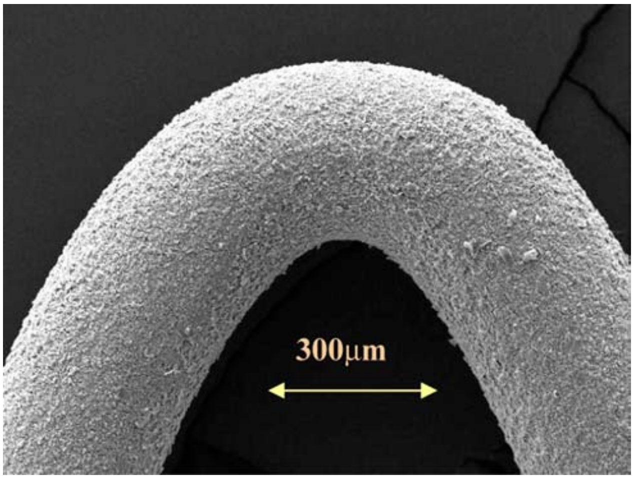

Fig. 6 SEM image of a sintered and bent PEEK coated Nitinol® wire, sintered at 340 °C for 20 min using a heating rate of 300 °C/h, showing structural integrity of the coating. (EPD Parameters: voltage = 20 volts, time = 5 mins, concentration of PEEK in ethanol = 6 wt%)

Bioactivity studies in SBF

The response of PEEK and PEEK/Bioglass® coated Nitinol® wires in contact with SBF was analysed using SEM and EDX. After 7 days of immersion in SBF small crystals developed on the surface of Bioglass® containing coatings, in regions close to Bioglass® particles. The morphology of the particles was similar to that reported for HA formed on dense Bioglass® discs immersed in SBF and tris buffer [25]. A SEM micrograph showing the coating surface after immersion in SBF for 7 days is shown in Fig. 7. EDX elemental analysis of the composition of the particles revealed a Ca/P ratio of ~0.68, which is close to the ratio of HA usually formed on Bioglass® surfaces upon immersion in SBF [8]. The concentration of the calcium phosphate particles formed on the surface of PEEK/Bioglass® wires increased with increasing time in SBF as expected. No similar precipitations were found on the surfaces of uncoated or PEEK coated wires for the period of incubation tested in this experiment. Since the size of the samples varied, a quantitative analysis indicating the amount of calcium phosphate formation in relation to the period of immersion in SBF could not be conducted. Certainly for the low period of immersion investigated, no degradation of the PEEK matrix was expected. It is therefore assumed that calcium phosphate formation

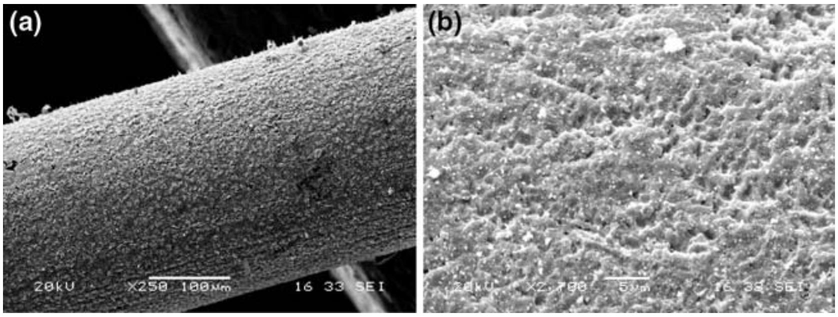

Fig. 5 SEM images of a sintered PEEK coated Nitinol® wire, sintered at 340 °C for 20 min using a heating rate of 300 °C/h at (**a**) low and (**b**) high magnification. (EPD Parameters: voltage = 20 Volts, time = 5 min, PEEK concentration of suspension = 6 wt%)

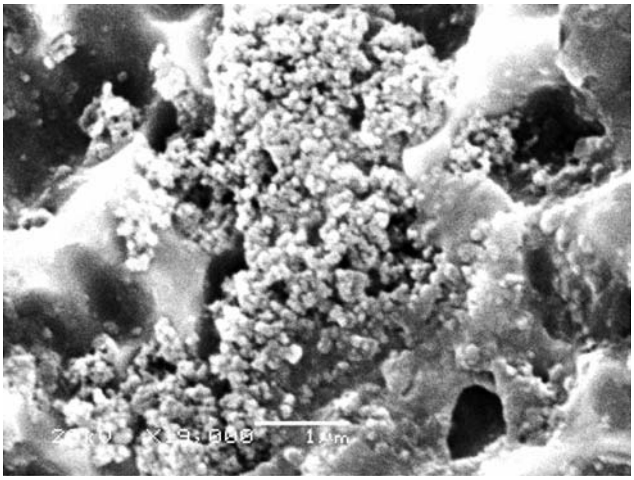

Fig. 7 SEM image of the surface of a PEEK/Bioglass® coated wire after 7 days of immersion in SBF, indicating formation of calcium phosphate crystals (identified by EDX)

started from Bioglass® particles exposed to the outer surface and thus in direct contact with the SBF. Similar behaviour has been found in several other systems where non-biodegradable polymers have been incorporated with bioactive particles, e.g. Bioglass® or HA, to induce bioactivity [26].

Conclusions

The work has shown that it is possible to use EPD to coat NiTi alloy wires with PEEK and with combined PEEK and Bioglass® particles. The achieved coatings had thickness in the range 10–15 μm which could be controlled by varying the deposition time under constant voltage (20 V) conditions. The coatings' surfaces were uniform and exhibited negligible microcracking or porosity. The length of wires that can be coated by EPD depends only on the dimensions of the EPD cell, the coating process can be easily scaled up to large sizes as appropriate. Upon sintering, the surface of the coatings became smooth due to viscous flow of the polymer. Both PEEK and PEEK/Bioglass® coatings exhibited high flexibility and good adhesion to the metallic substrate, which was tested qualitatively by manual bending wires and observing the coating microstructure. The PEEK and PEEK/Bioglass® coatings developed here should protect the NiTi substrate from corrosion in contact with body fluids impeding the leakage of ions. Besides this, in particular the Bioglass® containing coatings, being bioactive, will improve the bonding of bone or soft tissue to the implant improving thus the healing process. Long term studies in simulated body fluid are the focus of current research to investigate both the evolution of the coatings bioactivity and the effectiveness of the coatings as corrosion protective layers. Future studies will also concentrate on determining the roughness values of the produced coatings and their robustness under exposure to biofluids in realistic conditions.

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