

The effect of magnesium on bioactivity, rheology and biology behaviors of injectable bioactive glass-gelatin-3-glycidylpropyl trimethoxysilane nanocomposite-paste for small

Original

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The effect of Magnesium on bioactivity, rheology and biology behaviors of injectable bioactive glass-gelatin-3-glycidyoxypropyl trimethoxysilane nanocomposite-paste for small bone defects repair

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Abstract:	Injectable bioactive glass-based pastes represent promising biomaterials to fill small bone defects thus improving and speed up the self-healing process. Accordingly, injectable nanocomposite pastes based on bioactive glass-gelatin-3-glycidyoxypropyl trimethoxysilane (GPTMS) were here synthesized via two different bulk glasses containing 64SiO ₂ , 27CaO, 4MgO, 5P ₂ O ₅ (mol.%) or 64SiO ₂ , 31CaO, 5P ₂ O ₅ (mol.%). In particular, the effects of MgO on bioactivity, rheology, injectability, disintegration resistance, and cellular behaviors were investigated. The results showed that the disintegration resistance of the composite was improved by the replacement of MgO, thus leading to an increase in the amount of storage modulus (G') from 26800 to 43400 Pa, equal to an increase in the viscosity of the paste from 136 × 10 ³ to 219 × 10 ³ Pa.s. Since the release rate of ions became more controllable, the formation of calcite was decreased after immersion of the Mg bearing samples in the SBF solution. Finally, specimens' cytocompatibility was confirmed towards human osteoblasts by metabolic assay as well as the ability of human fibroblasts to penetrate within the pores of 3D composites was verified by a migration assay simulating the devices repopulation upon injection in the injured site.
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To the Editor of the *Ceramics International* Journal

Dear Prof. P. Vincenzini,

we submitted for a possible consideration on the *Ceramics International* Journal the original article entitled “*The effect of Magnesium on bioactivity, rheology and biology behaviors of injectable bioactive glass-gelatin-3-glycidylxypropyl trimethoxysilane nanocomposite-paste for small bone defects repair*” by Mehri Sohrabi, Bijan Eftekhari Yekta, Hamidreza Rezaie, Mohammad Reza Naimi-Jamal, Ajay Kumar, Andrea Cochis, Marta Miola and Lia Rimondini.

In this manuscript we demonstrated how the magnesium substitution in the bioactive glass composition can significantly improve the bioactivity, rheology and biology properties of injectable pastes aimed at small bone defects repair. Moreover, the presence of such “cells-friendly” polymer ameliorated the biological properties of the bioactive glass when used in direct contact with osteoblasts (MG63) and fibroblasts (HGF) cells, all involved in the self-healing process. Accordingly, we believe that the manuscript can be of interest for the readers of Your Journal.

Finally, we confirm that the manuscript is original, that it has been approved by all the Authors and that it is not actually under the evaluation of any other Journal.

Sincerely,

the corresponding Authors,

Prof. Bijan Eftekhari Yekta

A handwritten signature in black ink that reads 'B. E. Yekta'.

Prof. Lia Rimondini

A handwritten signature in black ink that reads 'Lia Rimondini'.

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The effect of Magnesium on bioactivity, rheology and biology behaviors of injectable bioactive glass-gelatin-3-glycidyoxypropyl trimethoxysilane nanocomposite-paste for small bone defects repair

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Abstract

Injectable bioactive glass-based pastes represent promising biomaterials to fill small bone defects thus improving and speed up the self-healing process. Accordingly, injectable nanocomposite pastes based on bioactive glass-gelatin-3-glycidyoxypropyl trimethoxysilane (GPTMS) were here synthesized via two different bulk glasses containing 64SiO₂, 27CaO, 4MgO, 5P₂O₅ (mol.%) or 64SiO₂, 31CaO, 5P₂O₅ (mol.%). In particular, the effects of MgO on bioactivity, rheology, injectability, disintegration resistance, and cellular behaviors were investigated. The results showed that the disintegration resistance of the composite was improved by the replacement of MgO, thus leading to an increase in the amount of storage modulus (G') from 26800 to 43400 Pa, equal to an increase in the viscosity of the paste from 136 × 10³ to 219 × 10³ Pa.s. Since the release rate of ions became more controllable, the formation of calcite was decreased after immersion of the Mg bearing samples in the SBF solution. Finally, specimens' cytocompatibility was confirmed towards human osteoblasts by metabolic assay as well as the ability of human fibroblasts to penetrate within the pores of 3D composites was verified by a migration assay simulating the devices repopulation upon injection in the injured site.

Keywords: *Bioactive glass; Nanocomposite paste; Magnesium; Bone.*

1. Introduction

Between the class of bone substitute, bioactive glasses (BG) represent a very promising category; thanks to their biocompatibility and bioactivity towards the host tissue, they may be used as porous scaffolds, fillers, dental grafts, coatings, and orthopedic implants [1-3]. BG bioactivity is due to their chemical similarity to the bone tissue; so, BG easily attach to the bone by forming a layer of hydroxycarbonate apatite (HCA) on its surface [1]. This process, which begins with the release of cations and the formation of silica hydrate finishes after the formation of a crystalline hydroxyapatite layer [4], according the following steps:

i) Replacement of Mg^{2+} , Na^+ , and Ca^{2+} cations with H^+ in the medium and hydrolysis of the silica groups and the formation of silanols ($Si-O-Na^+ + H^+ + OH^- \rightarrow Si-OH^+ + Na^+ + OH^-$)

ii) Once the hydroxyl groups amount in the medium increases, the silica network attach and the silica bond is broken ($Si-O-Si + H_2O \rightarrow Si-OH + Si-OH$)

iii) Condensation of Si-OH groups, polymerization of silica-rich layers, and release of alkaline and alkaline earth ions

iv) Migration of Ca^{2+} and PO_4^{3-} ions on the surface of the silica-rich layer and the formation of a $CaO-P_2O_5$ amorphous layer on its surface.

v) Crystallization of a $CaO-P_2O_5$ layer and the formation of crystalline hydroxyapatite carbonate

The precipitated HCA can bond with collagen fibers thus providing an adequate surface for adhesion and proliferation of osteogenic cells, creating a bridge with the bone [1].

Taking onto account the class of injectable pastes for bone small defects repair, it must be considered that BG are often used in combination with other biopolymers to improve mechanical properties. Accordingly, by a combination of degradable polymers such as chitosan, gelatin, sodium alginate or hyaluronic acid with a bioactive glass powder, an appropriate bioactive composite can be formed known as organic-inorganic hybrids [5,6]. The choice of the most appropriate polymer depends by the biocompatibility, the lack of side effects upon implantation (such as inflammation or allergy) and the physical-chemical properties such as degradation and reabsorption [7-11]. In particular, gelatin seems to hold such characteristics as it has been used for many applications including the synthesis of hydrogels, composites hybrids and scaffolds [12,13]. Thanks to the gelatin high affinity with cells, the presence of such polymer can be exploited to improve cells adhesion, spread, differentiation and proteins expression onto materials with low bioactivity or aimed at tissue healing. Accordingly, the use of gelatin as co-polymer can be applied towards injectable bioactive glass-based pastes for small bone defect repair developing composites with high bioactivity and injectability.

However, a predictable limitation of bioactive glass-gelatin composites is due to their sensitivity towards temperature that can cause poor stability upon injection in the body [14,15]. Therefore, to solve this problem, an appropriate cross-linker should be used as third component to improve the stability of the composites. Various materials such as polyethylenglicol, β -glycerophosphate, glutaraldehyde, hydroxyl propyl and methylcellulose, have been used as cross-linker for glass-polymer composites [16-20]. Trimethoxysilane

(3-glycidyloxypropyl) (GPTMS) is a biocompatible cross-linker forming silanol groups by hydrolyzing its methoxy groups leading to a silicate network [21]; due to the presence of such silanol groups, GPTMS can give rise to a strong bond with gelatin [22]. Moreover, the organic-inorganic compound derived from GPTMS cross-link can display ameliorate physical and chemical properties and a higher stability of the mineral and organic components [22].

Another key factor influencing the glass kinetics and the formation of the hydroxyapatite layer is the glass' chemical composition [23-25]. So far, the effect of oxides such as Ag, P, F, ZrO₂, Al, Zn, Sr, and Mg on the biological, physical, and chemical properties of such glasses have been studied [26-33]. In particular, magnesium seems holding the capacity to strongly influence the glass kinetics. The addition of MgO up to 13 mol.% into glass can delay the apatite formation without inhibiting the process as well as an increase of the alkaline phosphate activity (ALP) can be observed thus improving osteoblasts proliferation [34,35]. Besides, Mg ions affects the bone matrix quality as demonstrated by the evidence that its deficiency can causes bone fractures [34,35]. Despite these positive evidences, the role of magnesium influencing the physical-chemical behavior of the glass is still debated in literature. In fact, magnesium has been found to be a modifier element in several studies [36], although others have suggested that magnesium is an intermediate oxide entering the silica network as MgO₄ [27,37]. Therefore, there are contradictory results about the effect of magnesium oxide on the glass properties. Some researchers have shown that MgO delays the apatite layer's mineralization through reducing the mineral layer's thickness [38-40]. Some other studies have shown that the presence of MgO has no effect on the formation of the hydroxyapatite layer, but rather stimulates and promotes the early stages of mineralization [41,42]. Also, it has been found that the substitution of MgO with CaO can improve the stability as well as the mechanical properties of the glass [43].

Dealing with the ions influence on the glass behavior, in a prior work the Authors found that in a glass-gelatin injectable paste composite, an extra amount of Ca²⁺ ions determined the precipitation of calcite instead of apatite, when the injectable paste was tested by immersion in Simulated Body Fluid (SBF) solution [15]. Here, the role of magnesium was investigated in terms of mechanical, chemical and biological properties influence by comparing bulk 64SiO₂-31CaO-5P₂O₅ (mol.%) bioactive glasses and Mg-doped 64SiO₂-27CaO-4MgO-5P₂O₅ (mol.%) glasses aimed at injectable pastes for small bone defects repair. The pastes were investigated for their mechanical properties, injectability and bioactivity, while cytocompatibility was tested towards osteoblasts and fibroblasts in a migration assay. Results suggested that by adding Mg to the bioactive glass the release of Ca²⁺ was controlled thus avoiding the formation of calcite, bringing to a general improvement of the tested mechanical, chemical and bioactive properties.

2. Materials and Methods

2.1 Composites preparation

Two bioactive glasses 64SiO₂-31CaO-5P₂O₅ mol.% (BG) and 64SiO₂-27CaO-4MgO-5P₂O₅ mol.% (BG-Mg) were synthesized through the acid-base sol-gel method as previously described [10]. The precursors triethyl phosphate (TEP), tetraethyl orthosilicate (TEOS), calcium nitrate tetrahydrate powder, magnesium nitrate, ammonia solution, and nitric acid solution were purchased from Merck (Merck KGaA, Darmstadt, Germany) while bovine gelatin (Gel) and (3-Glycidyloxypropyl) trimethoxysilane (GPTMS) were purchased from Sigma (Sigma-Aldrich, St. Louis, USA). To prepare the gelatin solution, 3 wt.% gelatin powder was dissolved in distilled water. To prepare the composite pastes, the bioactive glass powder was mixed with the polymer solution and the GPTMS according to the amount detailed in Table 1, with a powder to liquid ratio (P/L) of 0.25 (referring powder to the glass and liquid to the polymer solution). As a result, two different pastes BG-Mg-Gel and BG-Gel were prepared, with the same powder-to-liquid ratio; finally, 20 wt.% (to the dry weight of polymer powder) GPTMS was added to each of them.

2.2 Physical-chemical characterization

2.2.1 Morphology and composition analysis

Composites' surface area, pore size and pore volume were determined by the Brunauer-Emmett-Teller (BET) analysis using an ASAP 2020 instrument (from Micromeritics Instrument Corporation, GA, USA) at 77K with nitrogen as absorbent. Glass powder phase specifications were measured using an X-ray diffractometry (XRD), with Cu-K α radiation and X-ray wavelength of 1.54050 Å (Dron 8, Bourestovink, Russia), operated at 40 kV and 40 mA at a step size of 0.04 and a count time of 2 s/step. The morphology of the glass particles was studied by scanning electron microscopy (SEM, TEScan, VeGa II, Czech).

2.2.2 Mechanical properties evaluation

Rheology was applied to evaluate pastes' mechanical properties using a MC-R300 rheometer with a plate-plate geometry device (from Anton Paar GmbH, Graz, Austria). The test was carried out at room temperature in an oscillatory mode, whereby 3 g of the paste was placed on the bottom plate of the device 1 mm distant from the top plate. To determine the linear viscoelastic region, the storage (G') and the loss (G'') modules were measured in terms of strain (γ), while the complex viscosity (η^*), G' and G'' were determined in terms of angular frequency (ω).

2.2.3 Pastes injectability and disintegration resistance

To measure the injectability of the pastes, 3 g of each paste was poured into the syringe with a needle diameter of 2 mm, and a compressive force was applied on the syringe's piston under the speed of 30 mm/min, using a computer-controlled mechanical testing machine (SANTAM STM-20). Finally, the compressive force versus displacement was measured, and the percentage of injectability was calculated according to the following equation 1 (Eq.1):

$$(Eq. 1) \text{ Injectability}(\%) = \frac{\text{Extruded paste volume}}{\text{Total paste volume}} \times 100$$

The pastes were immersed in the simulated body fluid (SBF) solution, which was developed based on Kokubo's method [44] and placed at 37 °C in the incubator for 5 min, 24 h, and 48 h, respectively, to determine their disintegration resistance (Dr). Dr was calculated according to the equation 2 (Eq.2) where W_s is the disintegrated paste weight after removing the SBF solution while W_i is the total paste weight before immersion.

$$(Eq. 2) Dr = \frac{W_i - W_s}{W_i} \times 100$$

2.2.4 Pastes bioactivity evaluation

Pastes' bioactivity was evaluated by their ability to form apatite *in vitro*. Accordingly, 1 g of each paste was immersed in 30 ml of the SBF solution at 37 °C in an incubator. The solution was refreshed every 48 h up to 14 days. At each time points, the pastes were taken out from the solution, rinsed with distilled water and air-dried at room temperature. To demonstrate the phase composition of the pastes, before and after immersion in the SBF solution, the dried pastes were analyzed by XRD whereas the morphology and chemical composition of the particles in the pastes, before and after immersion, was visually checked by SEM-EDX. The chemical band groups in the pastes were studied by Fourier Transform Infrared Spectroscopy (FTIR) at a wavelength of 4000-400 cm^{-1} with KBr pellets.

2.3 Biological evaluation

2.3.1 Cytocompatibility

Specimens were UV-sterilized (30 minutes/each side) and molded into 1 cm diameter, 1 cm height disks prior to perform biological assays. Human mature osteoblasts (MG-63, ATCC CRL-1427) were purchased from the American Type Culture Collection (ATCC, Manassas, USA) and used as bone-like model to test pastes cytocompatibility *in vitro*. Cells were cultivated into Dulbecco's Modified Minimal Essential Medium (DMEM, from Sigma-Aldrich) supplemented with 10% fetal bovine serum (FBS) and 1% antibiotics (penicillin/streptomycin) at 37 °C in a 5% CO_2 atmosphere. After reaching 80-90% confluence, cells were detached by trypsin-EDTA solution, harvested and directly drop-wise seeded onto specimens' surface at defined concentration (1×10^4 cells/specimen). Composites were incubated (37°C, 5% CO_2) for 24 and 72 hrs; at each time-points, cells' viability was determined by the metabolic colorimetric alamar blue assay (Alamar Blue™, from Life Technologies, ThermoFisher, USA) following Manufacturer's instructions. Briefly, supernatants were removed from each well containing cells and replaced with the alamar blue solution (10% v/v in fresh medium). Plates were incubated in the dark for 4 hours and then 100 μl were removed, spotted into a new black 96-well plate and fluorescence signals were evaluated with a spectrophotometer (Spark,

1 Tecan Trading AG, CH) using the following set-up: fluorescence excitation wavelength 570 nm, fluorescence
2 emission reading 590 nm. Finally, cells morphology was visually evaluated by SEM observations after 72 hrs
3 cultivation. Briefly, samples were washed twice with phosphate-buffered saline (PBS), fixed for 12 hrs by 4%
4 formaldehyde and dehydrated by the alcohol scale (50%, 70%, 90%, and 100%, 2 hrs each). Subsequently,
5 dehydrated specimens were surface covered with conductive silver and examined by SEM imaging.
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10 *2.3.2 Migration assay*

11 To evaluate the pastes' ability to support new tissue ingrowth within pores simulating the self-healing
12 process, a migration assay was performed using human primary fibroblasts (HGF, ATCC PCS-201-018).
13 Accordingly, 8 mm height - 4 mm width cylinders were realized and sterilized by UV light. Cells were
14 purchased from ATCC and cultivated into alpha-modified DMEM (α -MEM, from Sigma) supplemented with
15 10% FBS and 1% antibiotics. Once cells reached 80-90% confluence, they were detached by trypsin-EDTA a
16 seeded onto specimens' top side embedded into 20 μ l of a collagen matrix (PureCol™ EZ Gel solution, from
17 Sigma) used as a temporary substrate. After collagen gelation (2 hrs, 37 °C) the specimens were rinsed with
18 fresh medium (1 ml each) and cultivated for 15 days with medium changed every 3 days. Finally, specimens
19 were collected, vertically cut into 2 halves to visualize the internal part, dehydrated by the alcohol scale,
20 covered with Cr and observed with a JEOL NEOSCOPE JCM 6000 PLUS SEM (from Nikon Instrument S.p.A.,
21 Firenze, Italy). Images of the internal part were used to confirm cells' migration from the top side within the
22 pores.
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34 **2.4 Statistical analysis of data**

35 Biological experiments were performed using 6 replicates. Samples were statistically compared by the SPSS
36 software (v25, IBM) using the one-way ANOVA test after check of normal distribution and homogeneity using
37 Shapiro-Wilk's and Levene's test respectively. Tukey's test was used for post-hoc analysis. Differences were
38 considered as significant for $p < 0.05$.
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46 **3. Results and Discussion**

47 **3.1 Physical-chemical characterization**

48 *3.1.1 Composites morphology and structural characterization*

49 The morphology, absorption-desorption curves, and X-ray diffraction patterns of the samples' BG and BG-Mg
50 were investigated firstly (results are summarized in Table 2). The average particle size of BG and BG-Mg
51 glasses resulted of approximately 50 nm and 100 nm, respectively. The hysteresis BET loops, according to the
52 IUPAC classification, indicated that the two glasses were mesoporous. Based on the BJH adsorption –
53 desorption traces, the pore volume, the specific surface area, and the size of average pores of BG and BG-
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1 Mg resulted of 0.602 and 0.634 (cm³/g), 123.139 and 109.104 (m²/g), and 18.217 and 23.036 (nm),
2 respectively. The XRD results represented that the samples were amorphous.
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5 3.1.2 Composites injectability and disintegration resistance 6

7 Pastes injectability was studied under computer-controlled conditions by applying a compressive force
8 towards 2 mm needle syringe lodging the composites. Accordingly, the injected volumes % and the relative
9 applied forces required to complete the pastes extrusion are summarized in Table 3. Apparently, the entering
10 of 4 mol.% Mg ions into the glass composition has not influenced the injectability at all. So, it can be
11 speculated that the flow ability of the pastes is more dependent on the gelatin solution rather than to the
12 glass composition [15].
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17 After evaluating the role of Mg in the pastes injectability properties, the disintegration resistance (Dr) was
18 selected as the second parameter to evaluate Mg contribute. Accordingly, pastes were immersed in SBF for
19 5, 24 and 48 hours prior to evaluate their Dr in comparison to the starting values as prior described in the
20 Equation 1. Results are reported in Figure 1a showing that the resistance of the BG-Gel-GPT20 decreased
21 more rapidly than the Mg-doped ones starting from 24 hrs immersion (97±2% vs 95±2%) and being more
22 evident after 48 hrs (94±3% vs 89±3%). However, both pastes showed acceptable resistance and preserved
23 their shapes after 48 h immersion. The difference between the two samples was attributed to the presence
24 of magnesium in the BG-Mg-Gel-GPT20 structure. In fact, the pastes' Dr depends on two main factors: i) the
25 cross link between the polymer chains due to the ions released from the glass (schematized in Fig. 1b) and ii)
26 the bridging between the polymer and the glass (schematized in Fig. 1c). So, the glass particles can link the
27 polymer molecules through i) O⁻¹_{gelatin}-M²⁺-O⁻¹_{gelatin} and/or ii) O⁻¹_{gelatin}-M²⁺_{glass}-O⁻¹_{gelatin} bonds and increase the
28 disintegration resistance of the paste (M²⁺ refers to a two-valence cation like Ca²⁺ and/or Mg²⁺). Magnesium
29 plays a significant role in stabilizing the glass structure [34] and ameliorating the bond strength of the
30 mentioned links. A glass forms a more strong and stable three-dimensional network when the ionic field
31 strength of its cation increases, i.e. $z \cdot e / r^2$, where z, e, and r are the electrical valence of the cation, the
32 electrical charge of the electron, and the ionic radius of the used modifier cation, respectively. The
33 arrangement of ionic field strength for alkaline earth elements is Mg > Ca > Sr > Ba. It was also noted that the
34 addition of MgO instead of CaO into the glass led to increase the network coherence attributed to the small
35 size and larger field strength [45]. Thus, it can be concluded that the paste will be more resistant against
36 disintegration when its glass constituent contains magnesium oxide. The reduction in the amount of calcite
37 that is formed in the Mg-free sample seems to be another parameter that increases the disintegration
38 resistance of BG-Mg-Gel-GPT20. Apparently, with the formation of calcium carbonate, the concentration of
39 calcium ions in the solution is reduced, leading to a decrease in its probability for participation of the cross-
40 linker in the paste.
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3.1.3 Rheological characterization

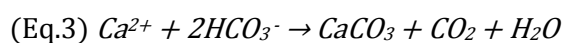
Another key element of the injectable pastes is represented by their mechanical strength. As a temporary substrate for bone repair, it is very important that the pastes can support the imposed compressive and shear forces meanwhile the self-healing occurs forming the new bone. Accordingly, here the GPTMS was used as cross-linker to improve the stability and thus the mechanical strength of the gelatin-bioactive glass composites. In fact, as schematized in Figure 2a, gelatin can bind the GPTMS silanol groups to form a large proteins chain. Moreover, it can be considered that hydrogen, ionic and covalent bonds can be formed between the bioactive glass and gelatin, through i) the bond between silanol in the glass-amine and gelatin hydroxyl groups, ii) calcium/magnesium in the glass-carboxyl groups of gelatin, and iii) calcium/magnesium in the glass-oxygen in the gelatin. Taken into account such interactions, the rheological behavior of the injectable pastes was studied to investigate the effect of magnesium addition on this regard.

In Figure 2b the changes in the storage and loss modulus of the samples in different angular frequencies, determined at a constant strain of 0.3% were shown. The storage modules of both samples are higher than the loss modules in the whole frequency ranges. Conversely, a non-homogeneous pattern was found in the sample lacking a cross linker in a previous study from the Authors [15]. This behavior can be attributed once again to the presence of the cross-linker which contributes creating a coherent and integrated structure, as was previously observed in the disintegration resistance evaluation. The presence of Mg is expected to increase the strength of the glass by creating stronger Mg-O-Si bonds than that of the Ca-O-Si; therefore, the storage modules of the Mg bearing paste should be higher than the Mg-free one due to the material's elastic nature. The results of the rheology test, which are given in Table 4, seem to confirm these hypotheses.

Finally, in Figure 2c the reduction in the complex viscosity of the two samples by increasing the angular frequency, at a constant strain of 0.3%, suggested for a shear-thinning behavior of the pastes. The complex viscosity of the pastes depends on the paste's modules. Therefore, it can be speculated that the MgO-containing paste has a higher viscosity than the MgO-free one, which originates from their higher bond strength as noted before.

3.1.4 Bioactivity evaluation

The Mg-induced bioactivity was evaluated as the ability of the injectable pastes to form apatite instead of calcite as previously observed for calcium [15]. Accordingly, the XRD patterns before (0 day) and after immersion in the SBF of the two pastes are given in Figure 3a after 1 and 14 days. As can be observed, negligible peaks of calcite can be observed before and after the immersion of the Mg containing sample in the SBF (BG-Mg-Gel-GPT20), while calcite (reference code 00-001-837) is the main crystalline phase precipitated in the Mg-free sample, due to the higher amounts of calcium oxide in its glass phase. The reaction between Ca^{2+} ions and the CO_3^{2-} deriving from the polymers occurring in the SBF environment causes calcite precipitation as described in the Equation 3 (Eq.3):



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3 Similar to the results here obtained, Seah et al. [46] shown that in the 45S5 glasses calcite is formed in a
4 highly concentrated homogeneous phase in the early days after immersion in SBF; however, it decomposes
5 gradually with time releasing calcium ions into the SBF solution and thus participating in the formation of
6 calcium phosphate. The highest intensity peak of apatite was observed for both pastes after 14 days of
7 immersion; however, apatite was found to precipitate later in the Mg-containing paste in comparison to the
8 bulk materials. In a previous study, K.M. Tohamy et al. [47] evaluated the influence of MgO replacing SiO₂
9 into SiO₂-Na₂O-CaO-P₂O₅-MgO glasses demonstrating that apatite layer formation decreased with
10 increasing amount of MgO due to the formation of diopside. Similarly, in another study [48] CaP layers were
11 found to be deposited on the lower MgO (8 to 2 mol%) content glass surface after 28 days of SBF study.

12 The FTIR spectra of the pastes, before immersion, i.e. zero-day, and after immersion, i.e. 14 days, in the SBF
13 solution are shown in Figure 3b. The chemical groups associated with the formation of apatite are indicated
14 in the spectra. The absorption bands at 460 and 1050 cm⁻¹ are related to the Si-O-Si vibrations, while the
15 chemical bond between glass Ca²⁺ and the carboxyl group from gelatin was detected at 1460 cm⁻¹ in
16 accordance with previous literature [15]. The hydrolysis observed by the bands at 1100 cm⁻¹ and 940 cm⁻¹
17 from the oxirane groups, as well as the band at 1254 cm⁻¹ from the GPTMS epoxy groups [49] caused the
18 formation of the Si-O-Si and Si-OH groups. These peaks overlap with the Si-O-Si from the bioactive glass and
19 represent the formation of a bond between the epoxy groups of GPTMS and the amino groups from gelatin
20 [50]. The bands marked in 560, 603, and 950 cm⁻¹ indicated the presence of ν₄-(PO₄)³⁻ thus giving
21 confirmation of the apatite formation [50]. The absorption bands at 1630 and 3700-3000 cm⁻¹ interval
22 showed the presence of OH groups. Also, the 870, 1450 and 1510 cm⁻¹ bands related to the CO₃²⁻ groups
23 involved in the formation of type B carbonated apatite via replacing of phosphorus groups [10,50] were
24 observed. In general, the intensities coming from the Mg-free paste peaks resulted as higher, thus suggesting
25 that it is probably more bioactive due to the higher specific surface area of the Mg-free sample, and its lower
26 bond strength. As a confirmation, the bond around 711 cm⁻¹ related to calcite [51] was not found after 14
27 days for the BG-Gel-GPT20 specimens, thus confirming the results of the XRD characterization.

28 Finally, SEM-EDX images (Fig. 3c-d and Fig. 4) showed the morphology of the particles in the pastes, before
29 and after the immersion in the SBF. Before immersion, the glass particles seem to be interconnected and
30 bonded together by the polymers and the GPTMS. In general, regardless of the sample, the morphology of
31 the particles is completely different before and after the immersion, showing a dense layer of spherical
32 aggregates, ascribable to the *in vitro* formed hydroxyapatite. The EDX spectra shown in Figure 4 reported an
33 increase in the intensity of Ca and P elements that can be an indication of the formation of a calcium
34 phosphate-based phase on the surfaces of the pastes. Based on the XRD and FTIR results, the precipitated
35 calcium phosphate can be claimed to be apatite. By replacing some Ca with Mg, apatite precipitates were
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1 observed early on the surface of both composites after immersion in the SBF solution. Previous literature
2 demonstrated that by substituting 5 mol.% of magnesium instead of calcium in 58S glass, the rates of apatite
3 formation, cell proliferation, and cell differentiation were increased [52]. In another study, it was found that
4 by adding 1 wt.% of MgO to the glass structure, a strong chemical bond with the bone was formed probably
5 due to an increased bioactivity [53]. In conclusion, glasses' bioactivity can be influenced by the rate of
6 dissolution and precipitation of the apatite layer, which is controlled by the composition and structure of the
7 glass.
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14 **3.2 Biological characterization**

15 *3.2.1 Cytocompatibility*

16 As debated, the here presented injectable pastes are aimed at small bone defects repair by direct injection
17 into the injured site. So, the specimens' cytocompatibility was tested towards osteoblasts (MG-63) as cells
18 representative for the bone tissue. To exclude any potential toxic effect due to the chemical composition or
19 surface properties, cells were cultivated in direct contact with pastes. Polystyrene was used as control to
20 compare test specimens' results as it represents the gold standard substrate for cells *in vitro* cultivation.
21 Results are reported in Figure 5.
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28 As can be seen, at both time-points the metabolic activity resulted as significantly higher (Fig. 5a, $p < 0.05$,
29 indicated by *) for the cells cultivated in contact with pastes in comparison with polystyrene (poly) control,
30 thus confirming the strong bioactive glasses' bioactivity even in the absence of biochemical stimulation such
31 as ALP [54]. Moreover, by comparing the test groups, on day 7 the viability of the Mg-containing paste seeded
32 cells was significantly higher than that of the Mg-free paste ($p < 0.05$, indicated by #). Based on the previous
33 studies, Mg can favor cell adhesion, new bone formation, and metabolic activity of osteoblasts [15,47]. Also,
34 literature showed that the proliferation and viability of osteoblasts as well as their ALP activity was superior
35 if stimulated by Mg in comparison with cells cultivated with bulk medium [49]. Varanasi et al. [51] shown that
36 the expression of collagen I, ALP, runx2, and osteocalcin increased by the dissolution of Mg-containing glass
37 in cell culture [51]. Finally, SEM images (Fig. 5b) representative for 7 days direct cultivation showed that the
38 cells are well attached and spread onto the surface displaying the typical stellate-like morphology of the
39 osteoblasts.
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51 *3.2.2 Cells Migration*

52 After evaluating pastes cytocompatibility towards osteoblasts cells cultivated directly onto top surface, a
53 migration assay using fibroblasts was assessed. The aim of such evaluation is to simulate cells migration from
54 the surrounding healthy tissue to the injured site filled by the paste: in this scenario, migrating cells must be
55 able to move within the pores of the injected paste in order to fill it and promote the self-healing process
56 while the composite acts as a temporary substrate. Accordingly, cells were seeded on the top of the
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specimens and cultivated for 15 days; afterwards, specimens were vertically cut in two halves to analyze by SEM imaging the internal pores looking for cells migrated from the top layer (as schematized in Figure 6a). Results are reported in Figure 6.

As can be observed in the SEM images (Fig. 6b) fibroblasts were detected within the internal pores of the pastes, thus confirming that they were migrated from the top surface (where they were seeded) to the inner part of the cylindrical devices. The cytoskeleton elongation suggested that cells were able to adapt to the pastes' topography and chemistry, thus allowing for the pores filling representing a crucial step to support the self-healing process.

4. Conclusions

Obtained results suggested that the Ca ions substitution by the Mg ones can improve several properties of the bioactive glasses-based gelatin injectable pastes thanks to the higher ionic field strength of the former ion. So, despite a similar injectability, bioactive glasses doped with Mg showed better mechanical properties and higher bioactivity thanks to the possibility to control the release of Ca^{2+} thus reducing the formation of calcite in favor of the bone precursor apatite. Finally, pastes resulted as cytocompatible for osteoblasts as well as they were penetrated by migrating fibroblasts simulating the materials repopulation after implantation in the injured site. Therefore, the Mg-doped bioactive glasses appear as a promising tool for the treatment of small bone defects by direct injection.

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16 Figure Legends

17
18 **Figure 1.** The disintegration resistance (Dr) test in SBF immersion demonstrated a higher resistance to
19 degradation from the Mg-doped composites (BG-Mg-Gel-GPT20) in comparison to the bulk ones (BG-Gel-
20 GPT20) (a). This is due to the superior structural stability conferred by the Mg ions linking the polymer chains
21 (b) that ameliorated the bond between the bioactive glass (BG) and the gelatin (Gel) as schematized in (c).

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27 **Figure 2.** Rheology. The use of GPTMS as cross linker allowed for bond with gelatin chains by the silanol
28 groups (a). The presence of Mg (BG-Mg-Gel-GPT20) improved the storage and loss modules (b) as well as the
29 complex viscosity (c) in comparison with the bulk BG-Gel-GPT20.

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35 **Figure 3.** Bioactivity. The XRD patterns (a) showed the presence of apatite after immersion in SBF for both
36 pastes as confirmed by the FTIR spectra too (b). SEM images of BG-Gel-GPT20 (c) and BG-Mg-Gel-GPT20 (d)
37 provided a visual proof of the apatite crystals formation.

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42 **Figure 4.** EDX spectra of control (a-b, BG-Gel-GPT20) and Mg-doped specimens (c-d, BG-Mg-Gel-GPT20)
43 before and after immersion in SBF.

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47 **Figure 5.** Cytocompatibility. Pastes demonstrated to be bioactive by significantly improving osteoblasts
48 metabolism in comparison with polystyrene (poly) control (a, $p < 0.05$ indicated by *); moreover, Mg (BG-Mg-
49 Gel-GPT20) led to a higher metabolic activity after 7 days in comparison to the BG-Gel-GPT20 bulk material
50 ($p < 0.05$, indicated by #). SEM images confirmed cells' adhesion and spread displaying the typical osteoblast
51 stellate morphology.

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57 **Figure 6.** Migration assay. By analyzing pastes internal pores using SEM imaging (schematized in a by red
58 circles) it was possible to observe cells migrating from the top to the internal part of the devices adhering
59 and spreading across pores (b) after 15 days cultivation.

<i>Composite</i>	<i>BG (g)</i>	<i>Gel (g)</i>	<i>GPTMS (wt.%)</i>
BG-Mg-Gel-GPT20	0.25	1	20
BG-Gel-GPT20	0.25	1	20

Table 1. Bioactive glass (BG) - gelatin (Gel) - trimethoxysilane 3-glycidyoxypropyl) (GPTMS) composites.

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<i>Composite</i>	<i>BET</i>		<i>XRD</i>	<i>SEM</i>	
BG-Gel-GPT20	pore volume		amorphous	particle size	50 (nm)
	specific surface	0.602 (cm ³ /g)			
	area	123.139(m ² /g)			
	size of average pores	18.217 (nm)			
BG-Mg-Gel-GPT20	pore volume		amorphous	particle size	100 (nm)
	specific surface	0.634 (cm ³ /g)			
	area	109.104(m ² /g)			
	size of average pores	23.036 (nm)			

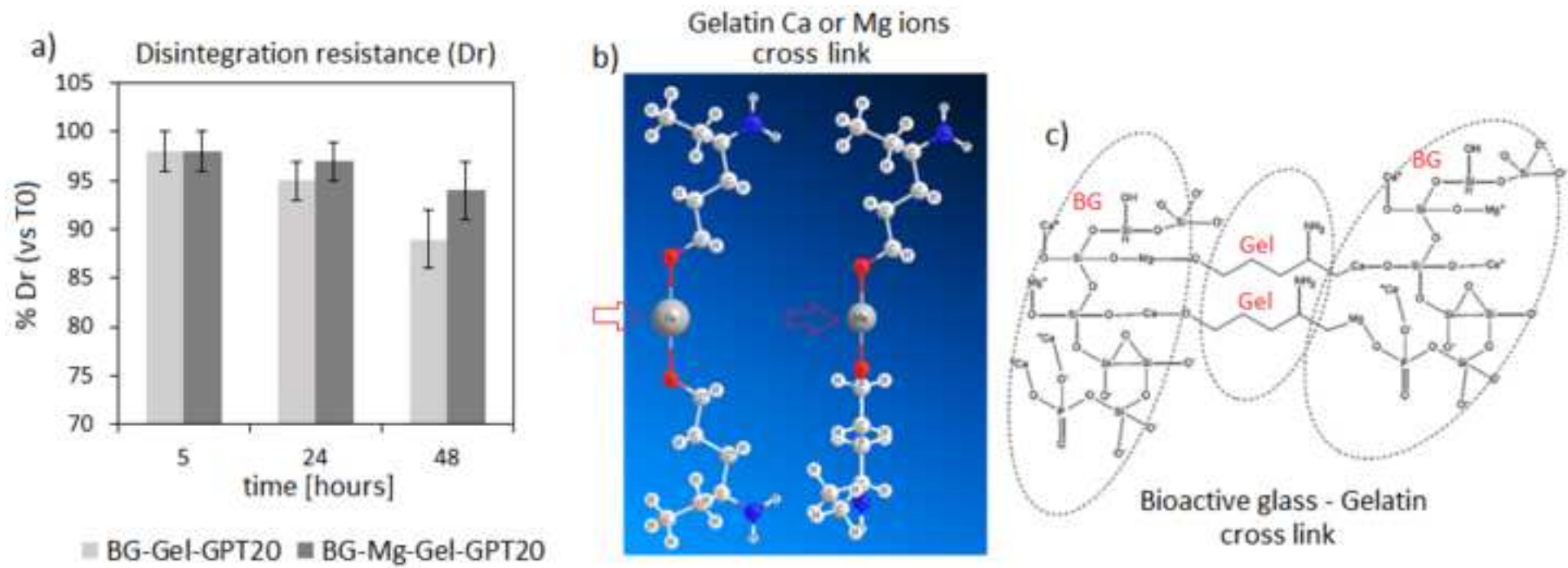
Table 2. Composites physical-chemical characterization summary.

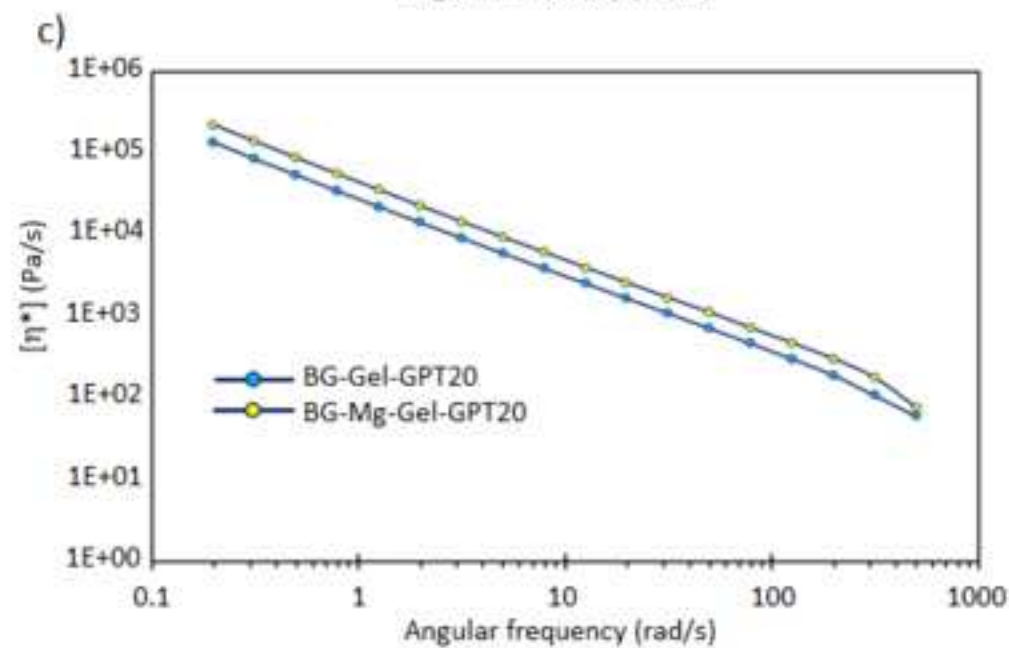
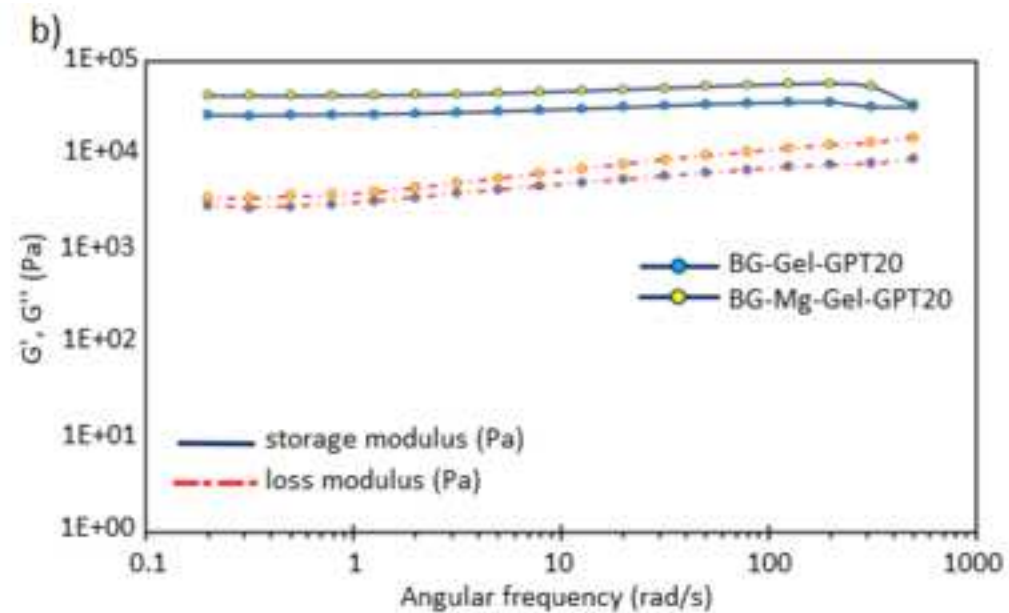
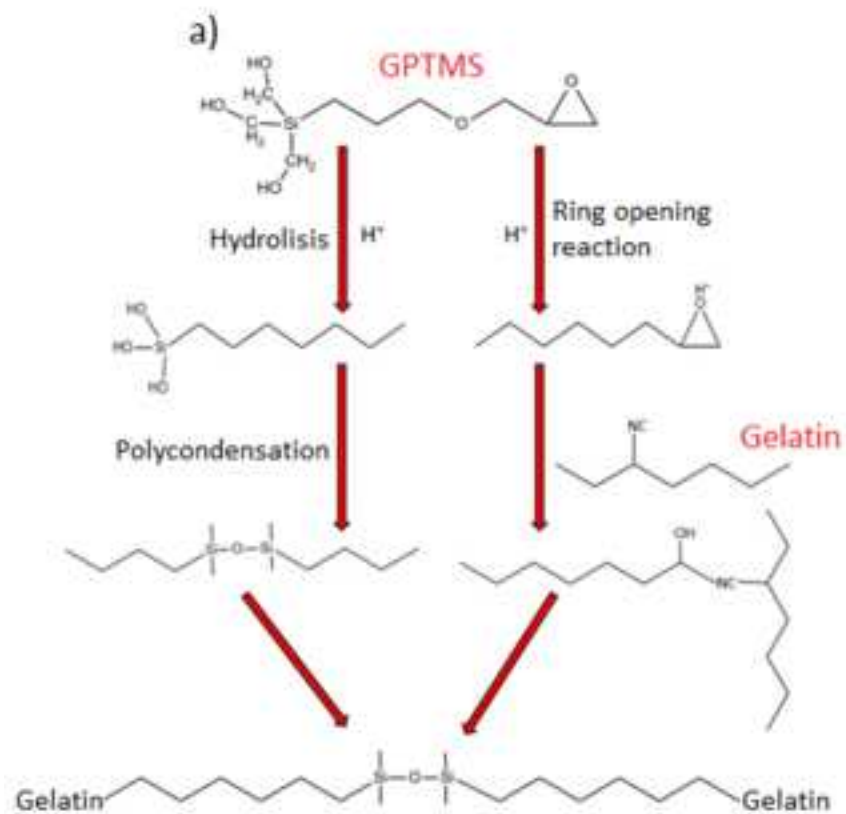
<i>Composite</i>	<i>Injection force (N)</i>	<i>Injectability (%)</i>
BG-Gel-GPT20	5±2	97±1
BG-Mg-Gel-GPT20	5±3	98±2

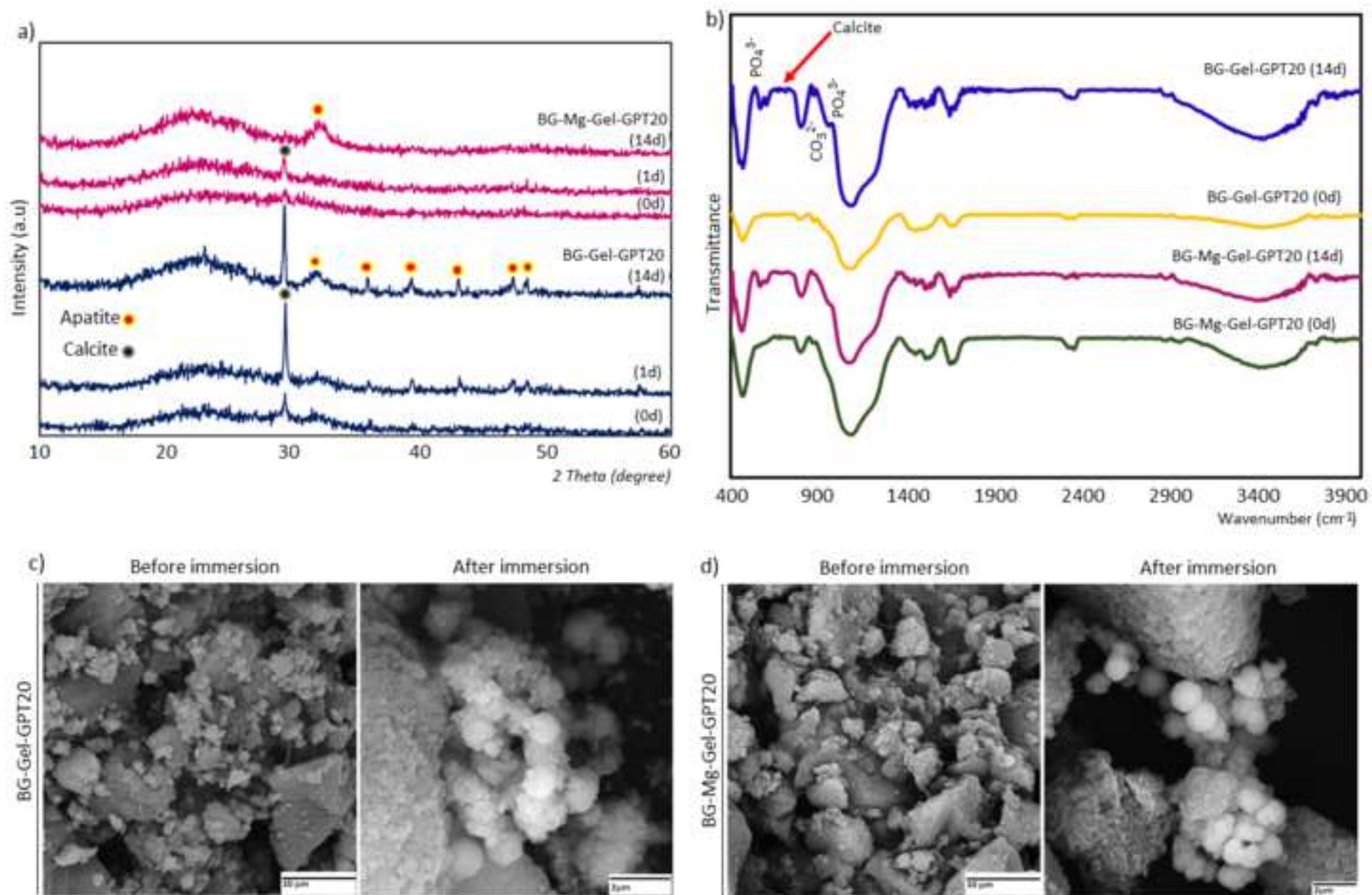
Table 3. Applied force and injection percentage (2 mm diameter needle). Results are expressed as means ± standard deviations of 3 replicates.

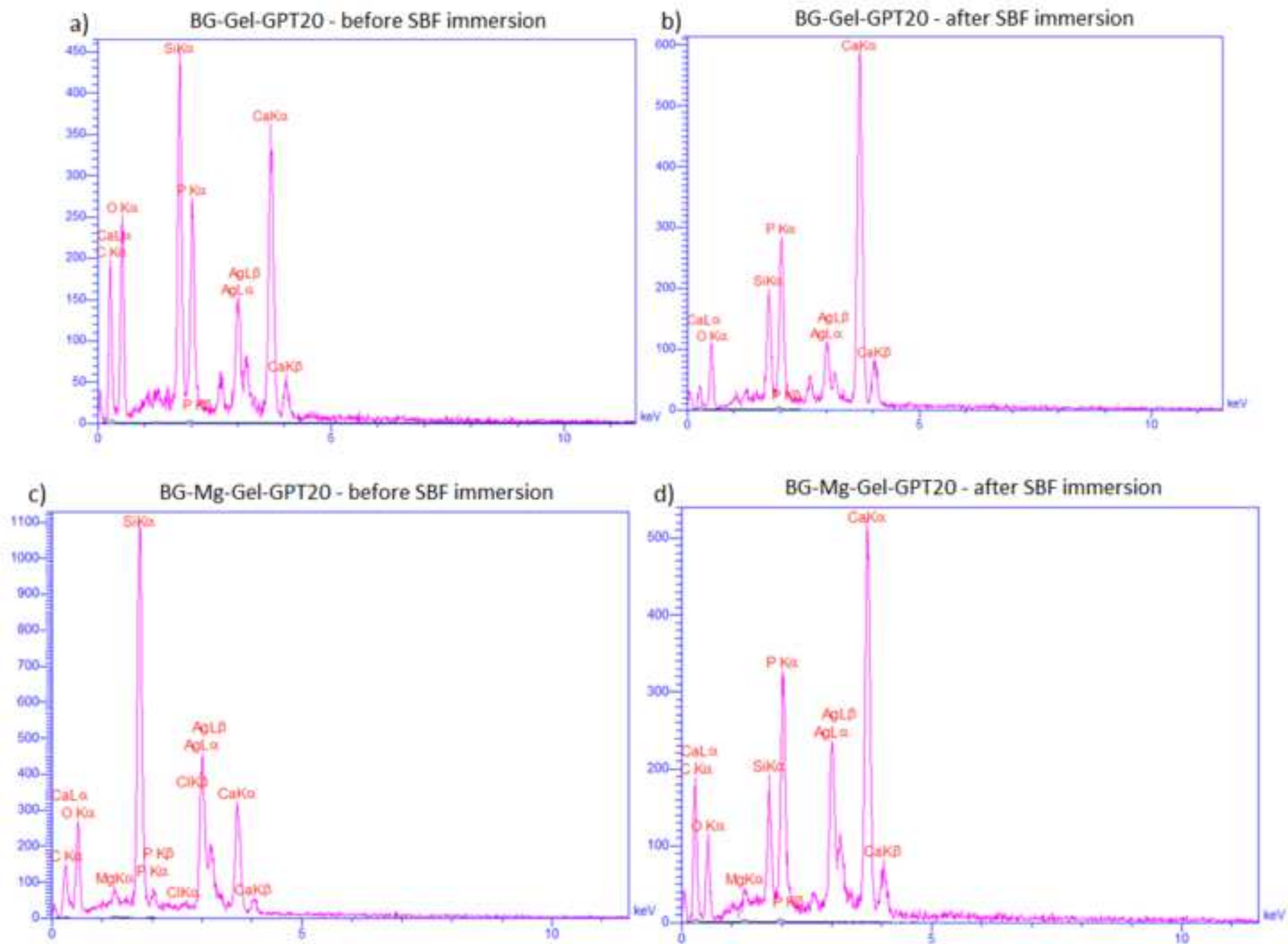
<i>Composite</i>	$G'_{(Pa)}$	$G''_{(Pa)}$	$\eta^*_{(max)} (Pa/s)$
BG-Gel-GPT20	268×10^2	2920	136×10^3
BG-Mg-Gel-GPT20	434×10^2	3610	219×10^3

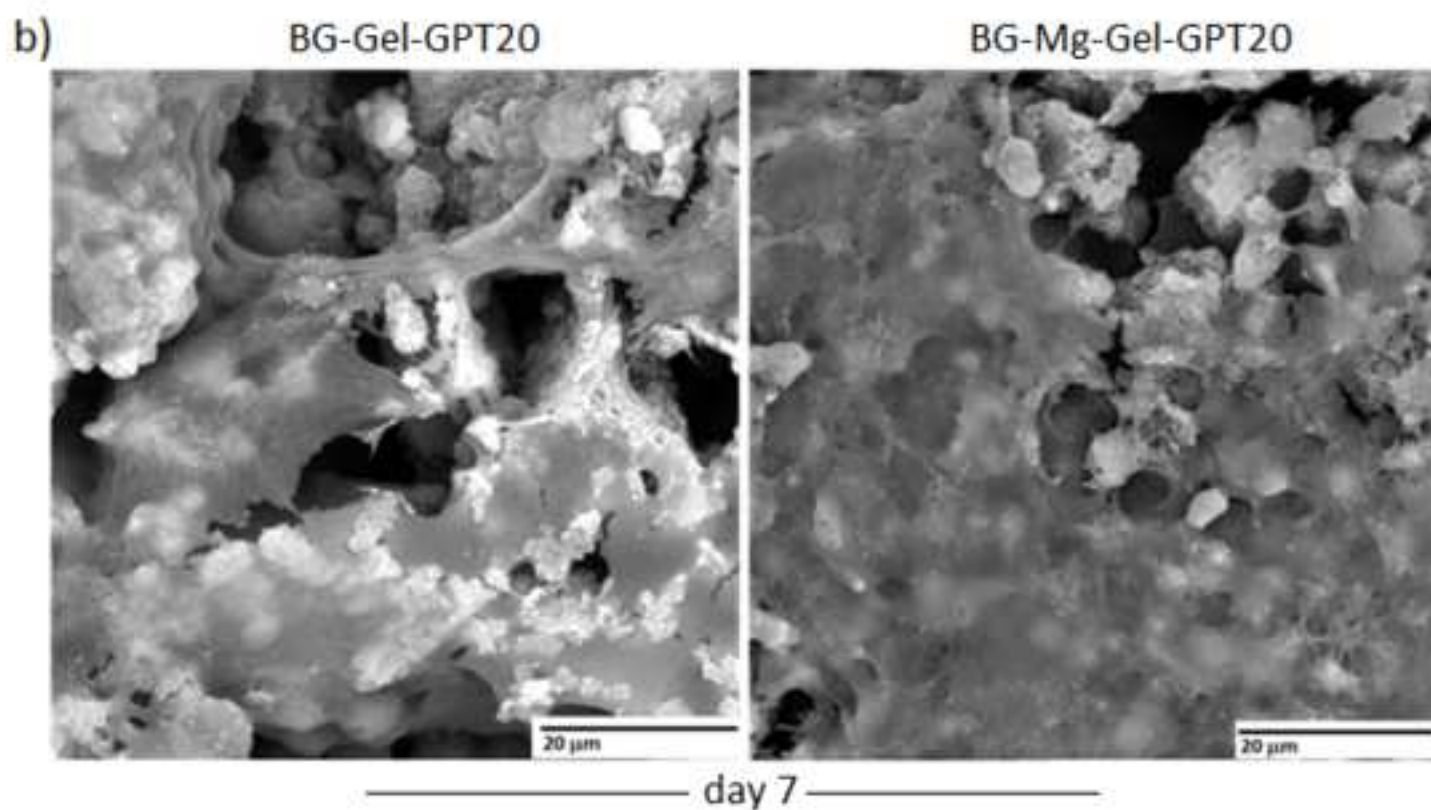
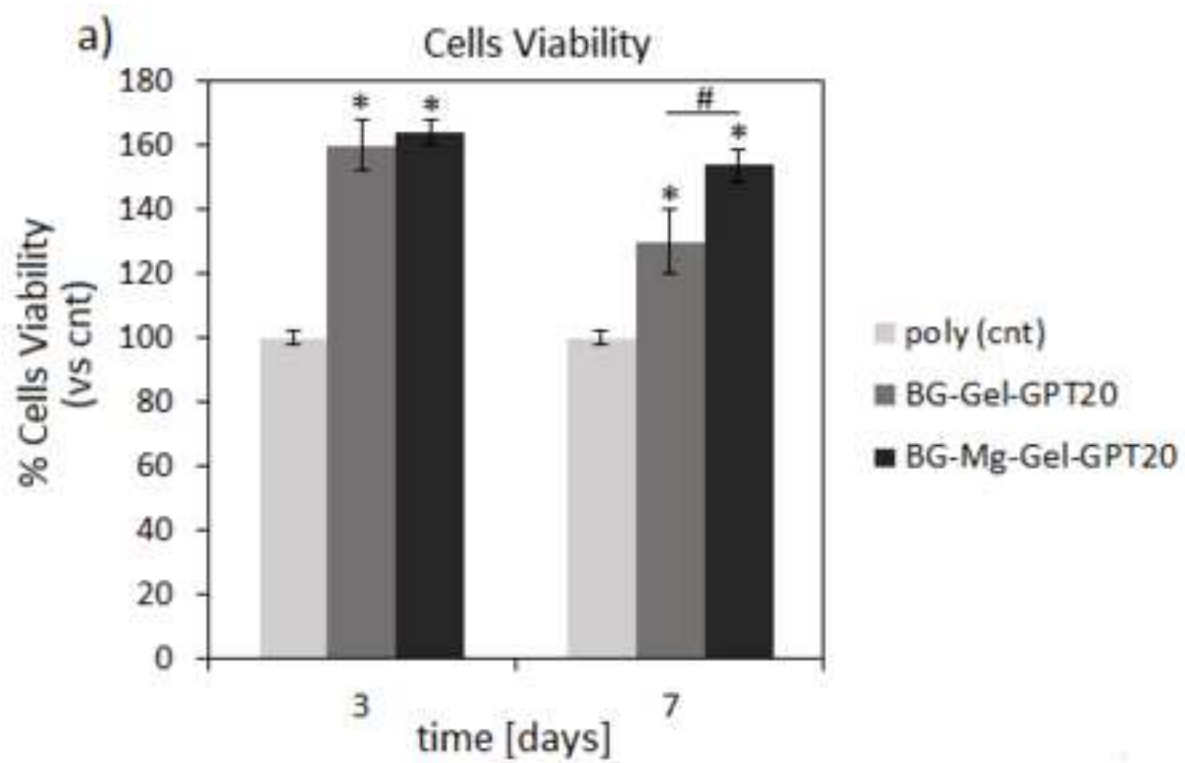
Table 4. Storage module (G'), loss module (G''), and viscosity (η^*) of composites.

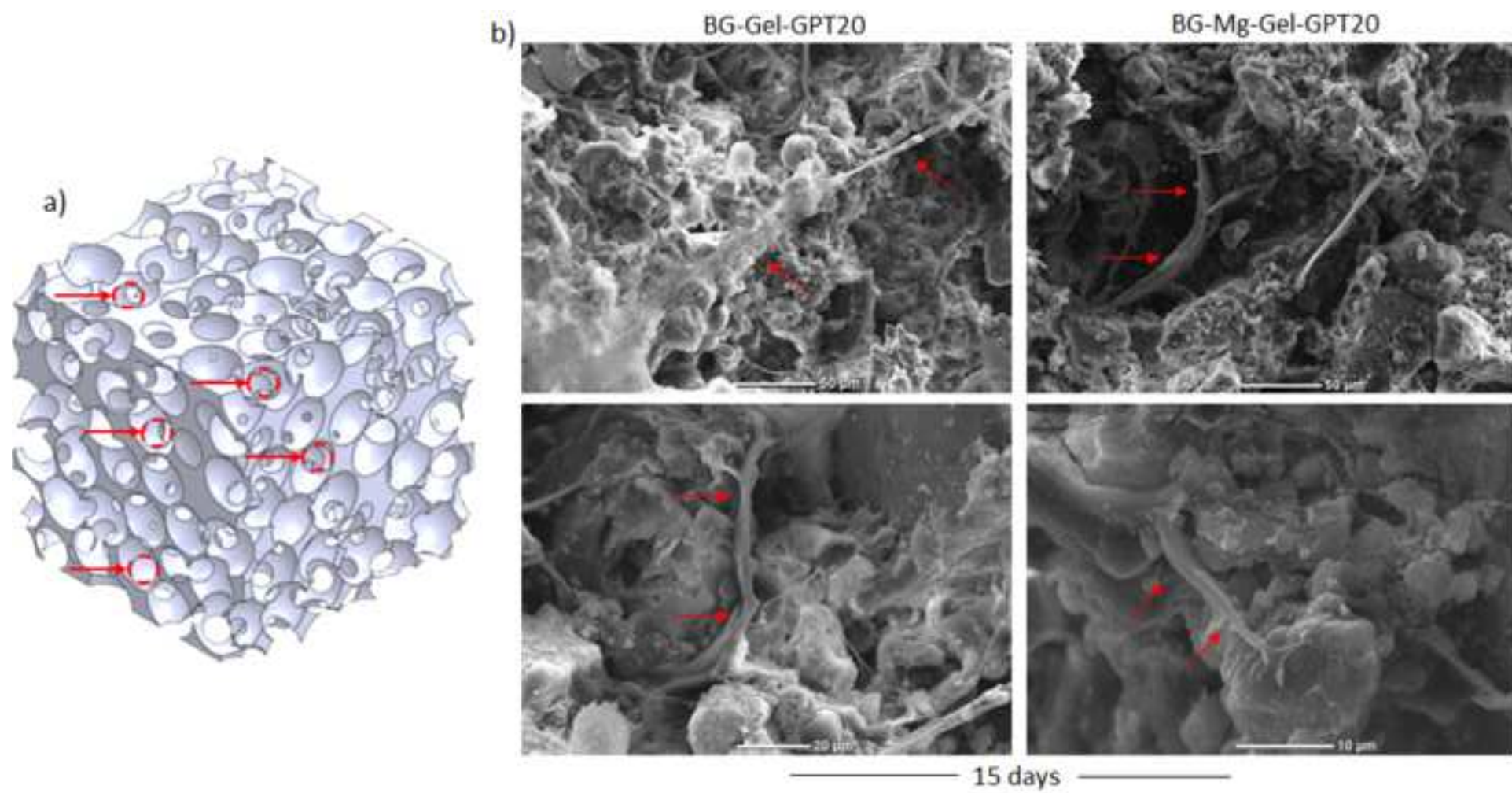














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Conflict of Interest Declaration:

Regarding the submission of the original article entitled "*The effect of Magnesium on bioactivity, rheology and biology behaviors of injectable bioactive glass-gelatin-3-glycidyoxypropyl trimethoxysilane nanocomposite-paste for small bone defects repair*" by Mehri Sohrabi, Bijan Eftekhari Yekta, Hamidreza Rezaie, Mohammad Reza Naimi-Jamal, Ajay Kumar, Andrea Cochis, Marta Miola and Lia Rimondini,

We confirm that the manuscript is original, that it has been approved by all the Authors and that it is not actually under the evaluation of any other Journal.

We confirm that the Authors have no conflicts of interest.

Sincerely,
the corresponding Authors,

Prof. Bijan Eftekhari Yekta

A handwritten signature in black ink that reads 'B. E. Yekta'.

Prof. Lia Rimondini

A handwritten signature in black ink that reads 'Lia Rimondini'.