

Fabrication of Bioactive Carbon Nonwovens for Bone Tissue Regeneration

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Abstract

The aim of tissue engineering is to repair or replace the function of defective or damaged tissue. One of the key factors is the creation of a scaffold as an artificial extracellular matrix (ECM) for cellular attachment, proliferation and differentiation. In scaffold-based bone tissue engineering, both the porosity and mechanical properties of the scaffold are of great importance. To mimic the structure of natural ECM, three fibrous scaffolds based on composite carbon fibres containing nanohydroxyapatite were fabricated using nonwoven techniques. The overall objective of the present work was to compare and analyse the properties of needlepunched nonwoven produced from PAN and PAN/HAp fibres before and after stabilization and carbonation processes. The characterisation of the scaffold showed that after the carbonisation process, the nonwoven had an interconnective microporous structure (70-150 µm), high porosity as well as mechanical and structural integrity. Soaking the nonwoven in simulated body fluid (SBF) at body temperature formed a bone-like apatite on the fibre surface. The formation of the bone-like apatite demonstrates the potential of nonwovens for integration with bone.

Key words: carbon nonwoven, bioactive scaffolds, polyacrylonitrile fibres, hydroxyapatite.

tain features of natural tissue in scaffold design. Needle-punched nonwoven fabrics provide a large surface area whose fibrous structure tends to mimic the architecture of natural tissue. Tissue engineering scaffolds can be either permanent or temporary, depending on the application and function of the neo-tissue.

Bone is a nanocomposite that consists of a protein-based soft hydrogel template (i.e., collagen, non-collagenous proteins and water) and hard inorganic components (HAp-hydroxyapatite $\text{Ca}_{10}(\text{PO}_4)_6(\text{OH})_2$). Specifically, 70% of the bone matrix is composed of nanocrystalline Hap, which is typically 20 - 80 nm long and 2 - 5 nm thick [2].

In the case of severe defects and loss of volume, bone would not heal by itself, and grafting is required to restore function to it without damaging living tissue. Graft materials not only replace missing bone but also help the body to regenerate its own lost bone [3]. Over the past four decades, several biomaterials have been developed and successfully used as bone grafts. Hydroxyapatite (HAp) is a class of calcium phosphate-based bioceramic which is frequently used as a bone graft substitute. HAp is not only bioactive but also osteoconductive, non-toxic, non-immunogenic, and its structure is crystallographically similar to that of bone [4]. It has also been proven that nano-HAp, compared to conventional micro-HAp, promotes osteoblast adhesion, differentiation and proliferation, as well as osteointegration and the deposition of calcium-containing minerals on its surface, which leads to the enhanced formation of

new bone tissue within a short period [5]. Unfortunately HAp behaves as a typical brittle material, therefore it is used only in low weight-bearing orthopaedic applications e.g. as a small bone defect filler, a coating-agent for metallic implants, and drug delivery.

Among the possible forms of implants, the fibrous matrix is highly promising in medicine for bone tissue regeneration by acting as a cell supporting scaffold. Carbon fibres and its composites are one of the most widespread groups of biomaterials used in orthopaedics. They have been used in the reconstruction of fibrous tissue such as ligaments and tendons, as well as for the regeneration of bone and cartilage defects [6]. Carbon material is so attractive for medical applications that individual researchers have implemented it in their own experiments without further physical and chemical characterisation, as well as without specifying the sterilisation methods. Therefore, on the one hand, we can find information about a high biocompatibility carbon fibres, on the other hand, even about toxic effects of fibre degradation products on the human body [7-10]. It was demonstrated by later studies that the cellular response to fibrous carbon material depends on the degree of crystallinity of the material, therefore only selected types of carbon fibres are suitable for the treatment of tissue. Highly crystalline, high modulus fibres are not suitable for medical purposes, while amorphous fibres are excellent for implants [6]. However, even low-crystalline CF, when implanted into bone defects, are encapsulated by connective tissue, being the reason for the long du-

■ Introduction

The success of tissue engineering greatly relies on the development of a suitable scaffold system for both *in vitro* tissue culture and subsequent *in vivo* neo-tissue formation. Ideal bone tissue engineered scaffolds should provide a three-dimensional (3D) matrix with a mechanical strength adequate to support the newly formed tissue, a high enough porosity to allow new tissue formation and growth within the scaffolds, a biomimetic structure for nutrient transport, as well as waste removal and good biocompatibility [1]. It is often beneficial to mimic cer-

ration of the process of bone restoration, which limits the applicability of such implants in the treatment of bone tissue defects. Carbon fibres can be subjected to various chemical and structural modifications. For example, by inserting HAp into the structure of polyacrylonitrile precursor fibres, it is possible to obtain a new generation of bioactive carbon fibres [11 - 13]. Our earlier research (made in cooperation with the Department of Man-made Fibres, Technical University of Łódź) regarding the modification of carbon fibres with hydroxyapatite indicated that the addition of hydroxyapatite to precursor fibres enhanced the proliferation of bone tissue cells, and the *in vivo* study established the biocompatibility of HAp containing carbon fibres [14, 15].

The current work was conducted as an extension of previous studies [13, 15]. Our goal was to mimic the composition of natural bone, therefore our research was focused on the application of HAp-containing PAN fibres to produce three-dimensional, porous, bioactive, carbon nonwovens which could serve as scaffolds for the treatment of bone tissue defects. Bioactive material in a biological environment should be covered with apatite similar to the natural one present in bones, which allows for the bonding of the biomaterial with bone. It is believed that the appearance of apatite on the implant surface (a process known as bioactivity) leads to the formation of chemical bonds on the implant - bone interface. In scaffold-based bone tissue engineering, both the porosity and mechanical properties of the scaffold are of great importance. In the design of scaffolds, high porosity usually leads to a decrease in biomechanical strength. The initial requirement of the scaffold is to hold cells and tissue together in spite of partial degradation, which reflects the importance of mechanical strength at the initial stages. Compared to the strengths of metals and ceramics for medical applications, the strengths of fibrous scaffolds are already very low. The mechanical and structural integrity of a scaffold are crucial to withstand *in vitro* culture conditions and surgical manipulations. Therefore it seems necessary to evaluate the mechanical properties of the scaffold designed. In order to solve the conflict between optimising the porosity and mechanical properties, we obtained three different types of carbon nonwoven scaffolds made of polyacrylonitrile-hydroxyapatite composite fibres. The overall objective of the

present work was to compare and analyse the properties of needlepunched nonwoven produced from PAN and PAN/HAp fibres before and after stabilization and carbonisation processes. Data from the experiment will help design bioactive nonwovens which would meet the requirements for bone scaffold materials.

Materials and methods

For this investigation, polyacrylonitrile fibres (PAN) containing nano-hydroxyapatite (HAp), prepared at the Technical University of Łódź, were selected. The hydroxyapatite used in the study was provided by the University of Science and Technology (Cracow, Poland) [16]. The specific surface area of the HAp powder was 71.4 m²/g. Fibres were obtained using PAN-terpolymer from Zoltek (Hungary), which had the following composition: 93 - 94% of weight meres of acrylonitrile, 5 - 6% of weight meres of methyl acrylate and about 1% of weight meres of alilosulphoniane. Ceramic powders (3% w/w) were added to the PAN spinning solution, in which Dimethylformamide (DMF) was used as a solvent. The modified PAN fibres were spun as described in [11, 17, 18]. The properties of precursor fibres and morphological changes occurring during the processes of thermal stabilization and carbonisation have been described in previous works [11 - 13]. A fibrous web (of a basic weight of approx. 120 g/m²) was prepared from PAN cut fibres (modified with HAp and not modified) by mechanical processing using a laboratory carding machine - 3KA of the Befama company. The fibres were bonded by needle punching using the same amount of needling and needling depth. As a result, two types of nonwoven fabrics were obtained:

- (1) PAN/HAp nonwoven - made of HAp-containing PAN fibres, and
- (2) PAN nonwoven - made of unmodified PAN fibres.

Afterwards these 'starting materials' - in order to obtain nonwovens with three dimensional structures - were bedded and then subjected again to mechanical needling with a different needle punching density (lp/cm²). Three types of nonwovens were obtained:

- (3) PAN/HAp with a needle punching density of 90 lp/cm²,
- (4) PAN/HAp with a needle punching density of 180 lp/cm², and
- (5) PAN with a needle punching density of 180 lp/cm².

A multi-stage process of heat treatment was carried out in an oxidising atmosphere at temperatures from 150 °C to 280 °C; these temperatures of stabilization were chosen based on previous experiments [13]. Subsequent carbonisation was performed for 15 min. at 1000 °C (heating rate: 5 °C/min, inert atmosphere: argon), and as a result five types of carbon nonwovens were obtained. Three types of the nonwovens obtained were made from HAp-containing carbon fibres:

- (1) CF/HAp 'one layer',
 - (3) CF/HAp (90 lp/cm²),
 - (4) CF/HAp (180 lp/cm²), and,
- two nonwovens made from unmodified carbon fibres were used as control materials:
- (2) unmodified CF 'one layer', and
 - (5) CF (180 lp/cm²).

Particle size distribution of the HAp powder used in this work was made by the dynamic light scattering method (DLS) using a Zetasizer Nano ZS, Malvern Instruments Ltd. A microscopic study of the 3D structures was made using scanning electron microscopy (Jeol, JSM-5500). Before the observation, the samples were coated with gold using a sputter coater. The diameter of the fibres were measured using a Lanameter MP-3 microscope. The average diameter of the fibres was determined by measuring the diameter at 100 different points.

The following parameters of the precursor and carbon nonwovens were measured: surface weight (PN-EN 29073-1:1994), thickness (PN-EN 29073-2:1994), air permeability (PN-EN ISO 9237:1998), pore size distribution, and apparent density. Measurement of the fabric thickness was carried out on a Thickness Tester (TILMET 73). A pressure of 2 kPa was applied for all the thickness measurements. Air permeability tests were carried out using a FX 3300 Labotester III from Textest AG, at the following conditions: sample surface - 1.76 cm², and pressure drop - 100 Pa. A PMI capillary flow porometer was applied to evaluate the pore volume/size distribution of nonwoven fabrics [19]. PAN nonwovens and nonwovens composed of fibres modified with hydroxyapatite were subjected to DSC analysis (5100 TA Instruments) to determine the influence of temperature on the stabilization process. The analysis was performed at the following conditions: heating rate - 10 °C/min, and nitrogen gas flow - 40 ml/min. The tensile

Table 1. Ion concentration of SBF and human blood plasma.

Kind of ion	Ion concentration, mM	
	SBF	Blood plasma
Na ⁺	142.0	142.0
K ⁺	5.0	5.0
Mg ²⁺	1.5	1.5
Ca ²⁺	2.5	2.5
Cl ⁻	148.8	103.0
HCO ₃ ⁻	4.2	27.0
HPO ₄ ²⁻	1.0	1.0

strength of the strips of the precursor, stabilised and carbonized nonwovens was determined on a Zwick-Roell Z 2.5. universal material testing machine at a constant speed of 2 mm/min. The specimens were tested under tension until failure. Average values of the nonwoven tensile strength were calculated for each specimen.

Measurements of the propagation velocity of a longitudinal ultrasonic wave were made using an MT-541 (UNIPAN) at the frequency $f = 100$ kHz. On this apparatus one head emits an ultrasonic wave, while the other one is a receiver, collecting the wave after transmission through the materials. Samples in the form of a strip (11.5×75 mm) were rolled. An emitter and receiver were placed on both sides of the rolled samples. Mean velocities were obtained by averaging three independent series of measurements. The ultrasonic method enables precise determination of the velocity (v) to produce the same estimation of the specific modulus (E/g), thus the velocity of the ultrasound wave was established as the main determinant of changes in the nonwoven elastic properties.

Bioactivity is a very important feature of nonwovens when using them for bone scaffold materials. The bioactivity of the nonwovens obtained was evaluated by soaking them in a simulated body fluid (SBF). The SBF solution was prepared according to the procedure described by Kokubo [20]. **Table 1** shows the ion concentration of the SBF solution and human blood plasma. Bioactivity tests were performed using $1.5 \times$ Simulated Body Fluid (SBF) of pH 7.4, at a temperature of 37 °C. SBF has the ability to form apatitic calcium phosphates on immersed osteoinductive materials within a few days to 2 weeks. Carbon nonwoven modified with HAp (sample number 4) and reference standard unmodified nonwovens (sample number 5) were chosen for the bioactivity test. Four nonwovens from each group were incubated in $1.5 \times$ SBF fluid in closed polyethylene containers. The SBF solution was replaced every 2.5 days. After 1, 3, 7 & 14 days of soaking, the samples were removed from the SBF, gently washed with deionised water, and dried at room temperature. The surface morphology before and after incubation in artificial plasma (SBF) was observed using scanning electron microscopy (SEM, Jeol JSM 5500).

Results and discussion

Figure 1 shows the particle size distribution of the hydroxyapatite powder used in the study as a bioactive filler. The DLS results revealed that the HAp powder consisted of nanosized crystallites (< 100 nm) as well as their agglomerates.

The microstructures of the original PAN fibres and HAp modified PAN fibres are presented in **Figure 2.a** and **2.b**, respec-

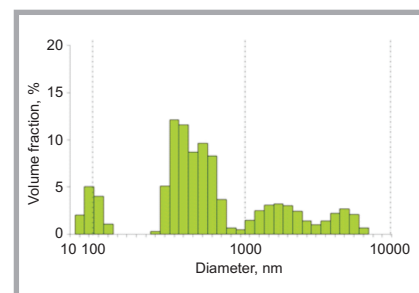


Figure 1. Particle size distribution of hydroxyapatite powder.

tively. The SEM micrographs illustrate typical surfaces of pure PAN fibres (**Figure 2.a, 2.c**). Hydroxyapatite particles and their agglomerates are clearly visible on the surface of the PAN/HAp fibres. The fibres were successfully converted into 3D nonwoven structures. Macroscopic observation of PAN/HAp nonwovens showed a ‘fluffy’ character for sample no. 3, whereas sample no. 4 was very rigid. The morphology of needle-punched precursor nonwovens is shown in **Figure 2.c** and **2.d**. Fibres of different diameter and relatively uniform fibre spatial arrangement were observed. The fibre diameter distribution of PAN and PAN/HAp precursor nonwovens is presented in **Figure 3.a**. The diameter of fibres varied from 8 – 14 μm for unmodified PAN nonwovens and from 8 – 22 μm for modified PAN/HAp nonwovens. In order to analyse the effect of a temperature increase on nonwoven oxidation, the materials were analyzed by the DSC method (**Figure 4**). Each of the samples analysed: (3) PAN/HAp 90 lp/cm², (4) PAN/HAp 180 lp/cm², and (5) PAN 180 lp/cm² were thermally stable up to 290 °C, and exothermic peaks were observed at 293 °C, 310 °C and 313 °C for PAN 180 lp/cm² (5), PAN/HAp 90 lp/cm² (3) and PAN/HAp 180 lp/cm² (4), respectively. **Tables 2** and **3** show a comparison between the properties of the precursor and those of carbon nonwovens produced from PAN and PAN/HAp fibres. It was observed that the processes of thermal stabilization and carbonisation cause a reduction in nonwoven thickness, whereas the density and surface weight of carbon nonwovens is always higher than those of corresponding precursor samples (**Table 2**). Air permeability is the capacity of nonwoven to let air pass through. At a constant pressure the amount of airflow through nonwoven depends on the porosity, thickness and mass. With an increase in nonwoven density (during the carboni-

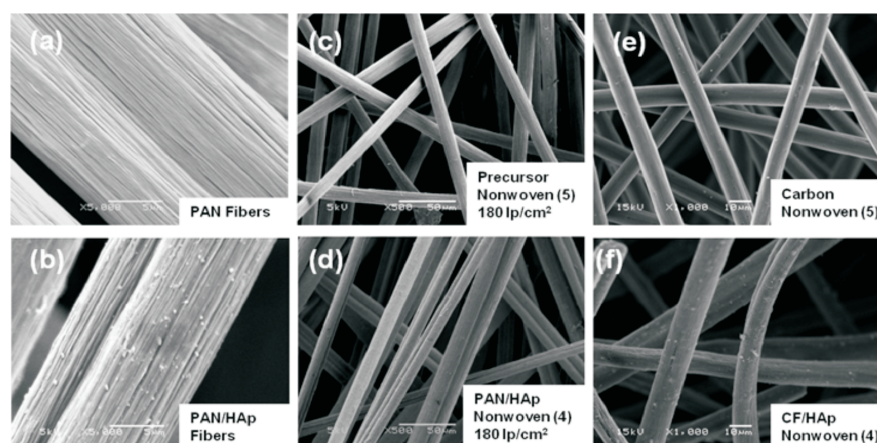


Figure 2. SEM photograph of the precursor fibres (a - b), precursor (c - d) and carbon nonwovens (e - f).

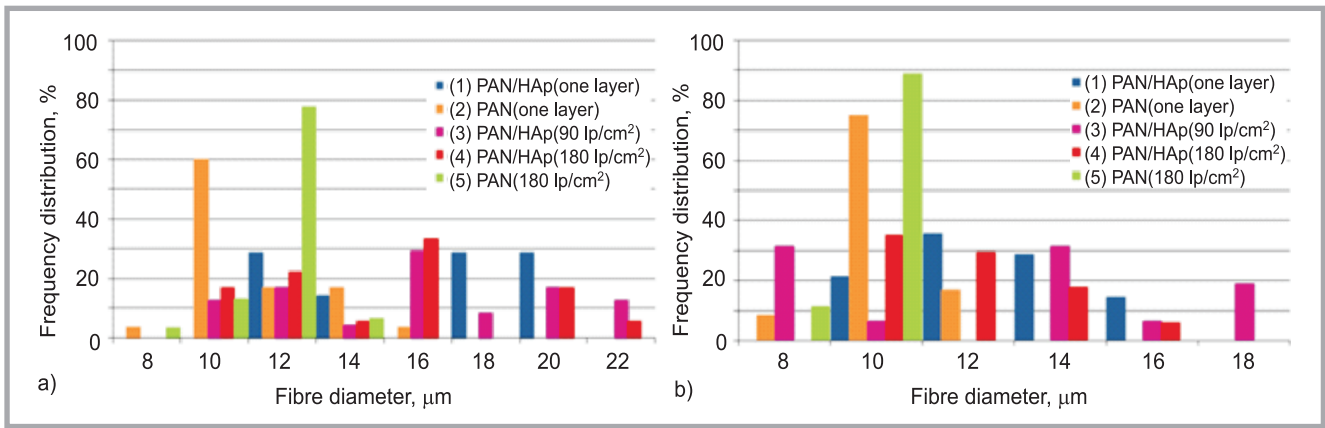


Figure 3. Diameter distribution of fibres for (a) precursor nonwovens and (b) carbon nonwovens.

Table 2. Characteristics of the precursor and carbon nonwovens.

Designation	Fibres	No. of layers	Punching density, lp/cm ²	Surface weight, g/cm ²				Thickness, mm				Apparent density, g/cm ³	
				Precursor nonwoven		Carbon nonwoven		Precursor nonwoven		Carbon nonwoven		Precursor nonwoven	Carbon nonwoven
				Volume	CV, %	Volume	CV, %	Volume	CV, %	Volume	CV, %		
(1)	PAN/HAp	1	-	143.4	0.9	153.3	2.5	2.20	9.66	1.67	1.14	0.065	0.092
(2)	PAN	1	-	87.4	5.9	80.3	5.4	2.59	9.36	1.55	4.50	0.034	0.062
(3)	PAN/HAp	2	90	115.6	0.9	210.0	7.2	3.70	3.39	1.93	3.4G	0.031	0.109
(4)	PAN/HAp	2	180	183.0	6.7	272.5	6.9	3.78	6.28	2.30	2.40	0.048	0.118
(5)	PAN	2	180	215.2	2.9	268.5	1.6	4.89	8.35	2.89	2.50	0.044	0.092

sation process), the air permeability decreases (Table 3).

Another important property of scaffolds is the interconnectivity of their pore network. Porosity results for the precursor and carbon nonwovens obtained are shown in Figure 5. The precursor samples exhibit a bimodal distribution of the pore size diameter, with a significant fraction of the porosity in the range of 200 – 300 µm for ‘one layer’ nonwovens (Figure 5.a) and 100 – 200 µm for ‘two layers’ nonwovens (Figure 5.b). Moreover, there is a very narrow distribution of pore sizes centred below 100 µm for samples 2, 3 & 5 and above 100 µm for samples no. 1 & 4. The main pore size fraction after the carbonisation process for “one layer” nonwovens was within the range of 100 – 130 µm for modified carbon nonwoven and 145 – 175 µm for unmodified carbon nonwoven (Figure 5.c). In the case of the “two layer” carbon samples, the main pore size fraction was within the range of 60 – 100 µm (Figure 5.d). Measurements of the porosity directly confirmed that needling as well as the carbonisation process influence the porosity of nonwovens, i.e. decrease the sizes of the main pore fraction and improve the structure homogeneity of carbon nonwovens. All pores in carbon

nonwovens are interconnected, which is very important for future cell growth in nonwoven as well as for migration and nutrient flow. For tissue engineering ap-

plications, the macropore diameter of scaffolds does not seem to be of great importance, but the interconnected pore diameter should be about 100 µm [21].

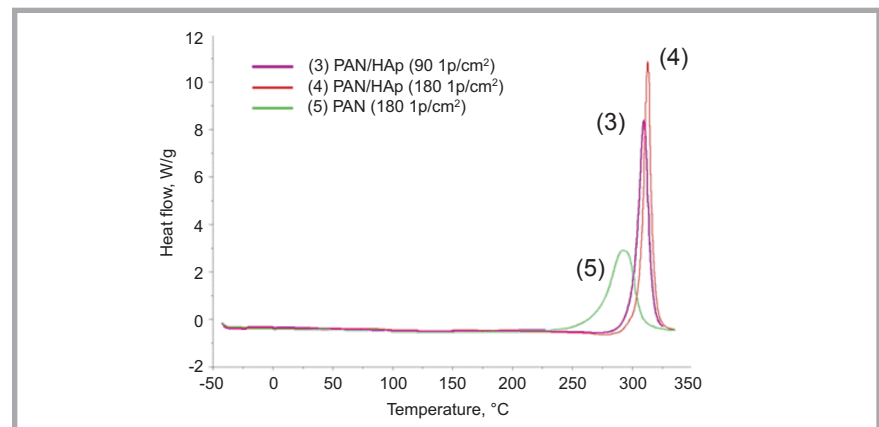


Figure 4. DSC curves of the unmodified and modified nonwovens.

Table 3. Comparison of air permeability characteristics of the nonwovens before and after the carbonisation process.

Designation	Fibres	No. of layers	Punching density, lp/cm ²	Air permeability, dm ³ /(m ² ·s)	
				Precursor nonwoven	Carbon nonwoven
(1)	PAN/HAp	1	-	179.00	60.40
(2)	PAN	1	-	142.00	86.23
(3)	PAN/HAp	2	90	64.13	38.03
(4)	PAN/HAp	2	180	114.36	71.16
(5)	PAN	2	180	116.42	103.95

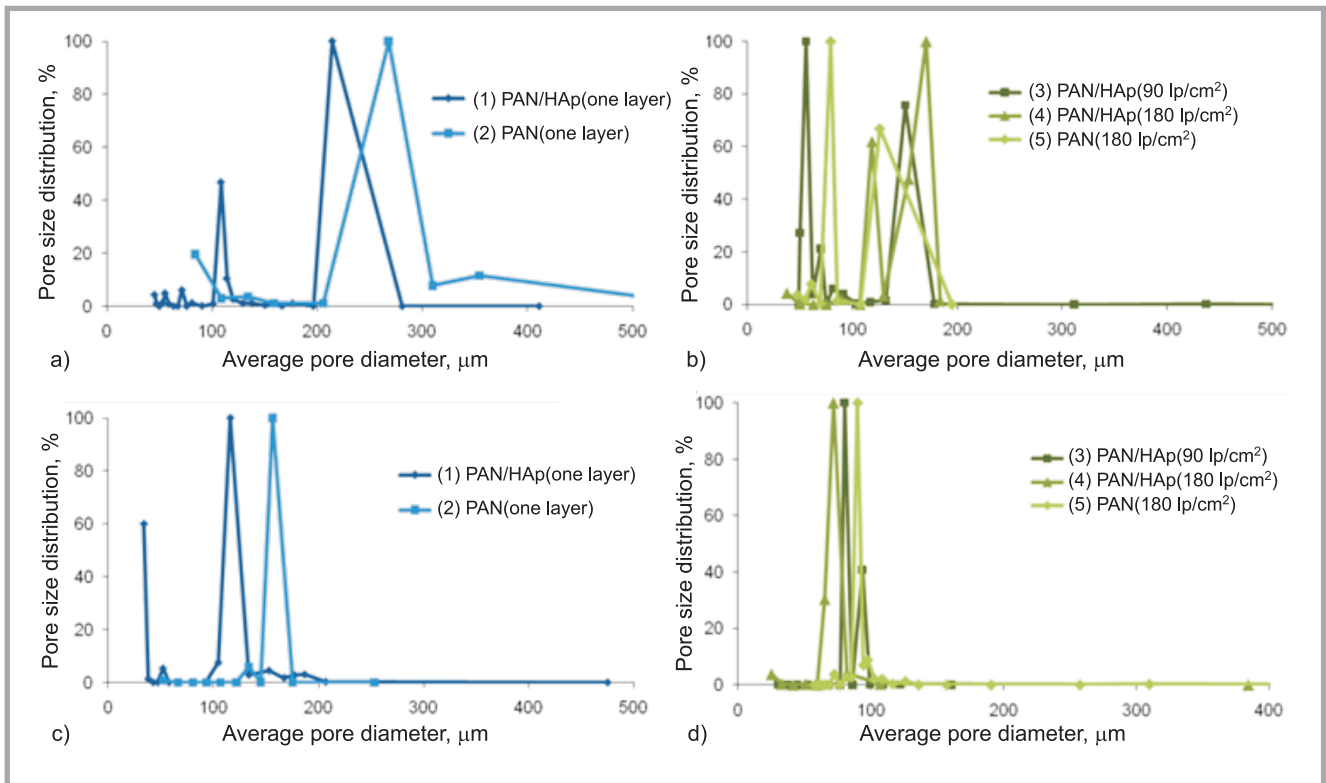


Figure 5. Pore size distribution for the modified and unmodified precursor (a-b) and carbon nonwoven fabrics (c-d).

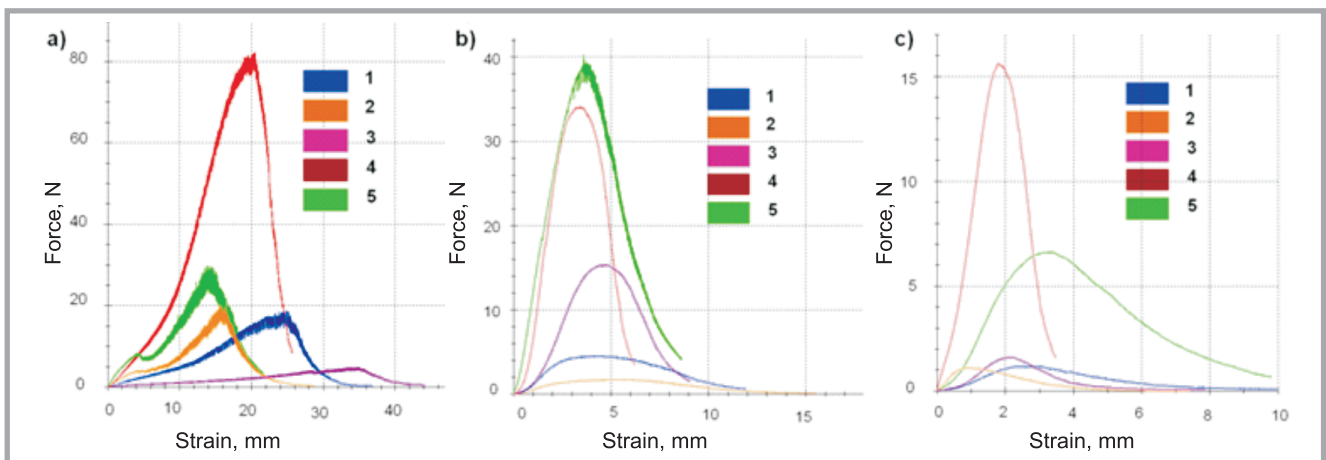


Figure 6. Force – strain curves for the (a) precursor nonwovens, (b) nonwovens after the stabilisation process, and (c) nonwovens after the carbonisation process.

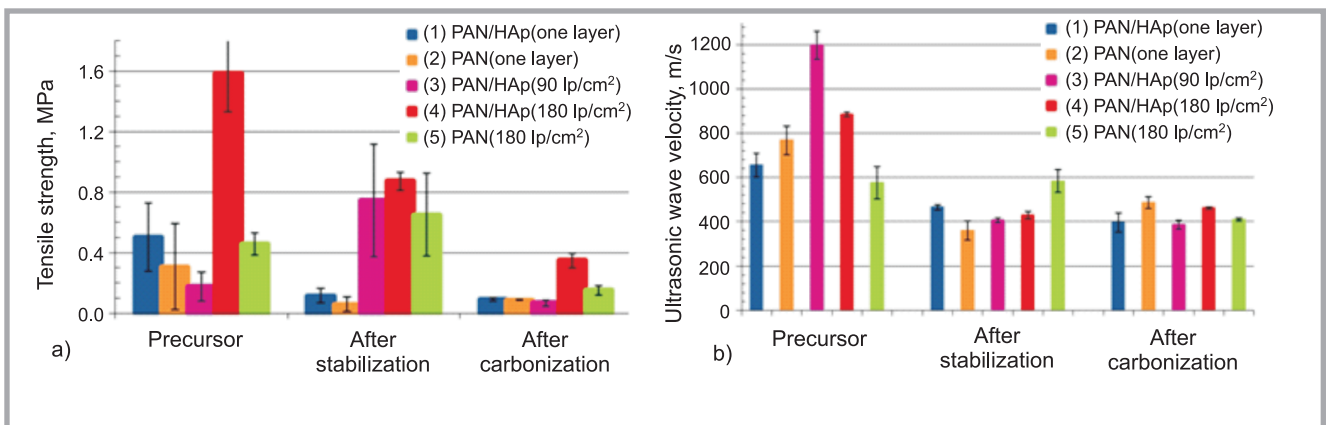


Figure 7. Mechanical characteristic of modified and unmodified nonwovens: (a) tensile strength and (b) ultrasonic wave velocity.

The microstructure of carbon nonwovens is presented in **Figure 2.e** and **2.f**. HAP particles were present on the fibre surface after the carbonisation process (**Figure 2.f**). The carbonisation process resulted in a decrease in fibre diameter for all types of carbon nonwovens (**Figure 3.b**). The thermal stabilization and carbonisation of polyacrylonitrile nonwovens were also accompanied by changes in mechanical properties. The results of the mechanical tests are displayed in **Figures 6 & 7**. Representative force - strain curves of the precursor nonwovens obtained are shown in **Figure 6.a**. The highest tensile force (80 N) was observed in the case of "two layer" PAN/HAp (180 lp/cm²) nonwoven (sample no. 4), whereas pure PAN nonwoven (with the same needle punching density - sample no. 5) was found to exhibit a 60% lower tensile force. The tensile force of "one layer" samples were comparable. The lowest force (5 N) was observed in the case of sample no. 3. The parameters of the tensile strength are presented in **Figure 7.a**. The mechanical properties of the precursor nonwovens obtained are strongly influenced by technological parameters. Better mechanical properties were observed in the case of the 'two layer' samples. Precursor PAN/HAp nonwoven (sample no. 4) was characterised by the highest tensile strength, almost four times higher than in the case of the control sample (no. 5), whereas the lowest tensile strength was observed in the case of sample no. 3. Both modified samples (no. 3 and 4) were subjected again to mechanical needling after bedding. It was shown that raising the needling number increased the mechanical strength of PAN/HAp nonwoven. The stabilization process resulted in a decrease in tensile strength for 'one layer' nonwovens (specimens no. 1 & 2). A moderate improvement in tensile strength was obtained for samples no. 3 and 5. After the carbonisation process, the tensile strength of all samples decreased; however, the highest tensile strength (0.4 MPa) was still observed for sample no. 4. The tensile strength of sample no. 4 is comparable to type I collagen - the major organic component of ECM in bone [22].

Velocities of the propagation of ultrasound longitudinal waves in the fibrous samples before the processes, as well as after thermal stabilization and carbonisation are presented in **Figure 7.b**. The highest values of ultrasound wave propagation were noticed in the case of

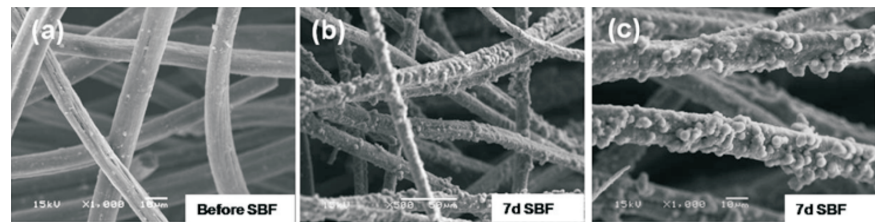


Figure 8. SEM microphotograph of CF/HAp nonwoven before (a) and after 7 days of immersion in SBF (b-c).

precursor sample no. 3. The higher ultrasonic wave velocity of these samples is a result of the manufacturing process i.e. a lower needling number resulted in the smaller damage of fibres. The ultrasonic wave velocity can be also connected to the rate of the stabilization process in the samples. From earlier studies it is known that the ultrasonic wave velocity decreases when the temperature and oxidation time increase; and after the final step the velocity starts to increase slightly (the carbonisation process begins). The data presented in **Figure 7.b** show that in the case of "one layer" samples (which have a lower thickness, see **Table 2**), stabilization is quicker due to easier air access, compared to the 'two layer' specimens. Sample 1 (modified with hydroxyapatite) is an exception - the carbonisation process had already begun. The value of the ultrasonic wave velocity is almost 20% higher for this sample compared to unmodified one layer sample no. 2. The precursor samples appeared to have the highest ultrasonic wave velocity. After stabilization the value of the velocity decreased, and no significant changes were observed after carbonisation.

Carbon nonwoven modified with HAp with the best mechanical properties (sample no. 4) and reference standard unmodified nonwovens (sample no. 5) were chosen for bioactivity testing. SEM images of a composite nonwoven before and after soaking in SBF are shown in **Figure 8**. The surface of CF/HAp nonwoven fibres before incubation in SBF was relatively smooth with small and sporadic prominences on the material surface (**Figure 8.a**). During incubation in SBF, the character of the nonwoven fibre surface was changed markedly by the abundant deposits (**Figure 8.b** and **8.c**). Spherical calcium phosphate was precipitated onto the CF/HAp fibre surface, proving its bioactivity. No apatite crystals were developed on the surface of pure carbon nonwovens (no. 5).

Conclusions

Needle punching is the oldest and best-established method of forming nonwoven textile materials. The results obtained in this work confirm that the needle-punching method can be used to produce three-dimensional bioactive scaffolds for tissue engineering, the application of which allows to avoid post treatment operations (by coating or covering nonwovens with films) that are usually necessary in order to obtain a bioactive character of implants. We demonstrated that HAp particles were successfully incorporated into PAN fibres and were visible on the surface of carbon nonwoven fibres after the carbonisation process. Comparison of the physical properties, such as the porosity, thickness, apparent density, air permeability, tensile strength, and propagation of the ultrasonic wave velocity was made for all samples. The results showed the significant effect of the carbonisation process on nonwoven properties, with special reference to pore size diameters and mechanical properties. It is well known that the pore size, porosity, pore size distribution, pore interconnectivity and the reproducibility of pores are crucial parameters for scaffolds as they provide optimal spatial and nutritional conditions for cells and determine the successful integration of natural tissue and the scaffold. All pores in the carbon nonwovens were interconnected, and the mean pore size for the modified PAN/HAp samples was about 100 µm. It was observed that the PAN/HAp nonwoven, with its compact structure, (sample no. 4) showed a higher tensile strength than other PAN/HAp nonwovens, the SBF test proved its bioactivity. The tensile strength of the PAN/HAp nonwoven was comparable to those of type I collagen fibres (the major organic component of ECM in bone). The results presented indicate that carbon nonwovens made of fibers modified with hydroxyapatite may be a prospective material for the production of bioactive implants which can es-

establish direct chemical bonds with bone tissue after implantation.



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Faculty of Textile Engineering and Environmental Protection

The Faculty was founded in 1969 as the Faculty of Textile Engineering of the Technical University of Łódź, Branch in Bielsko-Biała. It offers several courses for a Bachelor of Science degree and a Master of Science degree in the field of Textile Engineering and Environmental Engineering and Protection.

The Faculty considers modern trends in science and technology as well as the current needs of regional and national industries. At present, the Faculty consists of:

- The Institute of Textile Engineering and Polymer Materials, divided into the following Departments:
 - Polymer Materials
 - Physics and Structural Research
 - Textile Engineering and Commodity
 - Textile Engineering and Commodity
 - Applied Informatics
- The Institute of Engineering and Environmental Protection, divided into the following Departments:
 - Biology and Environmental Chemistry
 - Hydrology and Water Engineering
 - Ecology and Applied Microbiology
 - Sustainable Development
 - Processes and Environmental Technology
 - Air Pollution Control



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