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Tough and strong porous bioactive glass-PLA composites for structural bone repair

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Abstract

Bioactive glass scaffolds have been used to heal small contained bone defects but their application to repairing structural bone is limited by concerns about their mechanical reliability. In the present study, the addition of an adherent polymer layer to the external surface of strong porous bioactive glass (13–93) scaffolds was investigated to improve their toughness. Finite element modeling (FEM) of the flexural mechanical response of beams composed of a porous glass and an adherent polymer layer predicted a reduction in the tensile stress in the glass with increasing thickness and elastic modulus of the polymer layer. Mechanical testing of composites with structures similar to the models, formed from 13–93 glass and polylactic acid (PLA), showed trends predicted by the FEM simulations but the observed effects were considerably more dramatic. A PLA layer of thickness ~400 μm , equal to ~12.5% of the scaffold thickness, increased the load-bearing capacity of the scaffold in four-point bending by ~50%. The work of fracture increased by more than 10,000%, resulting in a non-brittle mechanical response. These bioactive glass–PLA composites, combining bioactivity, high strength, high work of fracture and an internal architecture shown to be conducive to bone infiltration, could provide optimal implants for healing structural bone defects.

Keywords

Bioactive glass composite scaffold; Structural bone repair; Mechanical properties; Work of fracture; Finite-element modeling

1. Introduction

Bone defects are a common occurrence in orthopedic surgery, resulting from trauma, malignancy, infection and congenital disease. Clinically, these defects are reconstructed through the use of various bone grafts. Whereas small contained bone defects are repairable

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with a wide variety of commercially-available osteoconductive and osteoinductive filler materials [1, 2], the repair of large (critical size) defects in structural bone is challenging. The available treatments such as bone allografts, autografts and porous metals are limited by costs, availability, durability, infection risk, donor site morbidity and uncertain healing. Consequently, there is a clinical need for synthetic biocompatible materials that can replicate the strength, bioactivity, morphology, porosity and load-bearing ability of living bone.

Bioactive glasses have been studied and developed as one such material because they are biocompatible, osteoconductive, convert to hydroxyapatite in vivo, heal readily to host bone and soft tissues and they are amenable to fabrication into porous three-dimensional architectures [3–7]. However, most bioactive glass scaffolds created by conventional methods have shown low strength, typically compressive strength in a range comparable to human trabecular bone (2 to 12 MPa) [5]. This low strength and a brittle mechanical response typical of glass have limited the application of bioactive glass scaffolds to the healing of contained or non-loaded bone defects.

Recent studies using additive manufacturing (rapid prototyping or three-dimensional printing) methods have shown the ability to create bioactive glass scaffolds with a porous gridlike microstructure which have vastly improved compressive strength, comparable to the strength of human cortical bone (100 to 150 MPa) [8–12]. Despite a low flexural strength (10 to 15 MPa), much smaller than that of cortical bone (100 to 150 MPa) [9], these strong porous scaffolds have shown the ability to repair critical size segmental defects in a rat or rabbit femoral model [13, 14]. Modification of the grid-like microstructure to produce a porosity gradient in the scaffolds resulted in a significant improvement in the flexural strength (34 ± 5 MPa) without sacrificing the ability to support bone infiltration [15]. As bending is an important loading mode in structural bone, this enhancement in flexural strength should improve the mechanical reliability of bioactive glass scaffolds in vivo. However, these scaffolds with high compressive and flexural strength still showed a brittle mechanical response.

The ability to modify or improve the mechanical response of porous ceramics or glass by infiltrating the pore volume or coating the pore surface with a biodegradable polymer composed of polycaprolactone (PCL) or polylactic acid (PLA) has been reported in several studies [16–22] and discussed in the literature [23, 24]. The polymer phase has been reported to act as a sealant to bond together the walls of pre-existing flaws in the surface of the brittle material and limit the initiation or growth of cracks, providing a defect healing mechanism [16–19]. Polymers with high elastic modulus, such as PLA, can also assist in partially supporting the applied load, thereby reducing the effective load on the ceramic or glass phase.

While complete infiltration of the pore space of biocompatible ceramics or glass scaffolds with a biodegradable polymer can improve their mechanical response, the absence of porosity could, at least initially, limit bone infiltration and integration of the scaffolds in vivo. An alternative approach, particularly for bioactive materials, is judicious coating of the pore surface of the scaffolds to provide an optimal combination of improved mechanical response, bioactivity and porosity conducive to bone or tissue infiltration. Coating the pore

surface of weak bioactive glass (45S5) scaffolds with poly(D, L-lactic acid) has been found to produce a small improvement in their compressive and flexural strength but a remarkable improvement in their work of fracture [21]. In comparison, scaffolds of 45S5 glass fully infiltrated with molten PLA did not show a significant improvement in mechanical properties over the coated scaffolds due to reaction between the molten PLA and the glass during the infiltration process [25]. When encapsulated in a sheath of PLA (~500 μm), the average flexural strength of bioactive glass (13–93) scaffolds (external diameter = 4.2 mm; porosity ~20%) formed from thermally-bonded fibers (300 to 700 μm) increased by approximately 2.5 times, from ~50 MPa to ~120 MPa [22]. The PLA sheath also produced a drastic modification of the brittle mechanical response of the glass scaffolds, resulting in a considerable improvement in the work of fracture and to an overall “non-brittle” mechanical response.

Based on our previous studies which showed the ability to create strong porous bioactive glass (13–93) scaffolds with a grid-like microstructure conducive to supporting bone infiltration [15], the present study was undertaken to modify their brittle mechanical response and improve their toughness. Finite element modeling (FEM) was used to simulate the mechanical response in flexural loading of model composites composed of a porous glass beam with a grid-like structure and an adherent polymer layer on the top and bottom surface of the beam. Then composites with microstructures mimicking the model structures were created from 13–93 bioactive glass and PLA, and tested in four-point bending to evaluate their mechanical response.

A novel feature of the present study is the combination of the bioactive glass scaffold geometry and the placement of the polymer layer. The beam geometry was used for the scaffold because of its simplicity, in addition to being the most commonly-used geometry in four-point bending tests. Placement of the PLA layer on only the top and bottom surface of the glass scaffold can provide the capacity to heal the most critical surface flaws in bending while preserving the bioactivity of the uncoated internal surface of the scaffold. Based on the results of a previous study in which thermally-bonded 13–93 glass fibers were encapsulated in a 500 μm PLA sheath [22], the present study also evaluated the effect of varying the PLA layer thickness: 200, 400 and 800 μm , equal to ~6.25, 12.5 and 25% of the bioactive glass beam thickness.

2. Materials and methods

2.1 Design of composite structures

Three different glass structures, with the external geometry of a beam (23.0 mm long $\times$ 4.7 mm wide $\times$ 3.2 mm thick) and composed of orthogonal glass filaments (diameter = 330 μm) in adjacent layers, were analyzed in FEM simulations (Fig. 1). The structures were composed of 6 layers of filaments in the long direction of the beam (referred to as “long” filaments) and 7 layers of filaments in the orthogonal direction (“short” filaments). These structures were selected on the basis of a previous study [15]. The structure designated L1S1 was composed of a uniform grid-like arrangement of the long (L) and short (S) filaments in the adjacent layers (Fig. 1a). In comparison, the structure designated L3S1 was derived from the L1S1 structure by redistributing some of the long filaments in the structure from the

interior to the top layer and bottom layer but keeping the number of filaments the same (Fig. 1b). The L4S1 structure was obtained from the L3S1 structure by adding additional long filaments to the second layer from the top and bottom, making these layers identical to the top and bottom layer (Fig. 1c). All three structures had the same arrangement of the short filaments.

The redistribution of the long filaments resulted in a porosity gradient in the L3S1 and L4S1 structures, with the top and bottom layers of the structure having a lower porosity than the interior. FEM simulations and four-point bending experiments previously showed that the flexural strength of scaffolds with the L3S1 and L4S1 structures (22 ± 3 MPa and 34 ± 5 MPa, respectively) was significantly higher than the L1S1 scaffold (15 ± 5 MPa) [15].

Composite structures were designed by adding adherent layers of polymer to the top and bottom surfaces of the beams (Fig. 2). Layers of thickness 200, 400 and 800 μm were used. Based on observations from experiments described later, a specific depth of polymer infiltration into the macroporous surface of the glass structure was assumed. Twelve different structures were analyzed in the FEM simulations, based on the structure of the glass scaffold and the thickness of the polymer layer (Table I).

2.2 Finite element simulations

Finite element simulations were carried out using ABAQUS/CAE 6.14-1 software (Dassault Sytemes Simulia Corp., Providence, RI). Twelve models shown in Table 1 were analyzed in four-point bending. The filaments in all the models were assumed to be composed of a dense homogeneous material with a Young's modulus of 70 GPa and a Poisson's ratio of 0.25, values which are comparable to those of a silicate glass [26]. The polymer layer in the models was assumed to be dense and homogeneous with a Young's modulus of 0.65 GPa (measured for the PLA sheets used in this study) and a Poisson's ratio of 0.36 (given in the manufacturer's specifications for PLA). The polymer layer was assumed to have an elastic response and to be perfectly bonded to the glass surface, without any delamination under flexural loading.

For the simulations in flexural loading, 4 rigid cylinders (diameter = 2.0 mm) were generated to simulate the fixture used in the four-point bending test, as described previously [15]. The two upper cylinders, each 5 mm from the midline, served as the inner span, while the two lower cylinders, 10 mm from the midline, served as the outer span. The cylinders were in contact with the model composites from the beginning of the simulations and the contact was assumed to be hard (no interpenetration between the materials) and frictionless.

The FEM models were partitioned into several parts in order to improve the meshing and apply the proper boundary conditions. The FEM models were discretized with an approximate mesh size of 120 μm and quadratic tetrahedral elements were used. This size was selected based on a previous study [15] as it provided a good balance between computing time and convergence to reliable results. The middle of the cross section perpendicular to the x axis was fixed at the x axis direction ($U1 = 0$) to prevent the scaffold from sliding in the x axis direction. In addition, the middle of the cross section perpendicular to the y axis was fixed at the y axis direction ($U2 = 0$) to prevent sliding in the y axis

direction during bending. The two lower cylinders of the fixture were fixed in all directions ($U_1 = U_2 = U_3 = 0$) and an evenly distributed load of 100 N was applied to the two upper cylinders.

A desktop computer with 16 CPU cores and 128 GB of RAM was used for the simulations. A typical CPU time to complete a simulation was ~4 h. As a brittle material, glass can withstand much higher stresses in compression than in tension. Consequently, the highest tensile stress within the inner span of the model was used as the primary parameter in the load-carrying capacity analysis.

2.3 Creation of bioactive glass – PLA composites

Bioactive glass (13–93) scaffolds with the geometry of a beam and microstructure corresponding to the L1S1, L3S1 and L4S1 structure in the FEM simulations were fabricated using a robotic deposition technique as described in detail elsewhere [15]. Briefly, as-received 13–93 glass frit (Mo-Sci Corp., Rolla, MO) was ground to form particles (~1 μm), mixed with a 20 wt. % Pluronic F-127 binder solution in water (Sigma-Aldrich; St. Louis, MO) to form a paste (40 vol. % glass particles) and extruded using a robocasting machine (RoboCAD 3.0; 3-D Inks, Stillwater, OK). After drying at room temperature, the scaffolds were heated in O_2 at a rate of $0.5\text{ }^\circ\text{C min}^{-1}$ to $600\text{ }^\circ\text{C}$ to burn out the processing additives and sintered for 1 h at $700\text{ }^\circ\text{C}$ (heating rate = $5\text{ }^\circ\text{C min}^{-1}$) to densify the glass filaments. The external dimensions of the as-fabricated bioactive glass beams were 23.0 mm long $\times$ 4.7 mm wide $\times$ 3.2 mm thick.

PLA sheets were prepared by dissolving as-received pellets (molecular weight = 150 kDa; Ingeo™ biopolymer 4043D; NatureWorks LLC, Minnetonka, MN) in CHCl_3 , casting the solution on a flat surface and evaporating the solvent in a fume hood for 24 h in dry flowing air. Sheets of thickness ~200, 400 and 800 μm were formed from solutions with different PLA concentration. The amorphous or crystalline nature of the PLA sheets was checked using X-ray diffraction (XRD) (D/mas 2550 v; Rigaku; The Woodlands, TX, USA). Dog-bone shaped specimens were cut from the sheets (800 μm thick) using a standard fixture and tested in tension at deformation rates of 2 to 10 mm min^{-1} to measure their Young's modulus.

A controlled and reproducible procedure was used to bond a PLA layer to the top and bottom surface of the bioactive glass beams. One surface of a PLA sheet was wetted with CHCl_3 and applied to one surface of the scaffold (in the xy plane). Then a pressure of ~20 kPa was applied to keep the sheet in contact with the scaffold surface and the system was allowed to dry in a fume hood for 24 h in dry flowing air. The process was repeated to bond a PLA layer to the opposite surface of the bioactive glass scaffold. Excess PLA was trimmed off the edges of the scaffold.

2.4 Measurement of bioactive glass–PLA adhesive strength

The adhesive strength between the PLA layer and the surface of the bioactive glass scaffold was measured using an adhesion testing apparatus (Romulus universal test equipment; Quad Group Inc., Spokane, WA). PLA sheets of thickness 800 μm were cut into disks (diameter = 2.6 mm) and bonded to one surface of the bioactive glass scaffold using the same process

described earlier for the preparation of the composite scaffolds. Then an aluminum rod (2.6 mm in diameter) was bonded to the PLA disk using an adhesive capable of withstanding a stress >100 MPa. The system was held in the testing apparatus and the aluminum rod was pulled at a rate of $\sim 2 \text{ mm min}^{-1}$ until delamination of the PLA layer occurred (Fig. 3). Five samples were tested. The adhesive strength was determined from the maximum recorded load, the total interfacial area between the scaffold and the PLA disc and the actual area of the PLA disk in contact with the glass scaffold. As the scaffold had a porous grid-like microstructure, the area of the PLA disc in actual contact with the glass was lower than the total interfacial area.

2.5 Mechanical testing in four-point bending

Testing of the as-fabricated bioactive glass–PLA composites in four-point bending was performed using a fully articulated fixture (outer span = 20 mm; inner span = 10 mm) at a crosshead speed of 0.2 mm min^{-1} using a 2 kN load cell. The stress was applied in the z direction (the same direction used in the FEM simulations). Eight samples in each group were tested according to the procedure described in ASTM C1674-11 and the load versus deflection curves were recorded. Apparent fracture energy was determined from the area under the load versus deflection curves using software (Image J; National Institutes of Health, USA). For composites with a PLA layer thickness of 400 μm or 800 μm , the tests were terminated prior to fracture because the sample came into contact with the side of the four-point bending fixture. After testing, the samples were examined in a scanning electron microscope (SEM) (S-4700; Hitachi, Tokyo, Japan) and by optical microscopy (KH-8700; Hirox-USA, Inc., Hackensack, NJ).

2.6 Statistical analysis

The data are presented as a mean $\pm$ standard deviation (SD). Analysis for differences between groups was performed using one-way analysis of variance (ANOVA) with Tukey's post hoc test. Differences were considered significant for $p < 0.05$.

3. Results

3.1 Finite element simulations

The deformation of a model composite composed of a glass scaffold with the L1S1 structure and an adherent polymer surface layer of thickness 200 μm in four-point bending is shown in Fig. 4. The deformation of the composite beam was magnified 15 times (scale factor = 15) for ease of observation. Similar deformation profiles were found for the other composites that were analyzed. The distribution of the maximum principal stress (S , Max. Principal (Abs.)) in the deformed structure is also shown in color, as represented by the scale in Fig. 4. Negative tensile stress was found in the top layers of the scaffold as they were in compression during bending. The highest tensile stress was present in the long filaments of the bottom layer in the y direction. As the polymer layer could deform more easily than the glass filaments due to its much lower Young's modulus, the tension was much lower in the PLA layer. According to the maximum principal stress theory (Tresca), failure will occur when the maximum principal stress in a system reaches the maximum stress at the elastic limit in simple tension. This is approximately valid for brittle materials such as glass.

Consequently, particular attention was paid to differences in the magnitude and distribution of the stress in the bottom layers of the glass filaments of the various models.

The magnitude and distribution of the tensile stress in the long filaments at the middle of the bottom layers of the models are shown in Fig. 5. The highest tensile stress was present at the contact region between the long and short filaments. Figure 6 shows the magnitude of the highest tensile stress for the various models. For models with same polymer layer thickness, the highest tensile stress was in the order $L1S1 > L3S1 > L4S1$, which is in consistent with the results of previous simulations for the same glass structures without a polymer layer [15]. For models with same glass structure but different polymer layer thickness, the simulations predicted a small decrease in the magnitude of the tensile stress (or the area with same magnitude of the tensile stress) in the glass filaments with increasing thickness of the PLA coating. For a PLA layer thickness of 800 μm , the highest stress in the filaments decreased by less than 10% when compared to the scaffold without a PLA layer.

The effect of the Young's modulus of the polymer layer was also analyzed. For a model composed of a glass scaffold with the L1S1 structure and a polymer layer of thickness 200 μm , Fig. 7 shows the magnitude and distribution of the tensile stress in the glass filaments near the middle of the bottom layers. The tensile stress decreased with increasing Young's modulus of the polymer layer.

3.2 Mechanical response of as-fabricated bioactive glass–PLA composites

X-ray diffraction patterns of the PLA sheets showed strong well-defined peaks corresponding to crystalline PLA and broad low-intensity bumps indicative of an amorphous PLA phase (results not shown). The results indicated that the PLA layers in the composites were semi-crystalline and not amorphous. The measured stress versus strain curves and the Young's modulus determined from the initial linear region of the curves at crosshead speeds of 2, 5 and 10 mm/min are shown in Fig. 8. As generally observed for polymeric materials, the Young's modulus increased with increasing strain rate. The PLA sheet showed an elastic response at strains less than 1 to 2% for the strain rates used, which was considerably larger than the strain to failure of the bioactive glass ($\sim 0.1\%$).

Optical images of as-fabricated bioactive glass–PLA composites with a PLA layer thickness of ~ 200 , 400 and 800 μm are shown in Fig. 9. The PLA layers appeared to have a uniform thickness and they partially infiltrated the macroporous surface of the glass scaffolds (the space between the short glass filaments at the surface). Higher magnification examination in the SEM revealed the presence of a few isolated pores within PLA layer. These pores existed through the entire thickness of the 200 μm PLA layer which meant that the glass surface immediately below these through-thickness pores was bare. Apparently, the 200 μm PLA layer was not sufficiently thick to provide a complete uniform layer over the entire surface of the glass scaffold. No through-thickness pores were observed in the 400 μm or 800 μm PLA layer. The measured adhesive strength between the PLA layer (800 μm thick) and the glass scaffold was 1.0 ± 0.1 MPa based on the total interfacial area. Because of the porous nature of the scaffold surface, the actual contact area between the PLA and the glass was estimated to be approximately two-thirds of the total interfacial area. If the actual contact area was used, the adhesive strength was ~ 1.5 MPa.

After the adhesion test, the surface of the PLA which was previously adherent on the glass scaffold showed a structure composed of bumps and valleys that replicated the macroscopic structure of the glass scaffold surface. This indicated some degree of mechanical interlocking between the PLA and the glass scaffold (Fig. 10a). The glass scaffold showed the presence of residual adherent PLA which but this was situated mainly at the junctions between the two outermost layers of glass filaments (Fig. 10b). The “smooth” surface of the glass filaments previously in contact with the PLA layer showed little residual PLA.

Typical load versus strain curves in four-point bending for composites formed from 13–93 bioactive glass scaffolds with the L1S1 structure and different PLA layer thickness are shown in Fig. 11a. The glass scaffold without a PLA layer showed a typical brittle mechanical response (rapid failure after an elastic response). The applied load showed a peak followed by a rapid decrease as the structure fractured and was unable to support the load. For a PLA layer thickness of 200 μm , the peak load increased by ~50% but a brittle mechanical response was still observed. In comparison, the composites with a PLA layer thickness of 400 or 800 μm showed a dramatic change in mechanical response. First, the peak load increased, showing the ability of the composite to support a higher load than the glass scaffold alone (without a PLA layer). When compared to the glass scaffold alone, the peak load increased by ~95% and ~110%, respectively, for composites with PLA layer thickness of 400 and 800 μm . Second, the composite did not fail in a brittle manner. Instead, the composite was able to support a considerable load at strains larger than 5% when the experiments were terminated. The load versus strain curves showed additional peaks and valleys, and the load supported by the composite when the experiments were terminated was even higher than the load at the first peak.

The mechanical response of composites formed from a bioactive glass scaffold with a gradient in porosity (L3S1 or L4S1 structure) and PLA layer thickness of 400 or 800 μm showed trends similar to the composite with the L1S1 glass structure and PLA layers of the same thickness (Fig. 11b). The measured load at the first peak in the load versus strain curve (referred to as the load-bearing capacity) is summarized in Fig. 12 for all the composites. For a PLA layer thickness of 400 μm , the load-bearing capacity of the composites was 72 N, 84 N, and 100 N, respectively, for scaffolds with the L1S1, L3S1, and L4S1 structure. In comparison, the load-bearing capacity of the bioactive glass scaffolds alone was 37 N, 46 N and 75 N, respectively, for the L1S1, L3S1 and L4S1 structure (Table II).

The area under the load versus deflection curve, referred to as the work of fracture (which is a measure of the toughness), for the bioactive glass–PLA composites with a PLA layer thickness of 400 μm or 800 μm showed a large increase over that for the bioactive glass scaffold alone or the composite with a 200 μm PLA layer. Figure 13a shows data for the work of fracture, G_{max} , measured at a strain corresponding to the first peak in the load versus deflection curve. This work of fracture is often referred to as the resilience. The data are normalized to the work of fracture for the bioactive glass scaffold with L1S1 structure (equal to 1.3×10^{-6} J). For composites composed of scaffolds with the L1S1, L3S1 and L4S1 structure and a PLA layer thickness of 400 μm , G_{max} increased to 465%, 710% and 970%, respectively, of the value for G_{max} of the glass scaffold. The work of fracture, $G_{0.05}$, at a strain of 5% (when the four-point bending tests were terminated) are shown in Fig. 13b

for the composites with the 400 μm and 800 μm PLA layer. The data are normalized to G_{max} for the corresponding bioactive glass scaffold (L1S1, L3S1, or L4S1). The work of fracture for the composites with a 400 μm PLA layer increased dramatically to over 10,000% of the value for G_{max} of the glass scaffold.

SEM images of a composite formed from a 13–93 glass scaffold with the L1S1 structure and 200 μm PLA layer after testing in four-point bending are shown in Fig. 14a, b. The crack initiated at the contact region between the long and short filaments, the same region that showed the highest tensile stress in the FEM simulations. Then the crack propagated further (upward in the z direction) via a path along similar contact regions between the long and short filaments. The initiation and propagation of the crack resulted in a brittle failure mode, similar to the glass scaffold alone, presumably because the composite was unable to support the applied load.

Examination of composites formed from a bioactive glass scaffold (L1S1, L3S1 or L4S1 structure) and PLA layers of thickness 400 or 800 μm after testing to different points of the load versus strain curve showed the presence of more than one crack in most specimens (Figure 14c, d, e). In general, the occurrence of these cracks was correlated with successive peaks in the load versus strain curve. The first crack was correlated with the first peak and additional cracks were correlated with subsequent peaks. Thus the composite did not fail catastrophically due to the propagation of a single crack. Instead, smaller cracks formed periodically in different parts of the glass scaffold which allowed the composite to maintain its overall integrity as a whole and to continue to support a considerable load.

4. Discussion

The results of the present study showed the ability to significantly improve the load-bearing capacity and to dramatically enhance the work of fracture of strong porous bioactive glass scaffolds in four-point bending. This was achieved by the addition of an adherent PLA layer to the external surface of the scaffolds. Finite element simulations of model composites predicted an improvement in the load-bearing capacity of the glass scaffold in four-point bending with increasing thickness and elastic modulus of the PLA layer. Mechanical testing of 13–93 bioactive glass–PLA composites created with structures similar to the models showed effects that were considerably larger and more dramatic than those predicted by the simulations. The load-bearing capacity of the bioactive glass scaffold increased significantly with the addition of the PLA layer. For a sufficiently thick PLA layer (400 μm , equal to 12.5% of the scaffold thickness), the work of fracture increased dramatically, resulting in a non-brittle mechanical response. These composites, combining bioactivity, high strength, high work of fracture and an internal architecture previously shown to be conducive to bone infiltration, could provide optimal synthetic implants for healing structural bone defects.

4.1 Mechanical response of bioactive glass–PLA composites

For a given applied load in four-point bending, the simulations predicted a reduction in the tensile stress in the bottom filaments of the glass structure with increasing elastic modulus and thickness of the polymer (Fig. 5–7). However, the predicted reduction in the tensile stress was <10% of the stress in the glass beam alone (without the polymer layer) for the

thickest PLA layer (800 μm) used in the simulations. This reduction in the tensile stress in the glass resulted from load sharing by the PLA layer but the effect was small due to the much smaller elastic modulus of the PLA layer when compared to the glass scaffold.

The structure and geometry of the as-fabricated bioactive glass–PLA composites showed a good approximation to the assumptions of the FEM simulations. The glass filaments were almost fully dense and the PLA can be considered to be elastic in four-point bending of the bioactive glass–PLA composites due to its much higher elastic limit (1 to 2%) when compared to the failure strain of the glass scaffold ($<0.1\%$). The PLA layer adhered to the glass filaments at the surface of the glass scaffold (Fig. 9), with a measured adhesive strength of 1.0–1.5 MPa in a standard pullout test. After the pullout test, residual PLA was found to adhere to the glass scaffold mainly at the junctions between the two outermost layers of glass filaments (Fig. 10). While perfect interfacial bonding assumed in the FEM simulations was apparently not present, these observations confirmed that the PLA layer adhered to the glass scaffold.

In general, the observed mechanical response of the bioactive glass–PLA composites in four-point bending showed trends predicted by the FEM simulations but the effects were considerably stronger and more dramatic. The addition of a 400 μm PLA layer improved the load-bearing capacity of the bioactive glass scaffold with the L1S1 structure by $\sim 100\%$. In comparison, the tensile stress in the glass filaments in the bottom layer of the models was predicted to decrease by only $\sim 3\%$ in the FEM simulations. Similar trends were observed for bioactive glass scaffolds with a porosity gradient (L3S1 and L4S1 structures). The difference between the predicted and observed load-bearing capacity of the bioactive glass–PLA composites is attributed to deviations between the model assumptions and the as-fabricated composites.

In the FEM model, the glass was assumed to be fully dense and devoid of flaws. In comparison, the mechanical response of brittle materials, particularly when subjected to a tensile stress, is known to be strongly dependent on the presence of surface flaws and internal microstructural flaws. In the present study, the bioactive glass filaments in the as-fabricated scaffolds were almost fully dense, containing only a few isolated fine pores. Consequently, the difference between the predictions of the FEM simulations and the measured mechanical response could be attributed to the presence of surface flaws in the glass filaments. In the creation of the bioactive glass–PLA composites, the surface of the PLA layer was softened by wetting with CHCl_3 and applied to the surface of the glass scaffold by means of a low applied load. In addition to adhering to the glass surface, the PLA presumably acted a sealant to bond together the walls of pre-existing flaws in the glass surface and limit growth or initiation of flaws, providing a flaw-healing mechanism as discussed previously [16–19].

A striking feature of the observed mechanical response was that the addition of the PLA layer did not just improve the load-bearing capacity of the bioactive glass scaffold. The PLA layer also dramatically modified the type of mechanical response. A PLA layer thickness of 400 μm ($\sim 12.5\%$ of the glass scaffold thickness) significantly enhanced the work of fracture, in addition to significantly enhancing the load-bearing capacity of the bioactive glass

scaffold (Fig. 12, 13). Increasing the PLA layer thickness from 400 μm to 800 μm did not provide a significant improvement in the load-bearing capacity. This is a further indication that the dominant effect of the PLA layer was to heal the surface flaws in the glass and, as predicted by the FEM simulations, that the PLA in the macropores at the surface of the scaffold had little effect on the strength.

The improvement in mechanical performance observed in the present study appears to be due mainly to healing of flaws in the surface of the glass scaffold that is under the maximum tensile stress rather than to a reduction in the average stress due to load sharing by the PLA layer. Consequently, this type of strengthening effect would not be as effective in loading modes such as compression and tension where the location of the maximum stress is not limited to the external surface of the glass scaffold. However, bioactive glass scaffolds similar to those used in the present study were previously shown to have high strength (comparable to human cortical bone) and good fatigue resistance in compression [9, 15]. Furthermore, the major loading modes in structural bones such as the long bones of the limbs are compression and bending. The high compressive strength of the glass scaffold together with the improvement in flexural mechanical response due to the PLA layer observed in the present study can provide an attractive combination of mechanical properties for structural bone repair.

Composites with a PLA layer thickness of 200 μm showed a brittle mechanical response typical of the glass scaffold, although there was a significant increase in the failure load (Fig. 12). As the same process was used to bond the PLA layers to the surface of the bioactive glass scaffolds, it is expected that the interfacial characteristics would be independent of the PLA layer thickness. The difference in mechanical response (brittle versus “non-brittle”) between the composite with the 200 μm PLA layer and those with the 400 μm or 800 μm PLA layer was attributed to processing effects. As described earlier, the 200 μm PLA layer was apparently not sufficiently thick to provide complete coverage over the entire surface of the scaffold. A few isolated through-thickness pores were present in the 200 μm PLA layer and the glass under these pores was essentially bare. These bare glass regions, although few in number and small in size, could nevertheless provide sources for brittle failure.

Another interesting feature of the results was the occurrence of multiple peaks and valleys in the load versus strain curves in four-point bending for composites with a PLA layer thickness of 400 or 800 μm (Fig. 11). These peaks were correlated with the initiation and propagation of successive cracks in the glass scaffold. Furthermore, at strains of up to 5% when the experiments were terminated, the load supported by the composite was comparable or even higher than the load at the first peak. For the glass scaffolds with a PLA layer thickness of 200 μm , a single crack initiated and propagated rapidly, resulting in failure following a peak load. In comparison, multiple cracks formed in succession in the glass scaffold with a PLA layer thickness of 400 or 800 μm , and the spacing between the two surfaces of the crack was much smaller than that for the glass scaffold with the 200 μm PLA layer (Fig. 14). Apparently, these multiple cracks with smaller separation between the crack surfaces enabled the composite to maintain its integrity and support a considerable load.

4.2 Potential of bioactive glass–PLA composites for structural bone repair

Although the tough and strong bioactive glass–PLA composites created in the present study were not evaluated in vivo, the bioactive glass scaffolds by themselves (without the PLA surface layer) were shown previously to have a microstructure conducive to bone infiltration in vivo [15]. When implanted in rat calvarial defects for 12 weeks, bioactive glass (13–93) scaffolds with microstructures corresponding to the L1S1, L3S1 and L4S1 structures were infiltrated with new bone. The amount of new bone that infiltrated the pore space of the scaffolds (35–37%) was not dependent on the microstructure of the three groups of scaffolds. Furthermore, the amount of new bone infiltration was higher than that observed for the same glass with a fibrous, trabecular or oriented microstructure [27]. As the PLA layers are present only on the external surface of the bioactive glass scaffold, the porous internal architecture can degrade, release ions that stimulate osteogenesis and support bone infiltration in more or less the same way as a glass scaffold without the PLA layer. The PLA layer on the external surface of the scaffold will eventually degrade away.

Scaffolds of 13–93 bioactive glass with a uniform grid-like (L1S1) microstructure (without a PLA layer), with a compressive strength of 72 ± 10 MPa and flexural strength of 15 ± 5 MPa, have already shown the capacity to heal segmental defects in rat femurs [13]. Scaffolds of 13–93 glass with a similar microstructure have also shown the capacity to heal segmental defects in rabbit femurs [14]. Bioactive glass (13–93) scaffolds with the L4S1 structure (without a PLA layer) have a higher compressive strength (88 ± 20 MPa) and a significantly higher flexural strength of (34 ± 5 MPa) than the scaffolds used in those studies. With the addition of a 400 μ m or 800 μ m PLA layer, the flexural load-bearing capacity of the L4S1 scaffold increased by –33% and ~50%, respectively, and its mechanical response changed dramatically from brittle to non-brittle. These composites formed from 13–93 bioactive glass with the L4S1 structure and an adherent PLA surface layer (400 or 800 μ m thick) can provide more reliable implants for structural bone repair.

In an aqueous phosphate solution such as the body fluid, the strength of bioactive glass scaffolds decreases with time as the glass converts to hydroxyapatite [9, 28]. In vivo, this decrease in strength should be compensated by an increase in strength due to new bone infiltration into the scaffold. In general, the degradation rate of the glass scaffold and the rate of new bone infiltration should be matched to maintain a sufficiently high strength to support physiologic loads. The degradation rate of the PLA layer in the composites created in the present study is also relevant to their mechanical performance in vivo. Ideally, the improvement in mechanical performance observed in the present study should be maintained at an acceptable level prior to sufficient bone infiltration of the bioactive glass scaffold. The effect PLA degradation on the mechanical response of the bioactive glass–PLA composites is currently being studied in vitro.

As a major focus of the present study was improving the flexural mechanical response of the bioactive scaffolds, a beam was used as the external geometry of the glass scaffolds in the FEM simulations and experiments. This geometry, coupled with the addition of a PLA layer to the top and bottom surface of the glass beam, makes the flexural mechanical response of the bioactive glass–PLA composites dependent on the loading direction. The present research will be extended to investigate the use structures in which the bioactive glass

scaffold has a circular cross section and angular symmetry, and the polymer layer is adherent on the circumferential surface of the scaffold. This will make the mechanical response independent of the loading direction along the long axis. The incorporation of an optimal porosity gradient in the radial direction of the bioactive glass scaffolds will be further analyzed using FEM while creation of the scaffolds with the requisite microstructure can be achieved using the robocasting technique described in the present study.

5. Conclusions

The addition of an adherent polylactic acid (PLA) layer to the external surface of strong porous bioactive glass (13–93) scaffolds with a grid-like microstructure significantly enhanced their load bearing capacity and dramatically improved their mechanical response. Finite element modeling (FEM) predicted an improvement in the load-bearing capacity of the glass scaffolds with increasing thickness and elastic modulus of an adherent polymer surface layer. Mechanical testing of 13–93 bioactive glass–PLA composites created with structures similar to the models showed trends predicted by the FEM simulations but the observed effects were considerably larger and more dramatic. A PLA layer thickness equal to ~12.5% of the scaffold thickness increased the load-bearing capacity of the bioactive glass scaffold by ~50% and dramatically enhanced its work of fracture by more than 10,000%, resulting in a non-brittle mechanical response. These bioactive glass–PLA composites, combining bioactivity, high strength, high work of fracture and an internal architecture known to be conducive to bone infiltration, could provide optimal implants for healing structural bone defects.

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Fig. 1.

Three different structures of the glass scaffold analyzed in finite element modeling. Each structure had the external geometry of a beam and was composed of 6 layers of long (L) filaments in the y direction and 7 alternating short (S) filaments in the x direction. The inset under each structure shows the arrangement of the long filaments in the xz plane.

Fig. 2.

Cross section of composite composed of glass scaffold and adherent polymer (PLA) surface layer of thickness 200, 400 and 800 μm . The polymer layer was assumed to partially infiltrate the top and bottom surface of the scaffold.

Fig. 3.
Schematic diagram illustrating the fixture used in the adhesive strength test.

Fig. 4.

Typical deformation of glass-polymer composite in four-point bending showing the magnitude and distribution of the stress in the glass filaments and polymer layer as determined by FEM simulations. The actual deformation of the model was magnified by a scale factor of 15 to better show the deformation. The scale on the left gives the maximum principal stress corresponding to specific colors. The box shows the region of most interest in the simulations.

Fig. 5.

Magnitude and distribution of the tensile stress in the long filaments of the bottom layers of different models when subjected to the same applied load in four-point bending. The scale on the left gives the maximum principal stress corresponding to specific colors for each row. The designations of the models are given in Table I.

Fig. 6. Highest tensile stress in the long filaments in the bottom layer of the models (normalized to the L1S1 model) determined from FEM simulations of four-point bending under an applied load of 100 N. The region of highest stress corresponded to the middle region of the bottom filaments between the inner span.

Fig. 7.

Magnitude and distribution of the tensile stress in the long filaments of the bottom layers of different models with polymer surface layers of thickness 200 or 800 μm and elastic modulus of 0.65, 2.1, 3.6 and 5.1 GPa when subjected to the same applied load in four-point bending. The scale on the left gives the maximum principal stress corresponding to specific colors.

Fig. 8. Stress vs. strain curves for PLA sheets of thickness 800 μm measured in tensile testing of dog-bone specimens (inset) at different deformation rates shown. The Young's modulus is also shown for each deformation rate.

Fig. 9. Side view of as-fabricated composites formed from bioactive glass (13-93) scaffolds with a grid-like (L1S1) structure and adherent PLA surface layer of thickness (a) 200 μm ; (b) 400 μm ; (c) 800 μm .

Fig. 10.

(a) Surface of PLA layer (previously in contact with the bioactive glass scaffold) after pullout test; (b) surface of bioactive glass scaffold after pullout test showing residual PLA adhering to the glass. The dashed lines in (a) outline grooves in the PLA layer which replicate the filaments (indicated by solid lines) at the surface of the glass scaffolds in (b).

Fig. 11. Typical load versus strain curves for bioactive glass (without PLA) and composites formed from bioactive glass (13–93) scaffolds with the different scaffold microstructures and PLA layers of different thickness.

Fig. 12.

Measured load at the first peak of the load versus strain curves in four-point bending for different groups of bioactive glass scaffolds (L1S1, L3S1 and L4S1) and composites formed from these scaffolds and PLA layers of thickness 200, 400 or 800 μm . (*significant difference between groups; $p < 0.05$).

Fig. 13.

(a) Work of fracture, $G_{\max}$, for the bioactive glass scaffolds and bioactive glass–PL A composites at a strain corresponding to the first peak in the load versus deflection curve (normalized to $G_{\max}$ for the L1S1 glass scaffold); (b) work of fracture, $G_{0.05}$, at a strain of 5% for the bioactive glass–PLA composites with a PLA layer thickness of 400 and 800 μm (normalized to $G_{\max}$ for the respective glass scaffold). (The data for the composites with the 400 μm and 800 μm PLA layer are significantly different from those for the corresponding glass scaffold; $p < 0.05$)

Fig. 14. SEM images of bioactive glass-PLA composites (yz plane) after testing to specific strains in four-point bending. (a) Glass scaffold with the L1S1 structure and PLA layer thickness of 200 μm ; a magnified image of the boxed area in (a) is shown in (b); (c) glass scaffold with the L4S1 structure and PLA layer thickness of 400 μm ; magnified images of the boxed areas in (c) are shown in (d) and (e). (The arrows in (a) and (c) indicate the PLA layer.)

Table I

Summary of models analyzed in FEM simulations.

Glass scaffold architecture	Polymer (PLA) layer thickness (μm)		
	200	400	800
L1S1	L1S1+200	L1S1+400	L1S1+800
L3S1	L3S1+200	L3S1+400	L3S1+800
L4S1	L4S1+200	L4S1+400	L4S1+800

Table II

Summary of measured mechanical properties in four-point bending for bioactive glass scaffolds (L1S1, L3S1 and L4S1 structures) and bioactive glass–PLA composites with PLA layer thickness of 200, 400 and 800 μm .

Material	Load-bearing capacity (N)	G_{max} (10^{-6} J)	$G_{0.05}$ (10^{-6} J)
L1S1	37 ± 11	1.3 ± 0.4	–
L1S1 + 200	55 ± 8	2.2 ± 0.1	–
L1S1 + 400	72 ± 10	5.9 ± 0.8	138 ± 12
L1S1 + 800	78 ± 10	8.9 ± 1.7	215 ± 18
L3S1	46 ± 8	1.5 ± 0.3	–
L3S1 + 400	84 ± 8	8.9 ± 0.8	182 ± 17
L3S1 + 800	97 ± 25	12.3 ± 0.9	231 ± 11
L4S1	75 ± 15	2.0 ± 0.4	–
L4S1 + 400	100 ± 18	12.2 ± 1.3	206 ± 11
L4S1 + 800	109 ± 10	16.8 ± 0.9	269 ± 16

Load-bearing capacity = Load at first maximum in load vs. strain curve

G_{max} = Apparent work of fracture, equal to the area under the curve at a strain corresponding to the first maximum in the load vs. deflection curve

$G_{0.05}$ = Apparent work of fracture, equal to the area under the curve at a strain of 5% in the load vs. deflection curve