Evaluation of Biodegradability of Surgical Synthetic Absorbable Suture Materials: An *In Vitro* Study

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Two types of sutures made by two different biopolymers were tested in terms of hydrolytic biodegradation in phosphate buffered saline solution, which simulates the physiological conditions, varying the pH of the medium and the immersion time. The determination of the degradation rate was conducted by measuring the weight loss of the sutures. The study revealed that both investigated surgical suture exhibit quite different hydrolytic degradation at various immersion times and pH level, a more intense degradation being recorded in the alkaline environment.

Keywords: biodegradable polymer, suture materials, weight loss, SEM

Due to their great advantages over natural materials, the synthetic bioabsorbable sutures became widely used in the surgical field [1-3]. An ideal suture must be sterile, easy to handle, causing minimal tissue damage or minimal tissue reactions, to provide high tensile strength, to present a favorable absorption profile and to be resistant to infection. Polymers used as biomaterials in sutures' fabrication must meet certain characteristic, namely appropriate tensile strength and Young moduls value, capillarity, handling, biocompatibility and biodegradability. The most widely used homopolymers and copolymers in obtaining absorbable sutures are polydioxanone, polyglycolic acid, the copolymer of glycolic acid and trimethylene carbonate, the copolymer of caprolactone and glycolide and the copolymer of glycolic acid and lactic acid [4].

After implanting the absorbable suture, the polymers used in the synthesis process, are broken down by enzymal and hydrolytic process. Strength, mass loss profiles and biocompatibility of the absorbable sutures are the most important characteristics in the degradation and absorption processes [4,5]. The literature shows that the strength and mass loss profiles of the absorbable sutures depend not only on the chemical differences between the used biopolymers, but also on some intrinsic and extrinsic factors, such as electrolytes, *p*H, applied stress, temperature, microorganisms and tissue type [6].

The objective of this work was to make a comparative analysis of the way and degradation degree of surgical sutures commonly used in orthopedic surgery. Two types of sutures made by two different biopolymers were tested in terms of hydrolytic biodegradation in phosphate buffered saline solution, which simulates the physiological conditions, varying the *p*H of the medium and the immersion time.

Phosphate buffered solution (PBS) was selected as testing medium because it is accepted as usual medium for testing the hydrolytic biodegradation of polymers for medical devices [7-11].

Experimental part

Materials and methods

In vitro degradation behaviour of the investigated absorbable sutures materials (table 1) in phosphate buffer saline solution of *pH* 4.45; 7,45 and 10 was monitored at 2, 4, 6 and 8 weeks.

In order to evaluate the structural characteristics of the biopolymers we used Infrared Spectroscopy [12, 13]. The infrared absorption spectra were recorded on a FT-IR JASCO 6200 spectrometer operating in the ATR (Attenuated Total Reflectance) mode. Spectra were recorded with a resolution of 4 cm^{-1}, by the superposition of approximately 160 spectra.

Scanning electron microscopy was used in order to observe the changes in the surface morphology of the suture materials during degradation [14-16]. Analyses were performed on a scanning electron microscope QUANTA F INSPECT.

In the physiological environment, the biodegradable polymers are expected to degrade by hydrolysis of the ester bonds. The evaluation of the in vitro degradation rate of the sutures was made by measuring the weight loss of the experimental samples.

Experimental samples were stored in standard atmosphere for two hours prior the initial weight measurement (W0) and then immersed in 25 mL PBS. After each pre-established time interval (2, 4, 6 and 8 weeks), the samples were taken out and rinsed with deionized water, kept in standard atmosphere until they

Samples	Suture material	Suture type
P1	Polyglycolic acid coated with polycaprolactone and calcium stearate	Absorbable multifilament suture
P2	Copolymer of polyglycolic acid and ε-polycaprolactone	Absorbable monofilament suture

Table 1
ABSORBABLE SUTURE MATERIALS USED IN EXPERIMENTAL STUDY

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reached constant weight and weighed (W, weight after immersion). The weight loss (WL %) was calculated using the equation 1:

$$WL\% = \frac{W_0 - W}{W_0} * 100 \quad (1)$$

Encoding the samples depending on pH value and immersion time is presented in table 2.

Results and discussions

Fourier Transform Infrared Spectroscopy (FTIR)

The spectra obtained by FTIR for the P1 and P2 samples are presented in figures 1 and 2. For the P1 sample peaks located at 2992.02 cm^{-1} and 2867.63 cm^{-1} were assigned to the asymmetric respectively symmetric stretching vibration of $–CH_3$ from calcium stearate, while peaks located at 2958.27 cm^{-1} and 2849.31 cm^{-1} are assigned to the asymmetric respectively symmetric stretching vibration of $–CH_2$ from polycaprolactone. At 1735.62 cm^{-1} we can identify a strong peak due to the carbonyl stretching vibration, at 1239.04 cm^{-1} a peak assigned to the asymmetric C-O-C stretching vibration from polycaprolactone and between 1400-1500 cm^{-1} peaks due to the bending vibration of $–CH_3$ and $–CH_2$ groups.

Regarding the spectrum of P2 sample we can easily identify at 1740.44 cm^{-1} a strong peak due to the carbonyl stretching vibration, at 2958.27 cm^{-1} and 2868.59 cm^{-1} peaks due to the asymmetric $-CH_2$ stretching vibration and symmetric $-CH_2$ stretching vibration, at 1299.79 cm^{-1} a peak assigned to the backbone C–C and C–O stretching modes in the crystalline phase, at 1259.29 cm^{-1} a peak due to the asymmetric C-O-C stretching vibration, at 1147.44 cm^{-1} a peak due to C-O stretching vibration coupled to the C-C stretching vibration.

Weight loss determinations

The evolution of the degradation rates by measuring the weight loss of the experimental sutures are shown in figures 3-5.

From the figures 3-5 it can be noted that in both investigated sutures, the degradation rate measured by weight loss, increases with immersion time, regardless of *pH* value and the type of material. Depending on the *p*H of the PBS medium, the highest values of the degradation rate were registered at *p*H=10, a basic medium accelerating the degradation process. Depending on the material type, the highest values of the degradation rate are noticed in the case of the surgical suture made of polyglycolic acid. This is because polyglycolic acid has a strong hydrophilic character unlike polycaprolactone, which is a polymer with weak hydrophilic character. An important issue was emphasized at sutures made of the copolymer of the caprolactone with glycolic acid. Although polycaprolactone is a weakly hydrophilic polymer, with a degradation time of more than 24 months, in the case of this copolymer with glycolic acid, the degradation rate increases significantly. This is because the combination with polyglicolic acid results in a less crystalline plastic and so the degradation rate is higher than that of the homopolymer.

Sample type	Immersion time	pH value		
		pH=4.45	pH=7.45	pH=10
P1	2 weeks	$P1\_2_{4.45}$	$P1\_2_{7.45}$	$P1\_2_{10}$
P2	2 weeks	$P2\_2_{4.45}$	$P2\_2_{7.45}$	$P2\_2_{10}$
P1	4 weeks	$P1\_4_{4.45}$	$P1\_4_{7.45}$	$P1\_4_{10}$
P2	4 weeks	$P2\_4_{4.45}$	$P2\_4_{7.45}$	$P2\_4_{10}$
P1	6 weeks	$P1\_6_{4.45}$	$P1\_6_{7.45}$	$P1\_6_{10}$
P2	6 weeks	$P2\_6_{4.45}$	$P2\_6_{7.45}$	$P2\_6_{10}$
P1	8 weeks	$P1\_8_{4.45}$	$P1\_8_{7.45}$	$P1\_8_{10}$
P2	8 weeks	$P2\_8_{4.45}$	$P2\_8_{7.45}$	$P2\_8_{10}$

Table 2
EXPERIMENTAL SAMPLES ENCODING

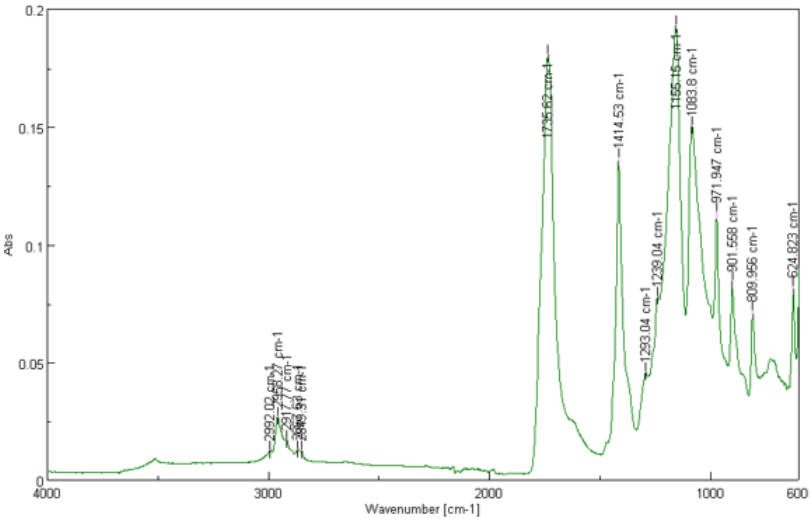

Fig. 1 FT-IR spectrum of the P1 sample

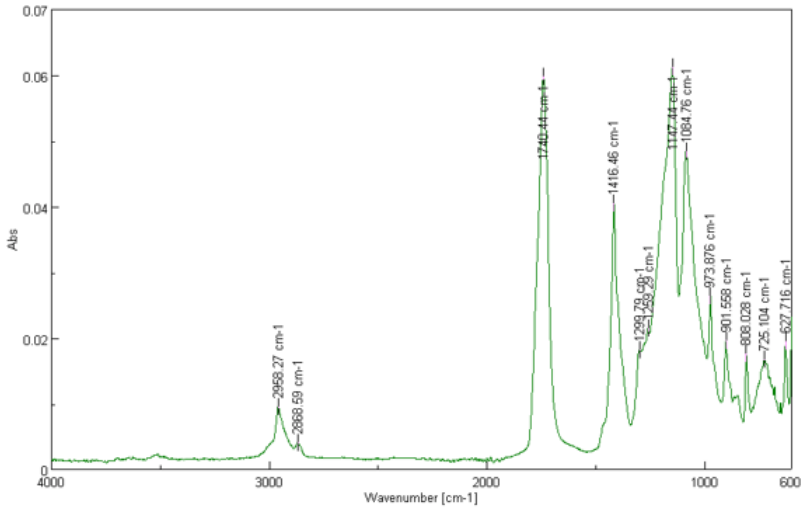

Fig. 2 FT-IR spectrum of the P2 sample

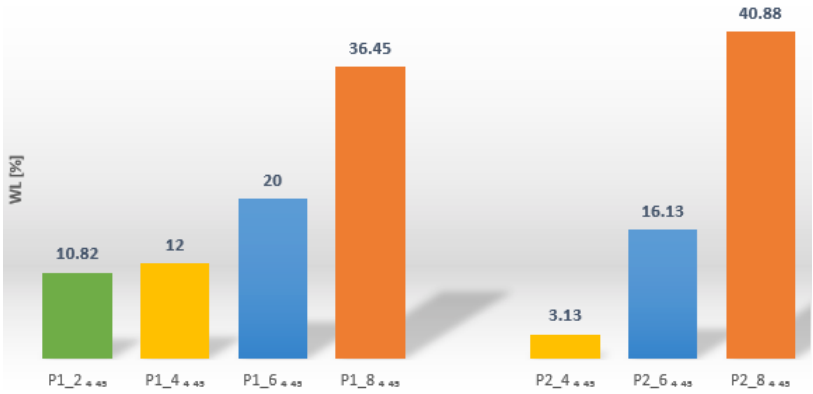

Fig. 3 The evolution of the sutures degradation rates at *p*H=4.45 for 2,4,6 and 8 weeks

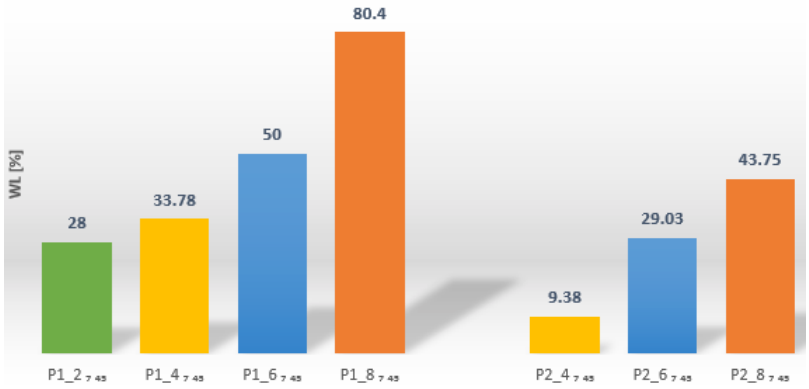

Fig. 4 The evolution of the sutures degradation rates at *p*H = 7.4 for 2,4,6 and 8 weeks

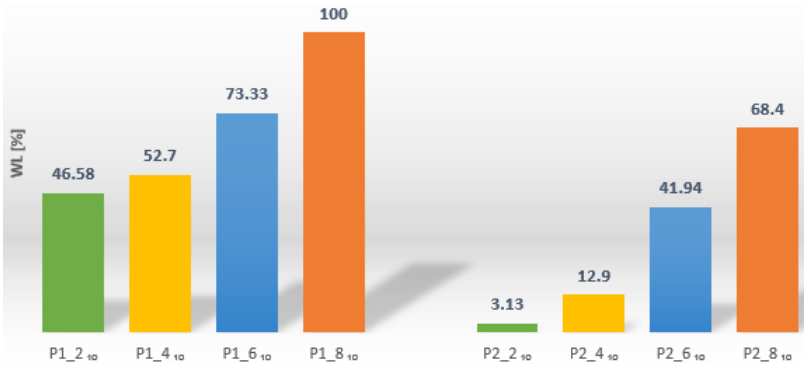

Fig. 5 The evolution of the sutures degradation rates at *p*H = 10 for 2,4,6 and 8 weeks

Scanning electron microscopy (SEM)

The SEM micrographs recorded on the experimental samples before immersion highlights the morphology of each type of investigated suture (figs. 6-7).

In the case of sample P1 is confirmed the type of the investigated suture, which is braided multifilament suture coated with polycaprolactone and calcium stearate. In the case of sample P2, the suture is monofilament with homogeneous structure.

The SEM micrographs obtained after immersion in PBS solution confirms the results obtained at determination of the degradation rate measured by the weight loss. It is noted that the degradation of the suture wires increases with immersion time, regardless of *p*H and type of material. Also, there is a more pronounced degradation at *pH* 10.

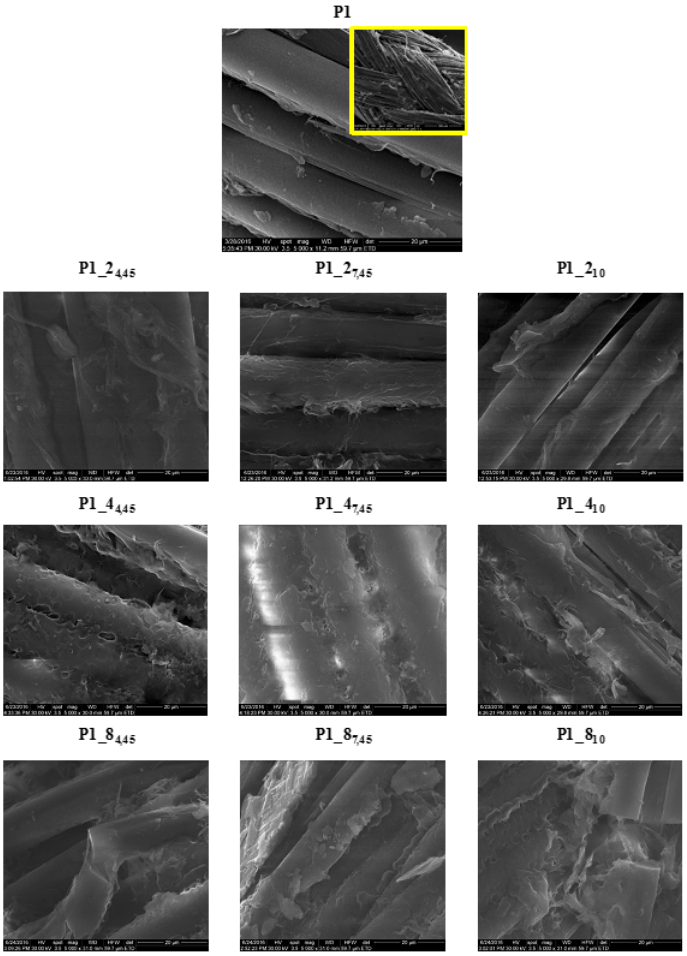

Fig. 6 SEM images of the sample P1 in different degradation time and *p*H level

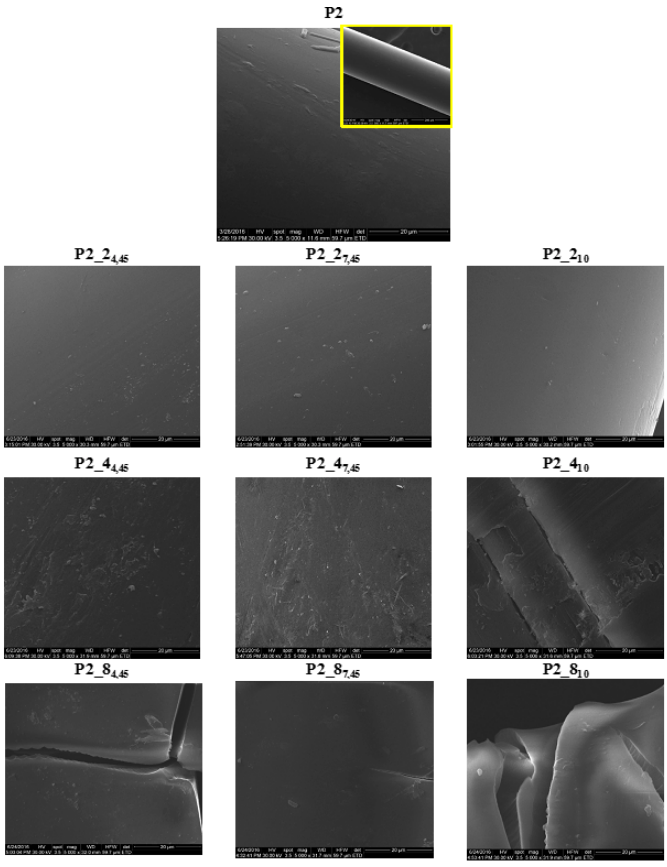

Fig. 7 SEM images of the sample P2 in different degradation time and *p*H

Conclusions

The degradation of the investigated surgical sutures immersed in PBS solution is the result of a process of ydrolysis. Some important issues, which affect the degradation rate, are the polymer hydrophilicity and mlecular weight. Water can easy penetrate hydrophilic olymers, and react easily with functional groups on these polymers. Regarding the polymer molecular weight, the lower molecular weight is, the higher degradation rate is.

This study demonstrates the influence of some extrinsic factors concerning the degradation rate of the biopolymeric surgical sutures. The experimental results obtained revealed that both investigated surgical sutures exhibit qite different hydrolytic degradation at various immersion times and *pH* values. They degrade faster in a hight-alkaline medium, and the sutures made by polyglicolic acid shown a total degradation at p*H* of 10.

In conclusion, we consider that the degradation rate of the biodegradable suture can be tailored to different applications depending on the type and area of surgery.

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