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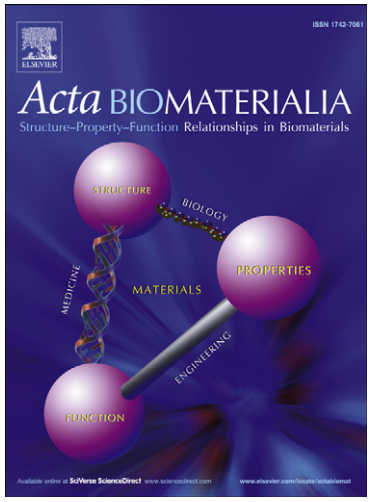

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Controlled Biomineralization of Electrospun Poly(ε-caprolactone) Fibers for Enhancing Their Mechanical Properties

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ABSTRACT

Electrospun polymeric fibers have been investigated as scaffolding materials for bone tissue engineering. However, their mechanical properties in particular stiffness and ultimate tensile strength cannot match those of natural bones. The objective is to develop novel composite nanofiber scaffolds by attaching minerals to polymeric fibers using an adhesive material mussel-inspired protein-polydopamine as "superglue". Herein, we report for the first time the use of dopamine to regulate mineralization of electrospun poly(ε-caprolactone) (PCL) fibers for enhancing their mechanical properties. We have examined mineralization of PCL fibers by adjusting the concentration of HCO_3^- and dopamine in the mineralized solution, reaction time, and surface composition of fibers. We have also examined mineralization on the surface of polydopamine-coated PCL fibers. We have demonstrated the control of morphology, grain size and thickness of minerals deposited on the surface of electrospun fibers. The obtained mineral coatings render electrospun fibers with much higher stiffness, ultimate tensile strength and toughness, which could be closer to the mechanical property of natural bone. Such a great enhancement of mechanical properties for electrospun fibers through mussel protein mediated mineralization was not seen in previous reports. Further, this study could also be extended to fabricate other composite materials for better bridging the interfaces between organic and inorganic phases.

Keywords: Electrospinning, Polydopamine, Fibers, Surface modification, Coating, Calcium phosphate

1. Introduction

A great advancement in development of bone scaffolds with various compositions and structures have been achieved using many techniques. Among them, electrospinning has been attracted much attention for fabrication of nanofiber scaffolds for use in bone tissue regeneration in that a non-woven mat of electrospun nanofibers can serve as an idea scaffold to mimic the extracellular matrix for cell attachment and nutrient transportation owing to its high porosity and large surface-area-to-volume ratio [1, 2]. Some studies have demonstrated that electrospun fibers of a polymer alone can serve as bone tissue engineering scaffolds and enhance bone regeneration to some extent. Other studies have suggested the use of some inorganic materials like bioactive glasses which have been processed into fibers or tubes by electrospinning technique for bone tissue regeneration [3-5]. However, the mechanical property of pure polymeric scaffolds is far from that of natural bone. And the nanofibers/tubes made of inorganic materials are very brittle and hard to manipulate. Recent efforts have been focused on the development of composite nanofiber scaffolds which can better mimic the composition and further match the mechanical property of natural bone. Incorporating an inorganic phase material (e.g., hydroxyapatite, octacalcium phosphate), which is one of the compositions of natural bone or bone precursors, to an organic phase material (e.g., biodegradable polymeric nanofibers) is generally used to enhance the mechanical property of nanofiber scaffolds. Two approaches of incorporation of inorganic phase materials include encapsulating inorganic phase materials (e.g., hydroxyapatite nanoparticle, nanorods) inside polymeric nanofibers and depositing inorganic phase materials on the surface of polymeric nanofibers to form uniform coatings [6-11]. The encapsulation of inorganic materials could improve the mechanical property of fibrous materials. However, the fiber surface is not optimized being a support to maintain desirable cell-substrate interaction. Direct deposition of inorganic materials on the nanofiber surface can not only enhance its mechanical property but also provide favorable substrate for cell proliferation and osteogenic conduction [11, 12].

Towards this end, many studies have investigated the deposition of minerals on the electrospun polymeric fibers. In principle, the morphology and grain size of minerals deposited on the fibers can be tailored by controlling the composition of mineralized solution, surface charge of substrate, and surface chemistry properties. It was demonstrated that Mg^{2+}, HCO_3^- acted as inhibitors of crystal growth and by varying the concentrations of them, the morphology

and grain size of minerals can be controlled [11, 13, 14]. The dose-dependent effects of amelogenin were also found to significantly inhibit apatite crystal growth and cause the octacalcium phosphate crystals to change from a plate-like shape to a curved shape [15]. In the same study, it was shown that the presence of bovine serum albumin in the mineralized solution can greatly alter the plate-like octacalcium phosphate (OCP) crystals into a round-edged, curved shape, indicating a general inhibitory effect [15]. The presence of cetyl trimethylammonium bromide (CTAB), poly-L-aspartic acid (PASP), and polyacrylic acid (PAA) in the mineralized solution was examined on the formation of crystals of minerals as well [16-18]. The charged surface was also thought to be an important factor which greatly affects the nucleation of minerals onto the substrates. It was reported that the negative surface was favorable for the heterogeneous nucleation of calcium phosphate [19-21]. Separate studies additionally examined the influence of different surface functional groups (e.g., carboxyl, carbonyl, amino, hydroxyl groups) and their combinations of substrates on the nucleation and growth of crystals during the mineral deposition [9, 10].

However, minerals attachment to polymeric materials in micro-/nanoscale is a big challenge as they represent inorganic phase materials and organic phase materials, respectively, exhibiting significant difference in mechanical properties. Prior studies demonstrated the control of morphology and grain size of minerals deposited on the electrospun polymeric fibers to enhance mechanical properties (e.g., stiffness) to a certain extent. However, none of them attempted to enhance the mechanical property by attaching minerals to electrospun polymeric fibers with a "filler" or "glue". In the present work, we aim to develop novel composite nanofiber scaffolds by attaching minerals (inorganic phase) to polymeric fibers (organic phase) through an adhesive material mussel-inspired protein-polydopamine as a "glue". We hypothesized that these hybrid fiber scaffolds could have superior mechanical properties compared to the unmodified fibers.

2. Materials and methods

2.1. Fabrication of PCL Fibers

The electrospinning setup used in the present work was similar to those described in our previous publications [22, 23]. Poly(ε-caprolactone) (PCL) (Mw=80,000 g/mol; Sigma-Aldrich, St. Louis, MO) was dissolved in a solvent mixture consisting of dichloromethane (DCM) and N, N-dimethylformamide (DMF) (Fisher Chemical, Waltham, MA) with a ratio of 4:1 (v/v) at a

concentration of 10% (w/v). Polymer solution was loaded into a 10 mL plastic syringe with a 22 gauge needle attached and pumped at a flow rate of 0.5 mL/h using a syringe pump. The working distance between the tip of the needle and the collector was about 15 cm and a voltage of 12 kV was applied. Alignment of electrospun fibers was achieved by making use of a high-speed rotating mandrel as a collector.

2.2. Fabrication of Composite Fibers

Composite fibers were fabricated in two different ways as indicated in Figure 1. One was to coat mussel inspired protein on the surface of plasma-treated, electrospun fibers and subsequently perform biomineralization of polydopamine-coated fibers. Specifically, nanofiber materials were treated with plasma for 8 min and then immersed in 0.2 mg/mL dopamine·HCl in Tris buffer (pH 8.5) for 4 h [22]. Polydopamine-coated nanofiber materials were then washed with DI water to remove excess monomer. Subsequently, the fiber mat was immersed in a supersaturated solution of 10 times concentration simulated body fluids (10×SBF) which was prepared from NaCl, $CaCl_2$, and $NaHPO_4{\cdot}H_2O$ in the presence of different amounts of $NaHCO_3$. The composition of 10×SBF was shown in Table 1. The ion concentrations in 10×SBF solution were 1 M of Na^+, 2.5×10^{-2} M of Ca^{2+}, and 1.0×10^{-2} M of HPO_4^-. The other is to directly biomineralize fibers in the 10×SBF solution in the presence of different amounts of dopamine and $NaHCO_3$. The minerals coated on electrospun fibers could be close to the composition of inorganic phase in native bone as mineralization in SBF mimics the mineral formation in human body.

2.3. Characterization of Fibers

The morphology of nanofiber scaffolds was characterized by scanning electron microscopy (SEM) (FEI, Nova 2300, Oregon). To avoid charging, polymer fiber samples were fixed on a metallic stud with double-sided conductive tape and coated with platinum for 40 seconds in vacuum at a current intensity of 40 mA using a sputter coater. SEM images were acquired at an accelerating voltage of 15 kV.

Fiber surface chemistry was examined by AXIS His X-ray photoelectron spectroscopy (XPS) (Kratos Analytical Inc., NY) and associated curve fitting software. For all samples, a survey spectrum was recorded over a binding energy range of 0-1100 eV using a pass energy of 80 eV.

In all cases, the survey spectra recorded the presence of Oxygen (O_{1s} 533 eV), Carbon (C_{1s} 285 eV), Calcium (Ca_{2p} 347.8 eV, Ca_{2s} 466.1 eV), Phosphorus (P_{2p} 133.7, P_{2s} 190.9 eV), and Nitrogen (N_{1s} 399 eV) at the surface.

2.4. Mechanical Test of Fibers

Fiber mats composed of uniaxially-aligned fibers were cut into sections and fixed onto a paper frame. Gauge length and width was set to 10 mm and 5 mm according to the frame size and the thickness of each sample, pre-determined by light microscope. Samples were mounted on a nano tensile tester (Nano Bionix, MTS, USA) and the edge of frame was cut before testing the fiber samples. Ten samples were stretched to failure at a low strain rate of 1%/sec at room temperature while related displacement and force values were recorded.

3. Results

3.1. Fabrication of Composite Fibers

In this work, we chose PCL as a model material because it is a biocompatible and biodegradable polymer that has been approved by FDA for certain human clinical applications [24]. We firstly coated electrospun PCL fibers with polydopamine following our recent study [22]. Specifically, PCL fibers were plasma treated and immersed in 0.2 mg/mL dopamine solution at pH 8.5 for 4 h. Morphology was similar between polydopamine-coated and un-coated fibers, with both fiber populations demonstrating consistent fiber diameters and limited surface roughness. Subsequently, we immersed polydopamine-coated PCL fibers in 10×SBF solution at 37^{o}C for 24 h in the presence of 0.01 M and 0. 04M $NaHCO_3$. Figure 2 shows the influence of $NaHCO_3$ concentration on the mineralization of polydopamine-coated PCL fibers. At a low concentration (0.01 M), large, thin, and plate-like minerals tended to be formed and loose structure of minerals was observed on the fibers (Figure 2, A and C). At a high concentration (0.04 M), smaller grain size of minerals displayed on the fibers and denser coating of minerals was seen compared to the samples fabricated in a low concentration of $NaHCO_3$ (Figure 2, B and D). The diameter of fibers in Figure 2D was around 1.2 μm. We also investigated the mineralization of polydopamine-coated PCL fibers at these two concentrations for longer period time (72 h). At low concentration, more plate-like minerals were seen on the surface of fibers and the fibrous morphology was still visible (Figure 3, A and B). In contrast, minerals coated on

the fiber surface in the presence of higher concentration of $NaHCO_3$ were dense and smooth (Figure 3, C and D). The fiber diameter was around 2.8 μm after 72 h coating.

In order to examine the influence of dopamine, we explored the mineralization of PCL fibers for 24 h in the presence of 0.04 M $NaHCO_3$ and a series of concentrations of dopamine (Figure 4). With increasing the dopamine concentration from 0.02 mg/mL to 0.2 mg/mL, there was no evident variation of mineral morphology and grain size (Figure 4, A-D). When the dopamine concentration reached 0.4 mg/mL and 1 mg/mL, the mineral morphology changed from plate-shape to nanorod-shape further to particle-shape (Figure 4, E and F). We also examined the influence of dopamine on the mineralization in the presence of low concentration of $NaHCO_3$. Plasma-treated PCL fibers were mineralized in 10×SBF solution containing 0.01 M $NaHCO_3$ and different amounts of dopamine for 2 h. In the absence of dopamine, few plate-like minerals were deposited on the PCL fibers (Figure 5, A and B). With increasing dopamine concentration, the morphology of minerals changed significantly (Figure 5). Also, more minerals were deposited on the fibers and the grain size of minerals decreased dramatically (Figure 5, E and F). We also examined the mineralization of PCL fibers in 10×SBF solution containing 0.04 M $NaHCO_3$ and 0.2 mg/mL dopamine at 37^{o}C for different times (Figure 6). The dense and smooth mineral coatings were observed. The fiber diameters increased with increasing the mineralization time.

3.2. XPS Characterization of Composite Fibers

Figure 7 shows a typical XPS survey spectrum of PCL fibers before and after mineralization. XPS wide scan in Figure 7A identified carbon and oxygen as the major constituents of PCL fibers, as expected. Figure 7, B-D, confirmed that the surface of mineral coated PCL fibers was primarily comprised of carbon, calcium, phosphorous, oxygen, and nitrogen. The spectra peaks for Ca2s, Ca2p, P2s, and P2p were originated from mineral coatings. N1s peak was originated from polydopamine. The surface composition was quantified in table 2. Ca/P ratio for mineralized samples is close to 1.67 the Ca/P ratio of hydroxyapatite. Figure 8 and Figure 9 show the typical C1s and Ca2p spectra of fibers before and after mineral coating. It is shown that the intensity of C1s band decreased while Ca2p and N1s bands increased.

3.3. Mechanical Test of Composite Fibers

Figure 10 demonstrates that the mineral coatings had functional consequences with regard to the mechanical properties of scaffolds composed of uniaxially-aligned composite fibers. HAP/PDA-PCL(24h) and HAP/PDA-PCL(48h) indicate the samples which were mineralized in the 10×SBF solution containing 0.04 M $NaHCO_3$ and 0.2 mg/mL dopamine for 24 h and 48 h. HAP-PDA-PCL(24h) indicates the samples were coated with polydopamine in 0.2 mg/mL dopamine solution at pH = 8.5 for 4 h prior to mineralization in the 10×SBF solution containing 0.04 M $NaHCO_3$ for 24 h. Evidently, the elastic modulus (Young's modulus) defined as the slope of stress-strain curve in the elastic deformation region increased dramatically from 86 MPa for PCL fiber samples to 459 MPa for HAP/PDA-PCL(24h) samples and further to 730 MPa and 768 MPa for HAP-PDA-PCL(24h) samples and HAP/PDA-PCL(48h) samples, respectively, suggesting the fibers become much stiffer after coating. It is seen that minerals can also significantly enhance ultimate tensile strength of fiber samples, increasing from around 36 MPa for PCL fibers to 157 MPa for HAP/PDA-PCL(24h) samples and further to 285 MPa for both HAP-PDA-PCL(24h) and HAP/PDA-PCL(48h) samples. The area covered under stress-strain curve is called toughness which is defined as the ability to absorb mechanical (or kinetic) energy up to failure. Accordingly, we quantified the area under the stress-strain curve and found that toughness showed the similar trend as elastic modulus and ultimate tensile strength. The toughness increased from 21.6 J/m^3 for PCL fibers to 80.5 J/m^3 for HAP/PDA-PCL(24h) samples and further to 137.4 J/m^3 for HAP/PDA-PCL(48h) samples. And the HAP-PDA-PCL(24h) samples showed the comparable toughness as HAP/PDA-PCL(48h) samples.

4. Discussion

Biomineralization of electrospun fibers provides a useful platform to fabrication of biomimetic materials for bone tissue engineering as they can recapitulate both the topography of extracellular matrix and composition of bone. Although previous studies examined the effect of different ions (e.g., HCO_3^- and Mg^{2+}) and proteins in the mineralized solution, surface charges and surface functional groups of substrates on the mineralization of electrospun fibers (e.g., morphology and grain size), the mechanical properties of obtained mineralized fibers was not optimum, which could be due to their ignorance of the usage of a 'glue' to bridge the two dissimilar materials between mineral coatings (inorganic phase) and polymeric fibers (organic phase). Our recent study demonstrated a uniform coating of polydopamine – a mussel inspired

protein which has been demonstrated to adhere to many different substrates ranging from metals, oxides, and polymers, on electrospun PCL fibers [22]. In the present study, we examined the mineral deposition to PCL fibers in two different approaches: mineralization of polydopamine-coated fibers and mineralization of PCL fibers in the 10×SBF solutions in the presence of dopamine. We demonstrated that the control of morphology, grain size, and thickness of mineral coating on PCL fibers with and without prior polydopamine coating. The mineral coatings result in significant improvement in mechanical properties (e. g., Young's modulus, ultimate tensile strength, and toughness) of fibers.

Based on the results of our research, we have proposed the two mechanisms for the mineralization of polydopamine-coated PCL fibers and plasma-treated PCL fibers in the presence of dopamine. In the first case, Ca^{2+} ions in the mineralized solutions bind to polydopamine and subsequently nucleation of CaP minerals takes place, followed by crystal growth. In the second case, Ca^{2+} ions bind to RCOO- on the surface of plasma-treated PCL fibers and nucleate and simultaneously dopamine polymerization occur in the mineralized solution. Other than the deposition to fibers, polydopamine could be deposited to certain facet of mineral crystals or whole surface of the minerals which may inhibit the growth of CaP mineral crystals along certain direction or inhibit the growth of CaP minerals.

Previous studies have demonstrated higher HCO_3^- concentration in the mineralized solution could lead to smaller grain size [11, 13, 14]. Our results are in line with those studies. However, the grain size of minerals deposited on the fibers was too large in some previous studies, resulting in elimination of fibrous morphology after mineral deposition and thus loss of ECM biomimetic capability of fiber scaffolds [12]. In the present study, fibers after mineral coating still remain the fibrous morphology. More importantly, for the first time we demonstrate that dopamine in the 10×SBF solution can affect the mineralization of PCL fibers. Increase of dopamine concentration in the mineralized solution can greatly decrease the grain size and change the morphology of minerals from plate to rod and further to particles. In addition, HCO_3^- ions were demonstrated not only affect the morphology and grain size during biomineralization but also the dissolution kinetics of minerals. A recent study showed that mineral coatings with increased HCO_3^- substitution presented more rapid dissolution kinetics in an environment deficient in calcium and phosphate, which can be used as carriers for regulation of growth factor release. The same principle should also be applied to our mineralized nanofiber scaffolds [25].

Recently, the macro-tensile measurements on nonwoven PCL fiber scaffolds showed a Young's modulus of around 3.8 MPa and the strain at break is at 170% [26]. In separate studies, it was demonstrated that the additive of nanohydroxyaptite could make the tensile strength increase from 0.81 MPa (0% nHAP) to 1.32 MPa (25% nHAP) and further to 1.44 MPa (50% nHAP) [27, 28]. Accordingly, the elastic modulus increased from 3.32 MPa to 3.5 MPa and further to 3.69 MPa. However, these studies demonstrated an increase of Young's modulus (stiffness) and ultimate tensile strength of electrospun PCL fibers to certain degree by creation of composite fibers or polymer blended fibers. Our recent study showed higher modulus, ultimate tensile stress, and toughness in the aligned nanofiber scaffolds relative to the random scaffolds [29]. In this work, we chose to test the mechanical properties of uniaxially-aligned PCL fiber scaffolds without and with mineral coatings, indicating biomineralization of fibers mediated by a mussel inspired protein can greatly enhance the mechanical properties including Young's modulus (≈ 9-fold), ultimate tensile strength (≈ 8-fold), and toughness (≈ 6-fold). Although our recent study demonstrated polydopamine coating could result in PCL fibers with a higher Young's modulus [22], PCL fibers without prior coating of polydopamine were directly mineralized in 10×SBF containing $NaHCO_3$ and dopamine showing great improvement compared to polydopamine-coated PCL fibers with regarding to Young's modulus and toughness.

In our future study, we will fabricate nanofiber scaffolds with gradations in both mineral content and fiber organization for repairing tendon-to-bone insertion site [30, 31]. In addition, adipose-derived stem cells (ADSCs) may be an optimal cell source for tendon-to-bone tissue engineering due to their pluripotency (i.e., they can differentiate into both tendon forming fibroblasts and bone forming osteoblasts), high proliferation rates, and capacity for matrix deposition [32-35]. We will also examine the response of ADSCs including attachment, proliferation and differentiation to the nanofiber scaffolds with dual gradients.

5. Conclusions

We have demonstrated the biomineralization of electrospun PCL fibers by making use of polydopamine as a filler or "bioglue" to bridge the minerals and polymeric fibers. We found that the morphology, grain size, and thickness of CaP mineral coating on PCL fibers can be readily controlled by adjusting the composition of mineralized solution, surface property of fibers, and

duration of mineralization. The mineral coating can greatly enhance the mechanical property of fibers, which may be useful as scaffolding materials for hard tissue engineering such as bone. This work also has significant implications for fabrication of other composite or hybrid materials.

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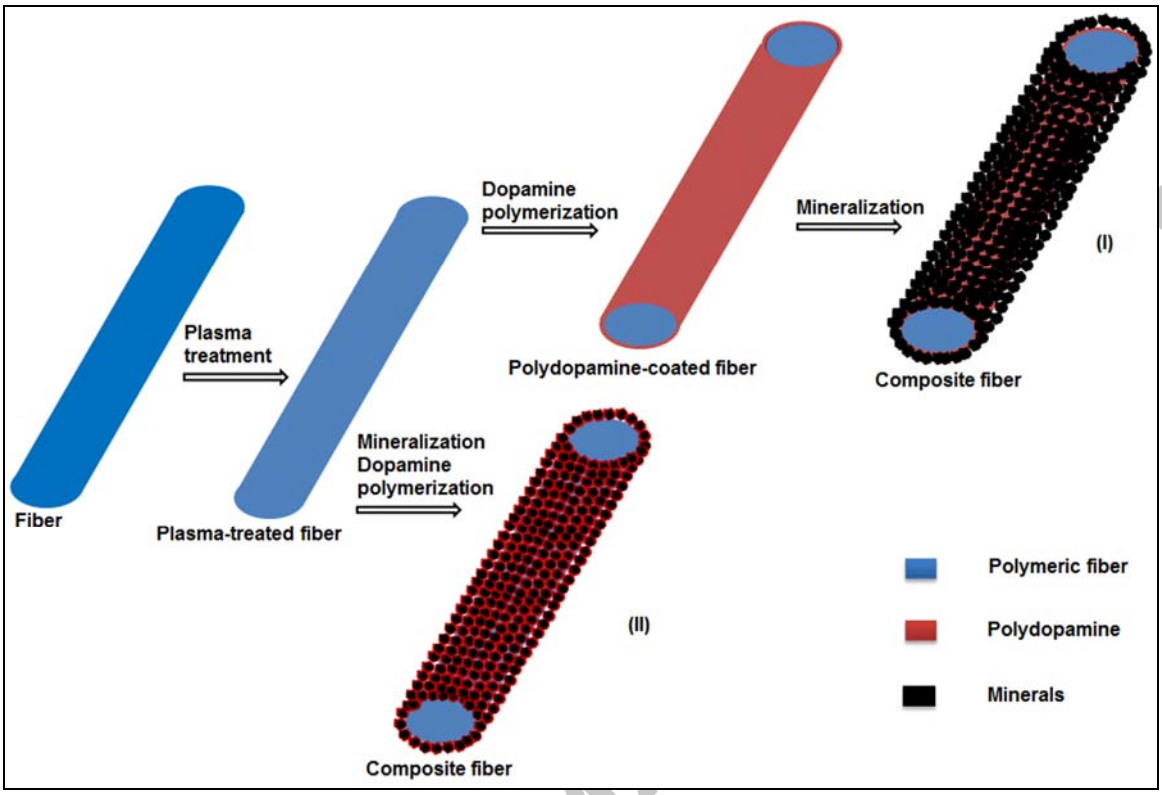


Fig. 1. Schematic illustrating the formation of composite fibers. Method I: Plasma-treated, electrospun PCL fibers were coated with polydopamine in the 0.2 mg/mL dopamine solution at pH 8.5 for 4 h and subsequently mineralized in the 10×SBF solution containing different amount of $NaHCO_3$. Method II: Plasma-treated, electrospun PCL fibers were directly mineralized in the 10×SBF solution in the presence of different amounts of $NaHCO_3$ and dopamine.

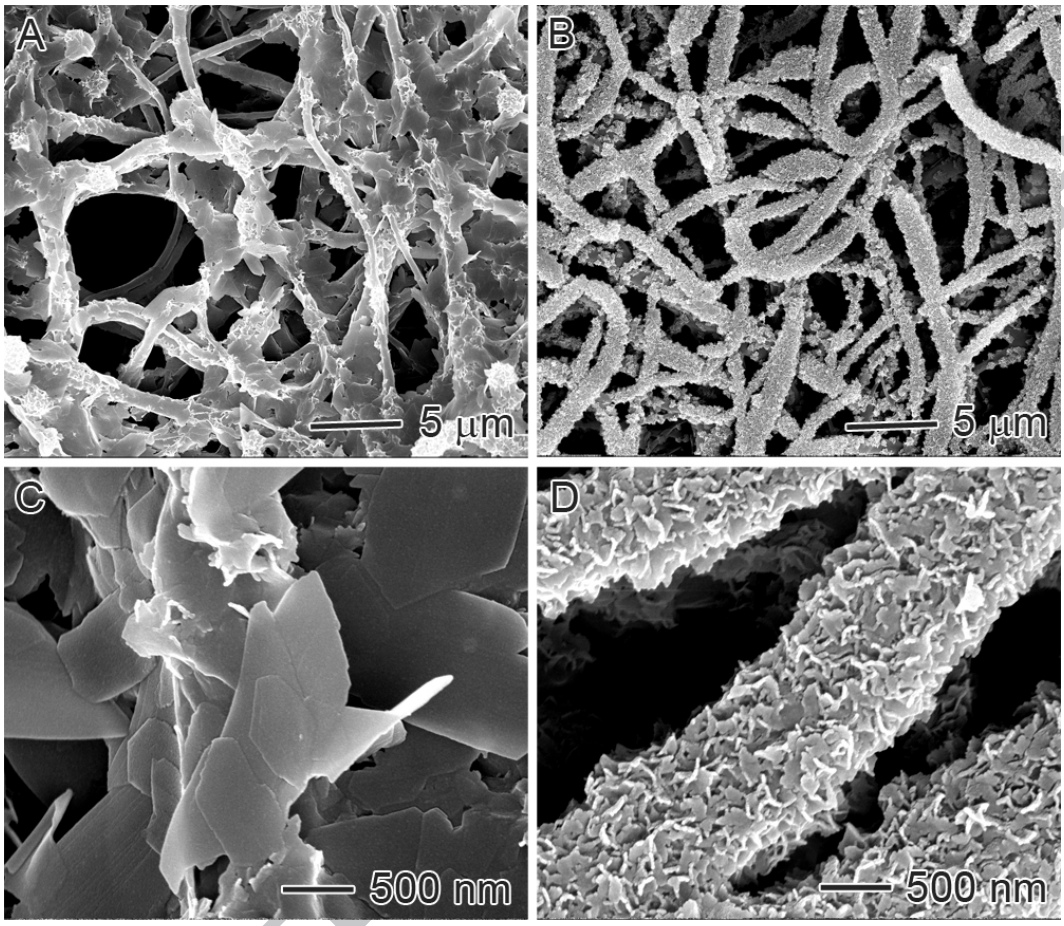


Fig. 2. SEM images of mineralized PCL nanofibers which were produced using method I shown in Fig.1. Plasma-treated, electrospun PCL nanofibers were coated with polydopamine for 4 h in 0.2 mg/mL dopamine solution at pH 8.5 and mineralized in the 10×SBF solution containing 0.01 M (A, C) and 0.04 M (B, D) $NaHCO_3$ at 37 °C for 24 h.

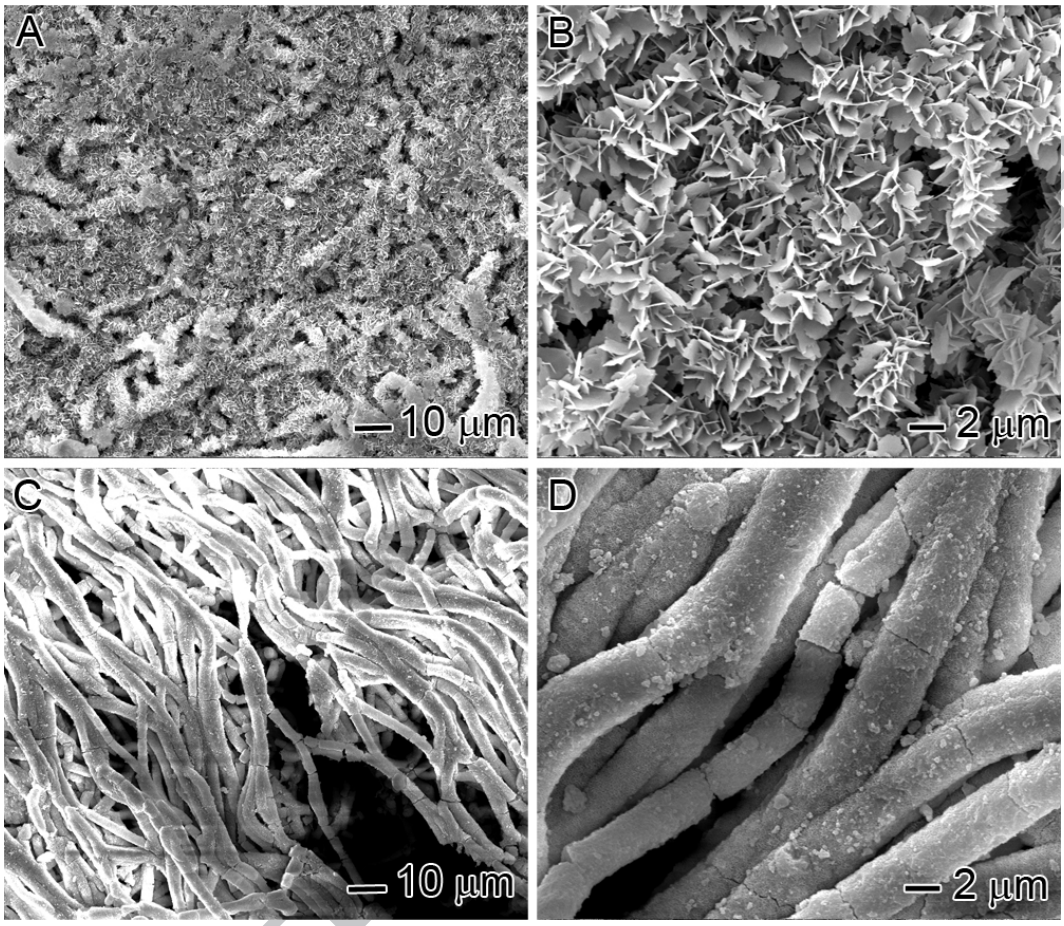


Fig. 3. SEM images mineralized PCL fibers which were produced using method I shown in Fig.1. Plasma-treated, electrospun PCL fibers were coated with polydopamine for 4 h in 0.2 mg/mL dopamine solution at pH 8.5 and mineralized in the 10×SBF solution containing 0.01 M (A, B) and 0.04 M (C, D) $NaHCO_3$ at 37 °C for 72 h.

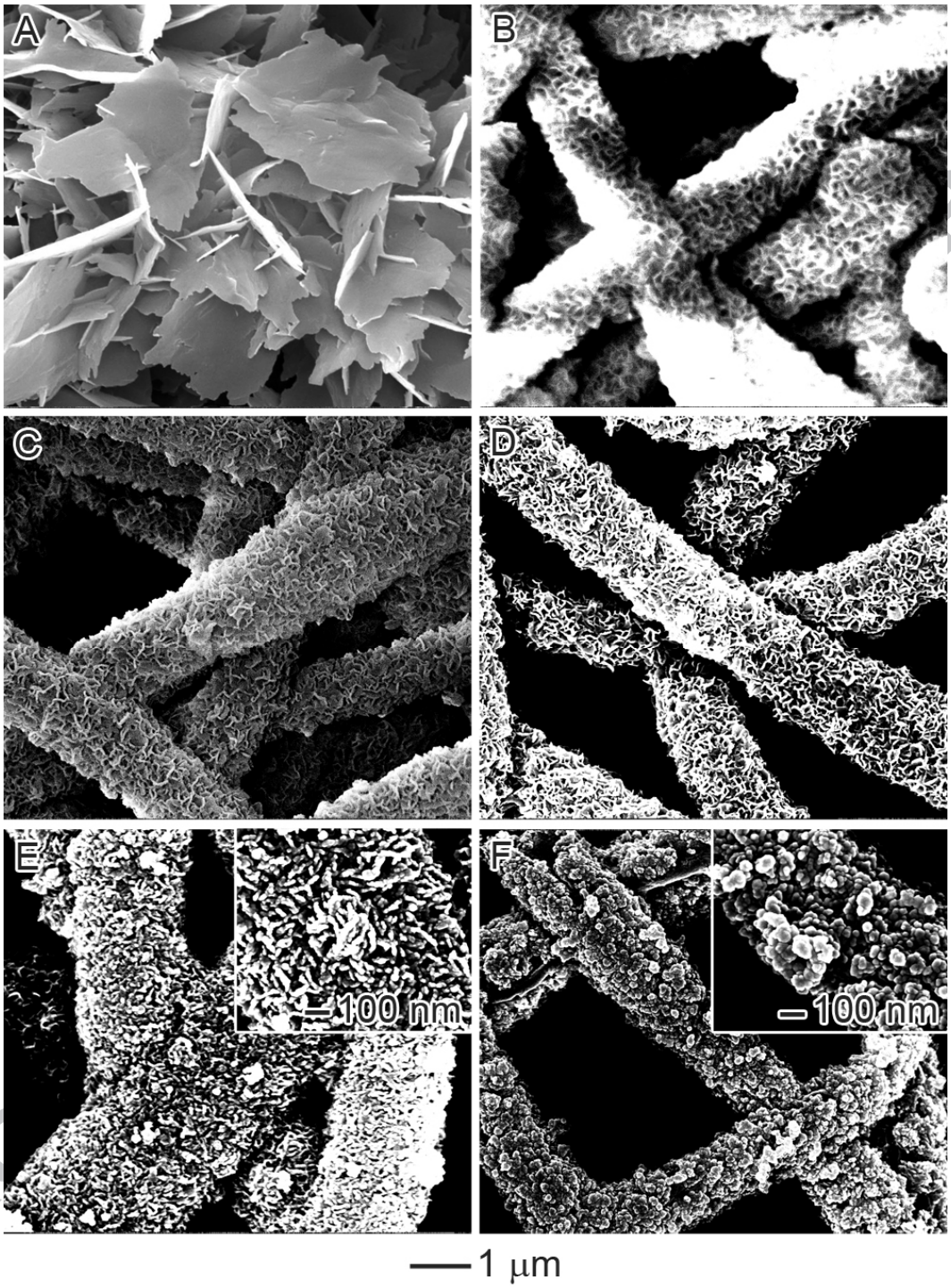


Fig. 4. SEM images of mineralized PCL fibers which were produced using method II shown in Fig.1. Plasma-treated, electrospun PCL fibers were mineralized in the 10×SBF solution containing 0.04 M $NaHCO_3$ and 0 mg/mL (A), 0.02 mg/mL (B), 0.1 mg/mL (C), 0.2 mg/mL (D), 0.4 mg/mL (E), and 1 mg/mL (F) dopamine at 37°C for 24 h.

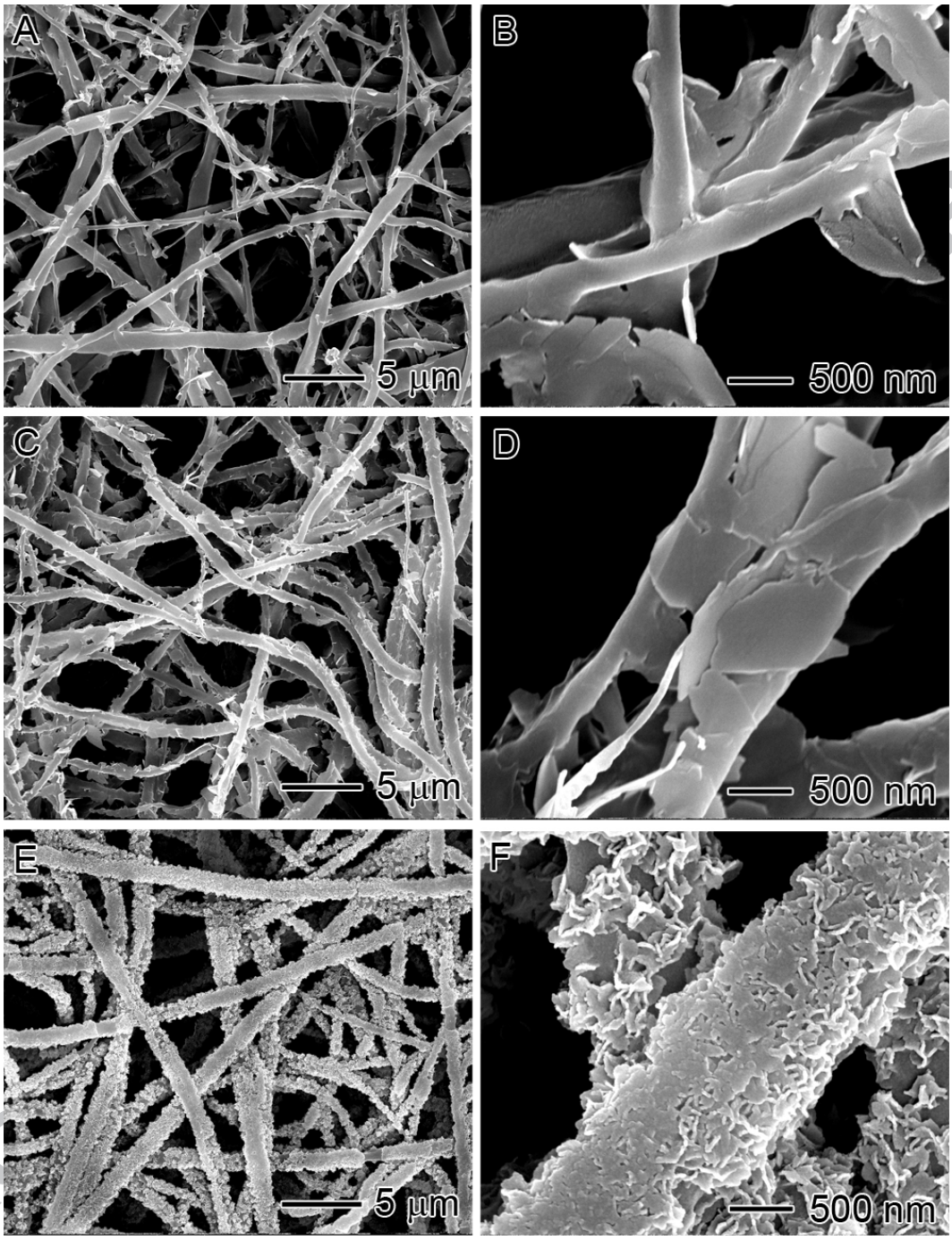


Fig. 5. SEM images of mineralized PCL fibers which were produced using method II shown in Fig.1. Plasma-treated, electrospun PCL fibers were mineralized in the 10×SBF solution containing 0.01 M $NaHCO_3$ and 0 mg/mL (A, B), 0.04 mg/mL (C, D), and 1 mg/mL (E, F) of dopamine at 37 °C for 2 h.

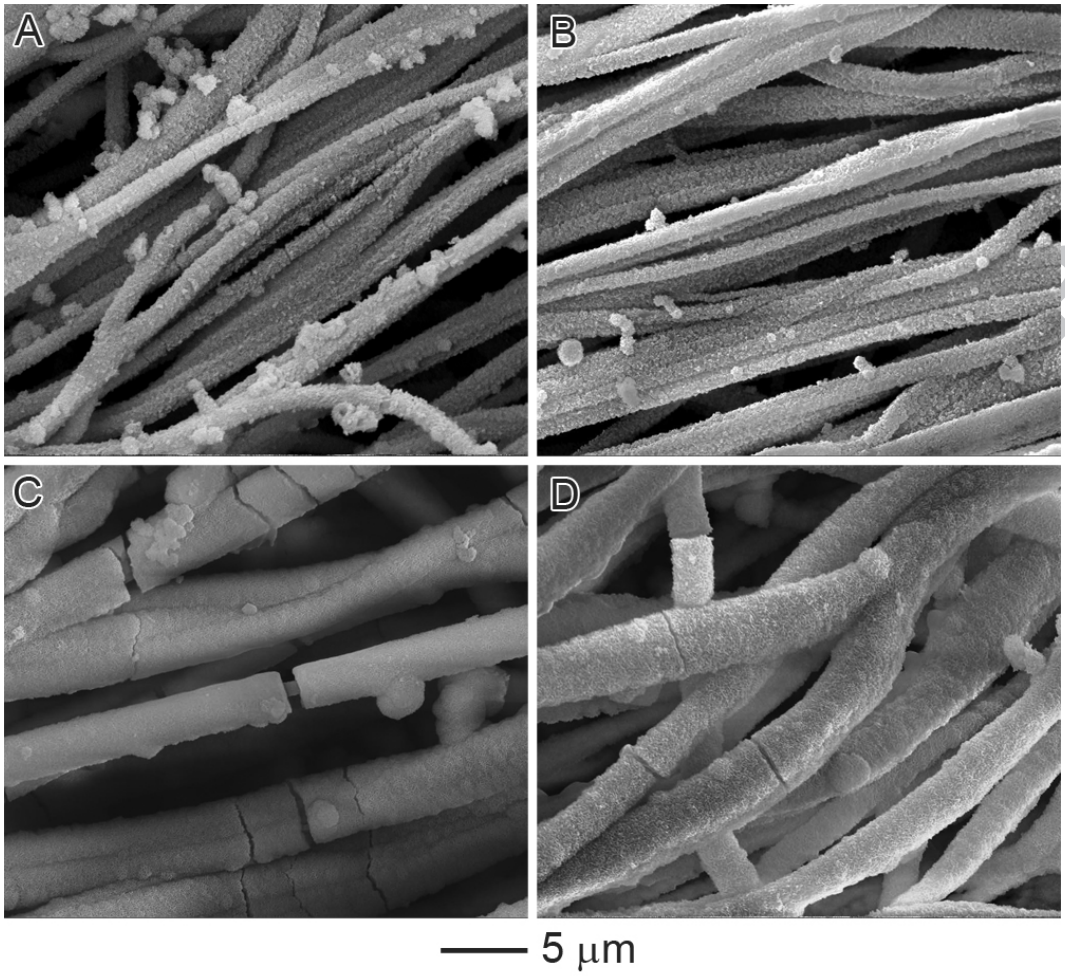


Fig. 6. SEM images of mineralized PCL fibers which were produced using method II shown in Fig.1. Plasma-treated, electrospun PCL fibers were mineralized in the 10×SBF solution containing 0.04 M $NaHCO_3$ and 0.2 mg/mL dopamine at 37 °C for different times: (A) 1h; (B) 24 h; (C) 48 h; and (D) 72 h. The reaction solution was changed every 24 h.

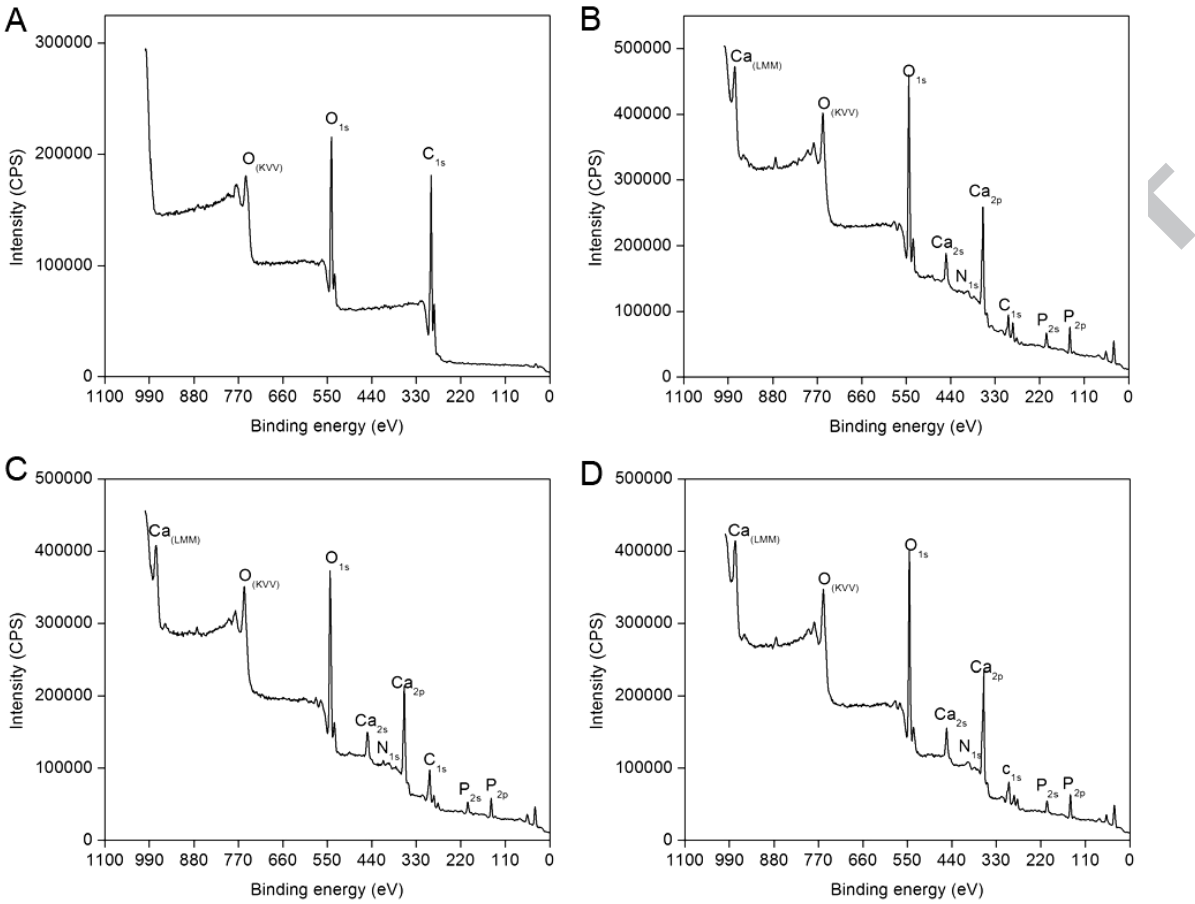


Figure 7**.** Survey scan XPS spectra for (A) plasma-treated PCL fibers, (B) HAP-PDA-PCL fibers (24 h), (C) HAP/PDA-PCL fibers (24 h), and (D) HAP/PDA-PCL fibers (48 h). HAP-PDA-PCL fibers (24 h): plasma-treated PCL fiber were coated with polydopamine in 0.2 mg/mL dopamine solution at pH = 8.5 for 4 h prior to mineralization in the 10×SBF solution containing 0.04 M $NaHCO_3$ for 24 h. HAP/PDA-PCL fibers (24 h) and HAP/PDA-PCL fibers (48 h): PCL fibers: plasma treated PCL fibers were mineralized in the 10×SBF solution containing 0.4 M $NaHCO_3$ and 0.2 mg/mL dopamine for 24 h and 48 h. The reaction solution was changed every 24 h.

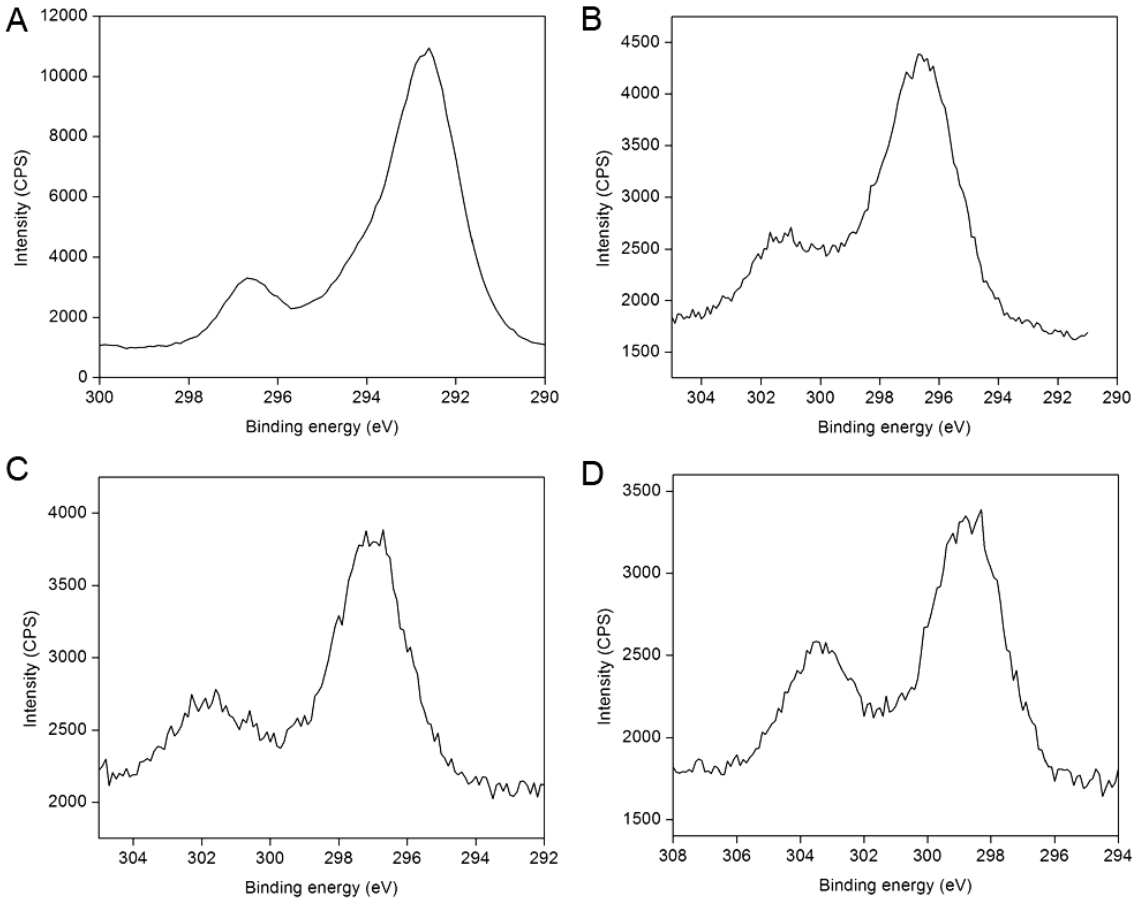


Figure 8. Typical XPS C_{1s} spectra for (A) plasma-treated PCL fibers, (B) HAP-PDA-PCL fibers (24 h), (C) HAP/PDA-PCL fibers (24 h), and (D) HAP/PDA-PCL fibers (48 h). HAP-PDA-PCL fibers (24 h): plasma-treated, electrospun PCL fiber were coated with polydopamine in 0.2 mg/mL dopamine solution at pH = 8.5 for 4 h prior to mineralization in 10×SBF solution containing 0.04 M $NaHCO_3$ for 24 h. HAP/PDA-PCL fibers (24 h) and HAP/PDA-PCL fibers (48 h): PCL fibers: plasma treated, electrospun PCL fibers were mineralized in the 10×SBF solution containing 0.04 M $NaHCO_3$ and 0.2 mg/mL dopamine for 24 h and 48 h. The reaction solution was changed every 24 h.

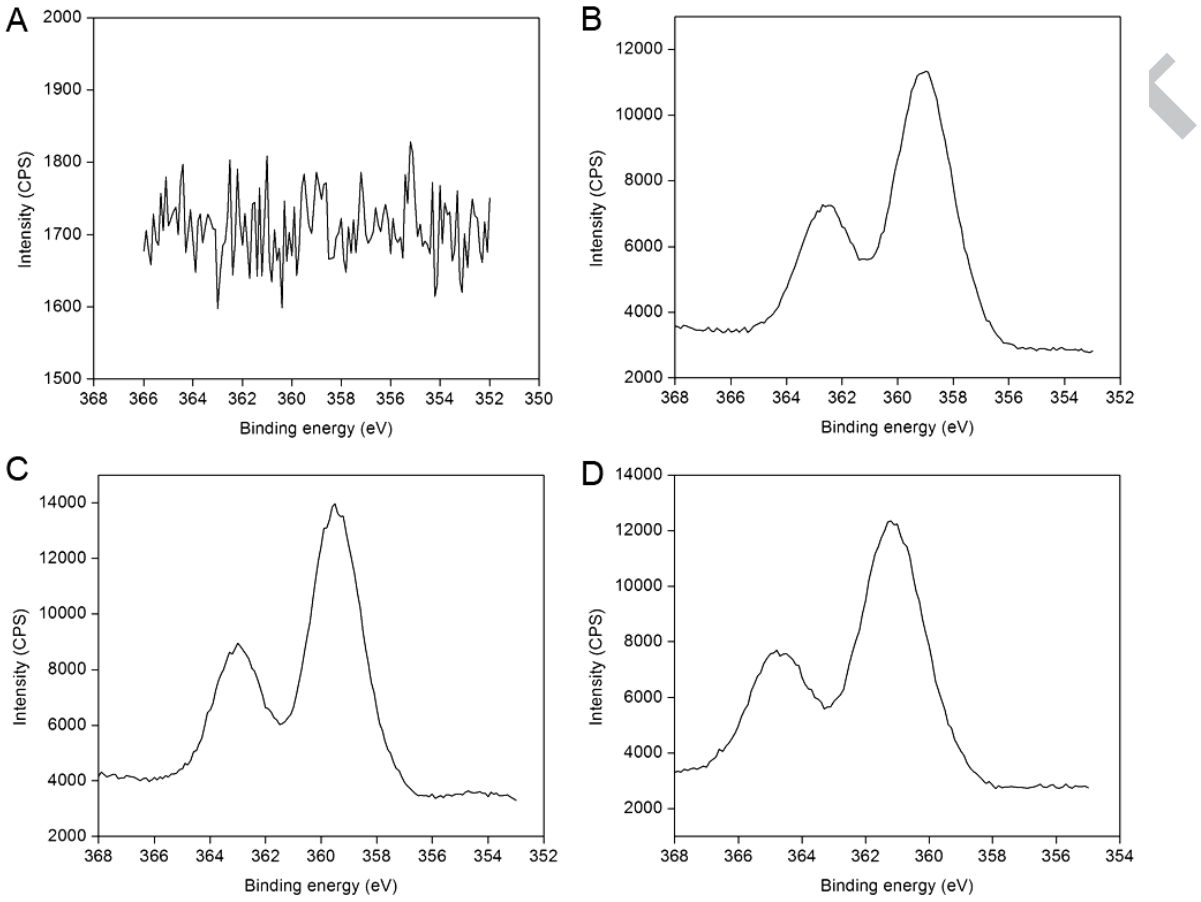


Figure 9. Typical XPS Ca_{2p} spectra for (A) plasma-treated PCL fibers, (B) HAP-PDA-PCL fibers (24 h), (C) HAP/PDA-PCL fibers (24 h), and (D) HAP/PDA-PCL fibers (48 h). HAP-PDA-PCL fibers (24 h): plasma-treated, electrospun PCL fiber were coated with polydopamine in 0.2 mg/mL dopamine solution at pH = 8.5 for 4 h prior to mineralization in the 10×SBF solution containing 0.04 M $NaHCO_3$. HAP/PDA-PCL fibers (24 h) and HAP/PDA-PCL fibers (48 h): PCL fibers: plasma treated, electrospun PCL fibers were mineralized for 24 h and 48 h in the 10×SBF solution containing 0.04 M $NaHCO_3$ and 0.2 mg/mL dopamine. The reaction solution was changed every 24 h.

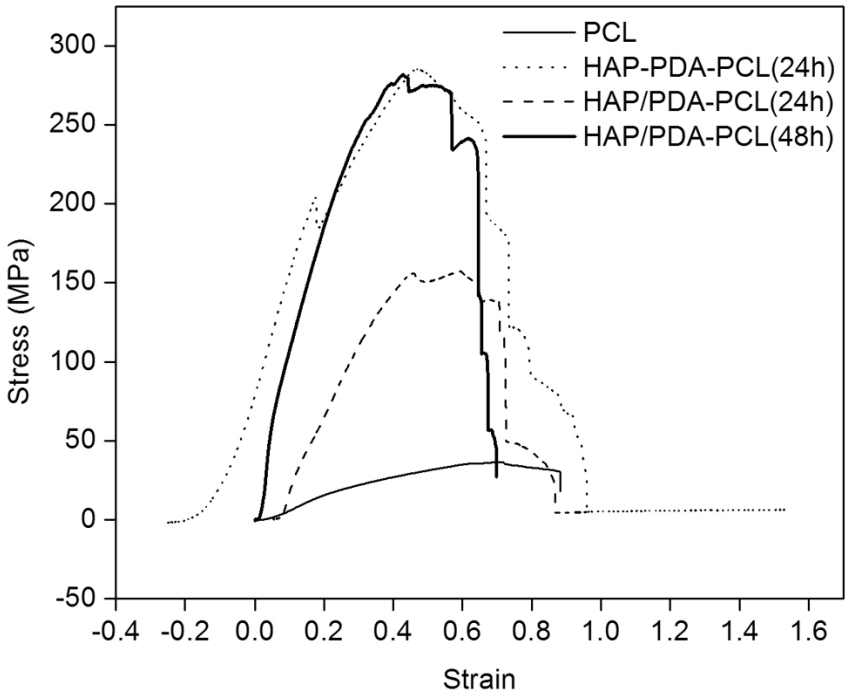


Figure 10. Stress-strain curves for various fibrous samples. HAP-PDA-PCL (24 h): plasma-treated, electrospun PCL fibers were coated with polydopamine in 0.2 mg/mL dopamine solution at pH = 8.5 for 4 h prior to mineralization in the 10×SBF solution containing 0.04 M $NaHCO_3$. HAP/PDA-PCL (24 h) and HAP/PDA-PCL (48 h): plasma treated, electrospun PCL fibers were mineralized in the 10×SBF solution containing 0.04 M $NaHCO_3$ and 0.2 mg/mL dopamine for 24 h and 48 h. The reaction solution was changed every 24 h.

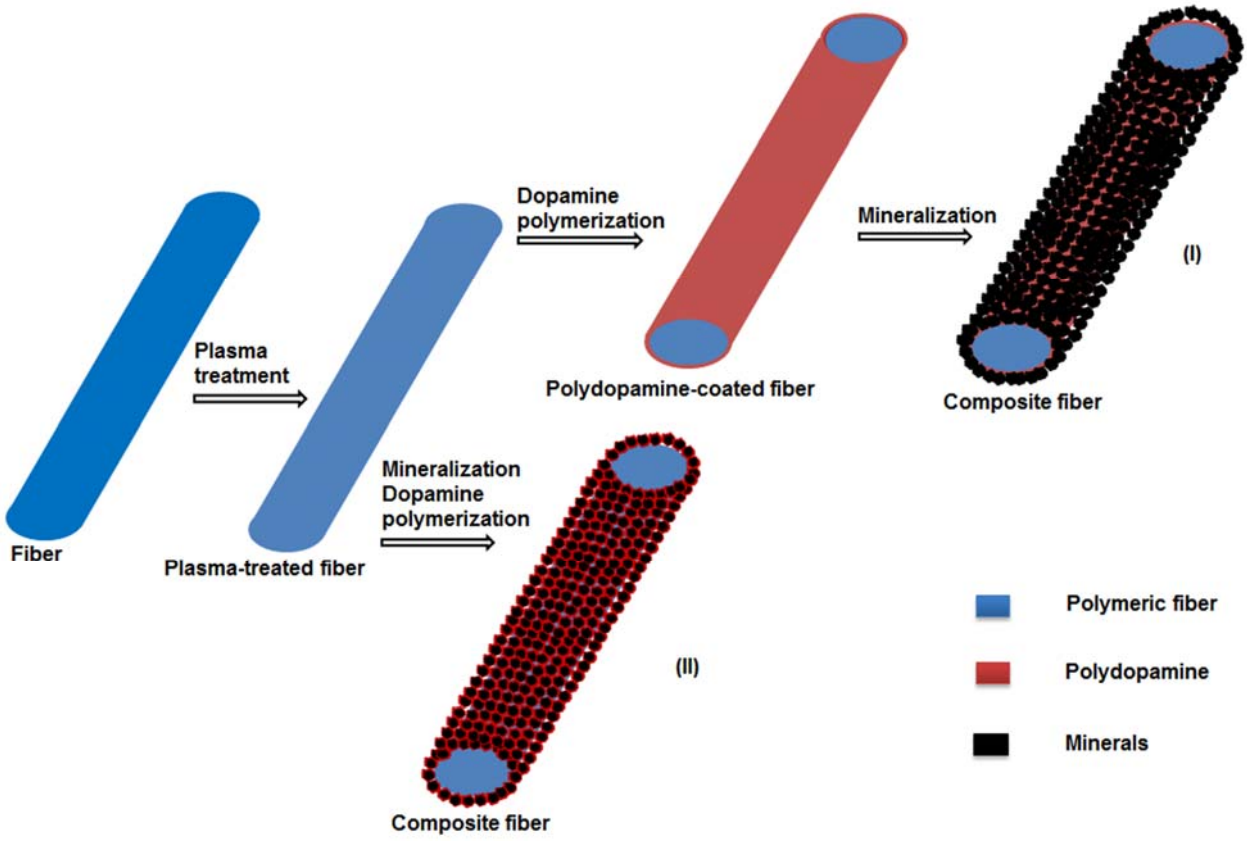
Table 1. Composition of mineralization solution

Composition of 10×SBF solution	Concentration (g/L)
NaCl	58.43
$CaCl_2$	2.77
$NaH_2PO_4 \cdot H_2O$	1.39

Table 2. Surface elemental analysis of various fiber samples

Atomic %	PCL	HAP-PDA-PCL(24h)	HAP/PDA-PCL(24h)	HAP/PDA-PCL(48h)
C	73.1	29.73	15.7	16.1
O	26.9	45.35	52.1	51.5
Ca		15.51	19.0	18.4
P		8.82	12.9	11.3
N		0.59	0.3	2.7

PCL fibers: plasma-treated PCL fibers. HAP-PDA-PCL (24 h): plasma-treated PCL fiber were coated with polydopamine in 0.2 mg/mL dopamine solution at pH = 8.5 for 4 h prior to mineralization in 10×SBF solution containing 0.04 M $NaHCO_3$. HAP/PDA-PCL (24 h) and HAP/PDA-PCL (48 h): plasma-treated PCL fibers were mineralized for 24 h and 48 h in 10×SBF solution containing 0.04 M $NaHCO_3$ and 0.2 mg/mL dopamine. The reaction solution was changed every 24 h.



Graphical abstract