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Mechanical properties of biodegradable polymer sutures coated with bioactive glass

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Combining commercially available Polyglactin 910 (Vicryl®) sutures with bioactive glass powder offers new possibilities for application of composite materials in tissue engineering. Commercial bioactive glass (45S5 Bioglass®) powder was used to coat Vicryl® sutures and the tensile strength of the sutures was tested before and after immersion in simulated body fluid (SBF) as a means to assess the effect of the bioactive glass coating on suture degradation. Different gauge lengths (126.6 and 111.6 mm) and strain rates (2.54, 11.4 and 25.4 mm/min) were tested. The tensile strength of composite sutures was slightly lower than that of as-received Vicryl® sutures (404 MPa versus 463 MPa). However after 28 days immersion in SBF the residual tensile strength of the coated sutures was significantly higher, indicating a protective function of the Bioglass® coating. The tensile strength results were similar for the different gauge lengths and strain rates investigated. A qualitative explanation for the effect of bioactive glass coating on polymer degradation is offered.

Introduction

Synthetic bioabsorbable sutures based on polyglycolic acid (PGA), polylactic acid (PLA) and polyglycolide lactide copolymers (PGLA) have been studied extensively for more than 30 years [1–7]. These sutures (e.g. Polyglactin 910 sutures) exhibit three important advantages: minimal tissue reactions, good mechanical properties and they can be manufactured in a reproducible and precise way [3]. The main biodegradation mechanism in these materials is the hydrolytic mechanism via the scission of ester linkages [4–7]. Several factors that influence the biodegradation mechanisms of Polyglactin 910 include pH of the degradation media [3, 4, 6], type of electrolytes in the media [6], external stress/strain applied [6, 7], γ-irradiation [5, 6, 8, 9], presence of enzymes, bacteria and lipids [10] and plasma modification [10]. Most studies have been carried out in buffered saline systems.

Tensile properties of Polyglactin 910 sutures reported at different pH levels of the saline medium [4] show that maximum retention of tensile properties is around pH 7.0–7.44. Either strong acidic or alkaline solutions resulted in smaller percentages of strength retention [4].

Polyglactin 910 has shown a very good performance as sutures in wound healing [1–7]. Thus, this material might be useful as a polymer substrate for incorporating a bioactive substance, such as bioactive glass, for tissue engineering applications. Bioactive glasses have been shown to form a mechanically strong bond to bone and to soft tissues as well as being resorbable at rates slower than Polyglactin [11, 12]. A particular commercial bioactive glass (45S5 Bioglass®), which contains 45% SiO_2, 24.5% Na_2O, 24.5% CaO and 6% P_2O_5 in weight percentage has been in clinical use for over 15 years predominantly to enhance bone repair and reconstruction [11]. 45S5 Bioglass® is considered a class *A* bioactive material, being both osteoconductive and osteoproductive [11]. It has been also shown that 45S5 Bioglass® can form a mechanically strong interface with soft connecting tissues in several animal models [12, 13]. Bonding occurs by the rapid formation of a thin layer of hydroxycarbonate apatite (similar to biological apatite) on the glass surface when implanted or in contact with biological fluids [11].

Thus, our hypothesis is that the bioactivity of biodegradable polymer substrates, including sutures, can be enhanced by coating with Bioglass® layers. Another reason to combine a bioactive glass with biodegradable polymers is to control the morphological changes of these materials, that involves hydrolytic degradation and yield acidic residues and compositional changes [14]. Our aim is to combine the bioactive glass particles that form a thin coating layer on the polymer surfaces, to control biodegradation of the polymer and inhibit possibly heterogeneous degradation, thus resulting in improved structural integrity of the polymer scaffold over times of exposure to body fluids.

Indeed the combination of biodegradable polymers and bioactive inorganic materials, e.g. bioactive glass and hydroxyapatite, forming hybrid composites is being

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increasingly explored for tissue engineering applications [15–17]. However, there have not been previous attempts to combine a bioactive material with surgical sutures and to assess the degradation behavior of the coated sutures in contact with simulated body fluids (SBFs).

The tensile strength of Polyglactin 910 sutures coated by 45S5 Bioglass® particles was measured at various strain rates (2.54, 11.4 and 25.4 mm/min) before and after immersion in SBF at 37 °C as means to assess the effect of the bioactive glass coating on the degradability of the sutures.

Materials and methods

Violet braided resorbable 3/0 Vicryl® sutures (2 metric) were obtained commercially from Ethicon Inc. (Edinburgh, Scotland). These are braided from fine filaments of glycolide and lactide polymers. The mean diameter of as-received sutures is 0.33 mm. The melt-derived Bioglass® 45S5 powder (particle size $< 5\,\mu m$) was provided by US Biomaterials (Alachua, Florida, USA).

The composites were prepared by a simple layer-pressing procedure, as described elsewhere [18]. Briefly, a weighed amount of Bioglass® powder was placed on a flat-surface steel plate forming a uniform layer. The sutures (75 cm in length) were then placed on the glass powder layer. Subsequently, a new glass powder layer was added, covering the sutures. A second, similar plate was placed on the top and the assembly was set in a uniaxial press. A two-step cycle was followed, involving pressures of 100 and 160 MPa applied in each case for 5 min. The time and the level of pressure applied were optimized by a trial-and-error approach to produce as uniform coating as possible. The optimization was based mainly on visual and scanning electron microscopy (SEM) observation of Bioglass® particles attached mechanically to the surface of the sutures. As discussed below, however, this simple processing technique did not provide a constant Bioglass® layer thickness. The amount of Bioglass® attached on the suture surfaces was around 0.08 g. The uniformity of the Bioglass® coatings was characterized by SEM.

As-received and coated Vicryl® sutures were then immersed in 75 ml of SBF. To ensure zero biological degradation due to the presence of bacteria or fungi 1% v/v of an antibiotic–antimycotic (Life Technologies Ltd., UK) was added. The preparation method of SBF has been previously described in the literature [19]. The pH of the solution was maintained constant at 7.25. Samples immersed in SBF were incubated at 37 °C in an orbital shaker (175 rpm) for 3, 7, 14, 21 and 28 days. The SBF solution was replaced twice a week. Once removed from incubation the samples were rinsed first in pure ethanol and secondly rinsed with deionised water. The samples were mechanically tested immediately after they were rinsed with deionised water. The tensile properties were measured using an Instron™ universal testing machine. The grips used were specially designed for testing of fibers. Samples were tested at two different gauge lengths (126.6 and 111.6 mm) and three different strain rates (2.54, 11.4 and 25.4 mm/min) following a procedure similar to that used by Chu [4]. For each case at least three samples were tested. A very high reproducibility of failure load data was achieved and in all cases the relative error was $< 2\%$. The diameter of as-received and coated Vicryl® sutures was measured under a confocal optical microscope to an accuracy of 1%.

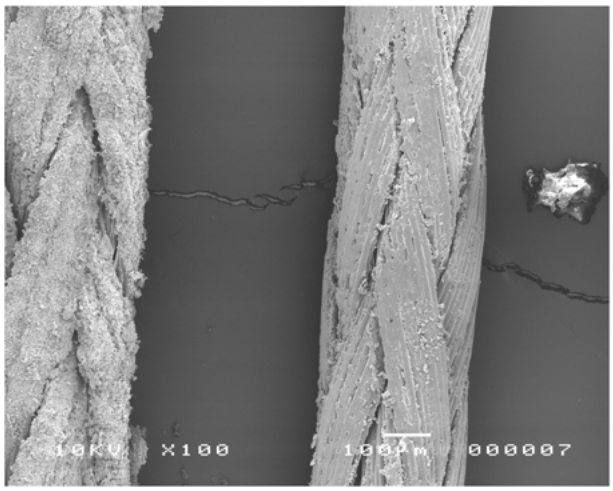

Figure 1 SEM micrograph showing sutures in the as-received state (right) and after coating with Bioglass® powder (left).

Results

Fig. 1 shows SEM micrographs of as-received and coated sutures. Although there is acceptable covering of the suture surfaces by Bioglass® particles, the structure of the glass coating and its thickness were not uniform, which is a consequence of having used, dried and agglomerated powders to prepare the samples.

The average maximum tensile strength of non-degraded Vicryl® sutures was found to be in the range 464–508 MPa ($\pm$ 3%), depending on the strain rate, as shown in Table I. These values are in agreement with

TABLE I Tensile properties of as-received Vicryl® sutures after the indicated incubation times in SBF at the given strain rates (gauge length = 111.6 mm)

Incubation time (days)	Average max. tensile load (N) ($\pm$ 0.8 N)	Average max. tensile strength (MPa) (Max. rel. error 3%)	Strain rate (mm/min)
0	36.1	464	2.54
3	37.2	478	2.54
7	35.6	458	2.54
14	27.5	353	2.54
21	17.5	226	2.54
28	0	0	2.54
0	39.2	502	11.4
3	39.1	502	11.4
7	36.1	469	11.4
14	29.6	382	11.4
21	17.4	226	11.4
28	0	0	11.4
0	39.5	508	25.4
3	39.7	509	25.4
7	37.1	476	25.4
14	29.5	379	25.4
21	12.9	166	25.4
28	0	0	25.4

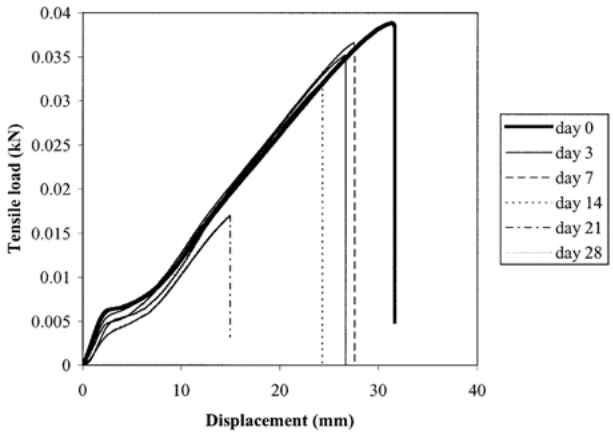

Figure 2 Tensile load versus displacement of Vicryl® sutures at the given immersion times in SBF (gauge length = 126.6 mm and strain rate = 2.54 mm/min). The sigmoidal shape of the curves is typical for this type of materials.

literature results [7]. The tensile strength was found to be little affected by changes in gauge length. Generally, smaller gauge lengths resulted in slightly higher tensile strength values due to the well-known fact that smaller test volumes exhibit less statistically distributed defects and therefore have a lower probability of failure. The gauge length also had little influence on the tensile strength of Bioglass®-coated Vicryl® sutures (data not shown in this paper). However, significant changes occurred for both types of materials as a function of immersion time in SBF, as shown in Tables I and II.

Figs 2 and 3 show the change of tensile load against displacement after different degradation times of as-received and coated sutures, respectively. Results for gauge length of 126.6 mm and strain rate of 2.54 mm/min are plotted. The curves for other gauge lengths and strain rates are similar. The shape of the curves has the typical sigmoidal shape reported by other researchers [3, 4] for similar sutures. The initial region represents the alignment of fibers that are stretched along their axis. The subsequent decrease in slope is attributed to failure of individual fibers before final fracture of the braided structure. This load–displacement shape was observed for all samples independently of gauge length, strain rate

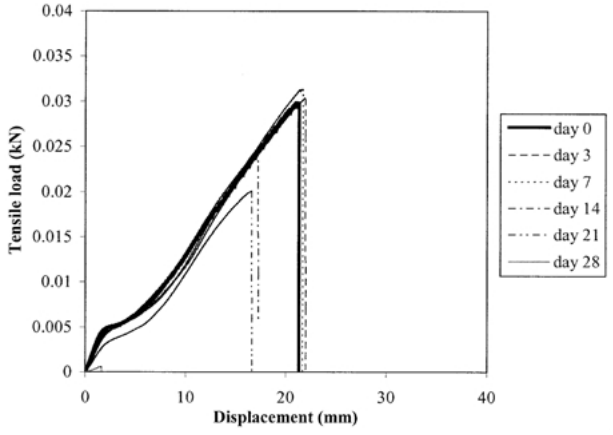

Figure 3 Tensile load versus displacement of Bioglass®-coated Vicryl® sutures at the given immersion times in SBF (gauge length = 126.6 mm and strain rate = 2.54 mm/min). The shape of the curve remains identical to that of as-received Vicryl® sutures (Figure 2), but fracture load and displacement are lower.

TABLE II Tensile properties of Vicryl® sutures coated with Bioglass® powder after the indicated incubation times in SBF at the given strain rates (gauge length = 111.6 mm)

Incubation time (days)	Average max. tensile load (*N*) (± 0.8 N)	Average max. tensile strength (MPa) (Max. rel error 3%)	Strain rate (mm/min)
0	31.5	404	2.54
3	31.3	402	2.54
7	31.6	406	2.54
14	27.2	349	2.54
21	18.1	233	2.54
28	1.2	16	2.54
0	34.0	437	11.4
3	34.7	445	11.4
7	31.8	408	11.4
14	28.6	367	11.4
21	12.2	157	11.4
28	1.5	19	11.4
0	33.1	426	25.4
3	32.6	418	25.4
7	30.3	388	25.4
14	28.6	367	25.4
21	14.9	191	25.4
28	1.0	12	25.4

and degradation time. The materials at low strain exhibit pure elastic deformation reaching a yield point at around a tensile load of ca. 5–8 N. Then, as-received and coated sutures deform plastically up to a maximum tensile strength (ultimate tensile strength) until they finally break at a slightly lower stress. It is worth noting, that comparing the load–displacement curves of as-received and Bioglass®-coated sutures, the yield strength, tensile strength and strain to failure are slightly lower for the composite samples. Similar results were obtained for the other gauge length tested.

Comparing the measurements at different strain rates for the as-received sutures, changes in tensile strength occurred as expected, i.e. slightly higher values were measured for higher strain rates, as reported in Table I. On the other hand, as degradation proceeds, the results became random with slightly higher or lower values for higher strain rates. The same behavior was observed for Bioglass®-coated composite sutures, as shown in Table II. Comparing the results in Table I and Table II, in general as-received Vicryl® sutures exhibit slightly higher tensile strengths than Bioglass®-coated samples. A possible reason for this behavior could be the technique used to prepare the composites, where the sutures may have been mechanically damaged during the powder pressing operation by the hard glass particles.

Fig. 4 shows the retained average tensile strength of as-received sutures versus the degradation time at different strain rates and a gauge length of 111.6 mm. Similar results were obtained for a gauge length of 126.6 mm. The loss of tensile strength after 21 days of immersion time in SBF ranges between 43 and 63%. The effect of strain rate for shorter immersion times is not clear from the data in Fig. 4, however. At low strain rates the strength retention values are slightly higher. This was also found for the larger gauge length (126.6 mm) and for the Bioglass®-coated sutures (Fig. 5). The mechanisms explaining this behavior are addressed below.

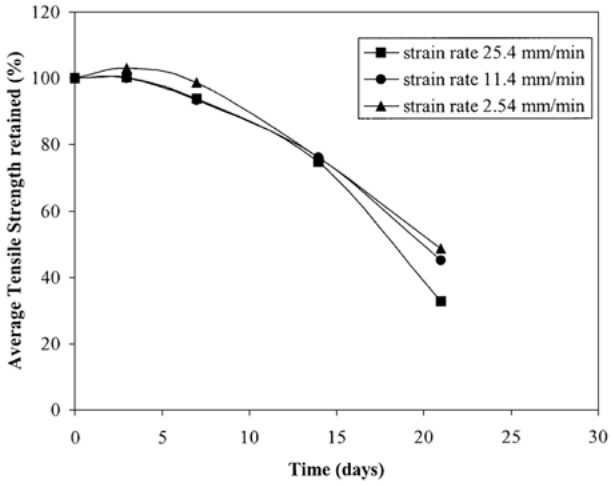

Figure 4 Retained average tensile strength of Vicryl® sutures as function of days immersed in SBF (gauge length = 111.6 mm).

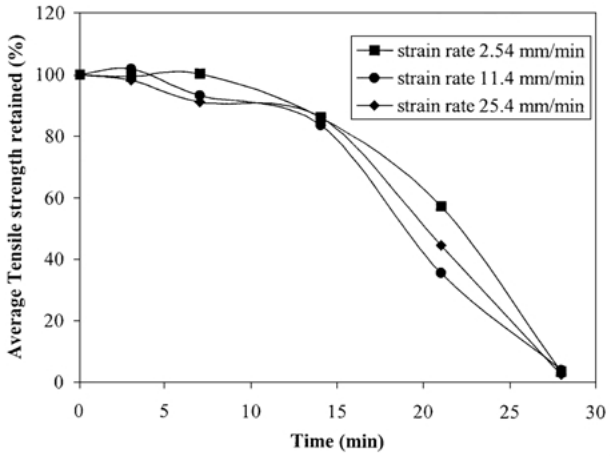

Figure 5 Retained average tensile strength of Bioglass®-coated Vicryl® sutures as function of days immersed in SBF (gauge length = 111.6 mm).

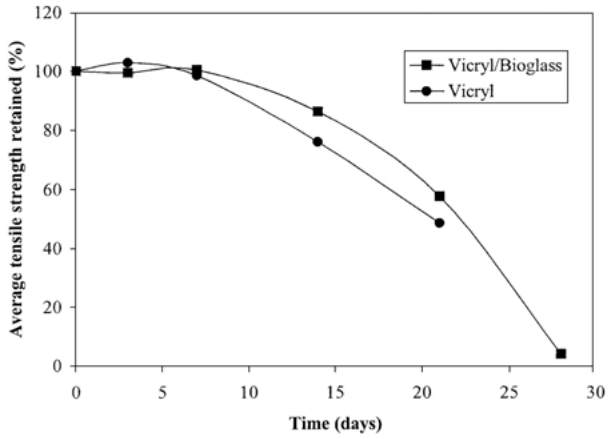

Figure 6 Retained average tensile strength of as-received and Bioglass®-coated Vicryl® sutures as function of days immersed in SBF (gauge length = 111.6 mm, strain rate = 2.54 mm/min).

In Figs 6 and 7 the performance of as-received and coated sutures is compared in terms of retained average tensile strength as function of days immersed in SBF. The plots are for different gauge lengths and strain rates to highlight the negligible effect of these two variables, at least in the range of values investigated here.

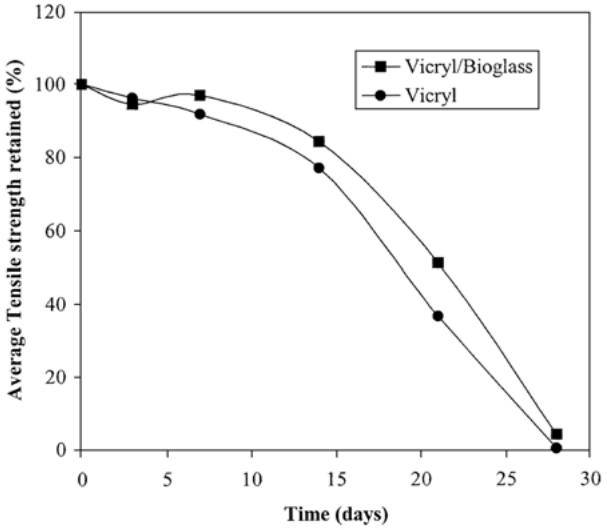

Figure 7 Retained average tensile strength of as-received and Bioglass®-coated Vicryl® sutures as function of days immersed in SBF (gauge length = 126.6 mm, strain rate = 11.4 mm/min).

Discussion

The degradation mechanisms of biodegradable polymers have been described on the basis of a microfibrillar model of the fiber structure [20]. The model considers a microfibril with alternated crystalline and amorphous regions aligned in the direction of the fiber axis. Within the microfibril some polymer chains will pass through several crystalline and amorphous regions and some polymer chains will simply return back to form chain folds. The crystalline regions are composed of chains in a certain order of conformation whereas the amorphous regions are composed of chain folds, chain ends and tie-chain segments. The tie-chain segments can be formed in two types: interfibrillar and intrafibrillar. Their main role is to tie crystalline regions together and to support and transmit the tensile loads from the amorphous to the crystalline regions. As reported in the literature [14], degradation in biodegradable polymers occurs first in the amorphous region followed by the crystalline region. By immersing a material into an aqueous solution, the water molecules penetrate first the amorphous regions but they do not penetrate the crystalline regions. As degradation proceeds, tie-ends, chain-ends and chain folds degrade into fragments until a point where the fragments can dissolve in the aqueous medium. Consequently, spaces that were occupied by the chain segments now become vacant and can be visible as cracks. It is believed that these cracks are initiated on the surface of the fiber and grow along the circumference of the fiber to the interior. As microcracks are formed more water molecules penetrate these areas. The microcracks propagate quite fast into the interior until failure occurs. Consequently, degradation introduces chain scission resulting in a lesser degree of entanglement of the long chain molecules and therefore lower axial elastic moduli and tensile strength, as illustrated in Figs 2 and 3. The remaining chain segments in the amorphous regions can now move easier and reorganize themselves so that an increase in crystallinity can be observed [3]. The degree of crystallinity reaches a maximum at the end of the first stage. The

loss of tensile strength during the first degradation stage is also at a maximum. During the second stage, hydrolysis destroys the crystalline lattice resulting in reduced crystallinity whereas the complete loss in tensile strength from the value at the end of the first stage of degradation to zero occurs at a slower rate. It has been reported that the first stage of degradation is predominant during the 21-day immersion period, whereas the second stage becomes more important afterwards [3]. This is in good agreement with our experimental results, shown in Figs 4 and 5. Particularly, Fig. 4 shows that lower strain rates lead to slightly higher strength retention values. This can be attributed to the following: as chain scission occurs due to degradation, shorter chains find the time to reorganize themselves at lower strain rates leading to slightly higher retention strengths. However, this effect seems to be random. The percentage of strength retention decreases or increases slightly after three days, then as degradation proceeds the retention decreases slowly so that the shape of the curve becomes slightly sigmoidal. After 14 days of degradation the curve slope changes suddenly leading to a significant drop of strength retention that lasts another 14 days until the tensile strength becomes zero. Similar effects were observed for the Bioglass®-coated sutures (Fig. 5). In general, composite (coated) sutures behaved similarly to as-received sutures. Dramatic change of the tensile strength retention occurs here also after 14 days of immersion in SBF. It seems that between the 3rd and the 14th day of degradation the strength retention decreases quite slowly, reaching – specifically at higher strain rates – a plateau which results in a much faster reduction of strength retention between the 14th and the 28th day of degradation.

The direct comparison between the mechanical performance in tension of as-received and coated sutures (Figs 6 and 7), leads to the conclusion that the composite sutures perform slightly better at long degradation times. Specifically, in almost all cases the strength retention of the coated sutures decreases at a slower rate after 7 days of degradation. A detailed microscopical investigation of the morphology of the as-received and coated sutures at different degradation stages has indicated a significant effect of the presence of a glass coating on the degree of surface degradation after 7 days of immersion in SBF [21].

These results are thus quite encouraging in that the mechanical performance of commercially available biodegradable Vicryl® sutures in contact with SFB can be altered by coating them with a layer of Bioglass® powder, even if the homogeneity of the coatings reported here was far from optimal. Qualitatively, the Bioglass® coating is acting as a protective ''shield'' that affects both the extent and rate of degradation of the sutures. The rapid exchange of protons in water for alkali in the glass will provide a pH buffering effect at the polymer surface. Also, dissolution of the glass will lead to nucleation and growth of a polycrystalline hydroxyapatite layer on the sutures.

One way to quantitatively assess the effect of the Bioglass® coating on polymer degradation would be by using gel permeation chromatography (GPC), which allows the determination of molecular weight changes. Application of this technique to the present commercial sutures was not possible however, as they had been γ-irradiated and therefore extensive cross-linking has occurred. As a result, the polymer is not solvable in common solvents making it impossible to determine the molecular weight changes by GPC.

From this experimental study there are still several other questions that remain open and need further investigation. For example, there is the need to assess the effect of the thickness and the microstructure of the Bioglass® coating (e.g. size, shape and distribution of coating porosity) on the degradation rate and strength retention. In this regard optimized processing techniques must be developed which allow a high control of the Bioglass® coating thickness and of the homogeneity of the coating microstructure. Novel techniques should also allow to coat resorbable polymer fibers without affecting their initial mechanical properties. A step forward in this regard is the development of wet techniques, i.e. using a stable dispersion of glass particles in water and coating by controlled slurry-dipping, which is the focus of current research.

It is interesting to point out that the combination of Bioglass® and biodegradable sutures is not limited to glass powder coatings. Using Bioglass® fibers, which are currently under development [22], would also provide an interesting approach. Bioglass® fibers exhibit low fracture strength and brittle behavior, in comparison to the high strength and viscoelastic behavior of polymer sutures, as measured in this study. However combining both resorbable sutures and Bioglass® fibers may open a new way to the design of hybrid bioactive and biodegradable 3D scaffolds for tissue engineering, the aim being to achieve better mechanical performance and higher bioactivity of the constructs.

Conclusions

Composites based on biodegradable Vicryl® sutures coated with Bioglass® particles were prepared and mechanically tested after being immersed in SBF. The aim was to assess the effect of the bioactive glass layer on the mechanical properties of the sutures before and after degradation. The tensile properties of the sutures in the as-received condition were slightly reduced by the glass coating. However, the strength retained by the sutures at different degradation times was improved by the Bioglass® coatings, indicating a qualitative effect of the bioactive material on the mechanism of polymer degradation. Further research on the optimization of the coating process and on the degradation mechanisms active on coated sutures is expected to give essential information for the use of these materials in tissue engineering and wound healing.

Acknowledgments

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