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Modelling the relationship between tensile strength and porosity in bioceramic scaffolds

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Abstract

A model describing the relationship between tensile strength and total porosity in brittle open-cell macroporous foams is developed and applied to silicate ceramic scaffolds produced by sponge replication and subsequent sinter-crystallization. The tensile strength of the scaffolds decreased from 7.4 to 2.3 MPa as the total porosity increased from 0.40 to 0.79. The results of the model, which is based on the concepts of fracture mechanics, were in good agreement with the experimental data (R^2 = 0.88), which supports the good predictive capability of the approach presented. In principle, this model could help biomaterials scientists not only to estimate the tensile strength of highly-porous bioactive glass and ceramic scaffolds, which is often difficult to determine experimentally, but also to improve the rational design of porous bioceramics with customized properties.

Keywords: Bioceramics; Scaffold; Porosity; Mechanical properties.

1. INTRODUCTION

Implantable bioceramics for bone repair and regeneration are often produced in the form of scaffolds, i.e. 3D porous templates that direct and support the growth of new healthy tissue [1,2]. The presence of open and interconnected large macropores (> 100 μm) is vital to allow cell migration and spreading, as well as blood vessel growth (vascularization) inside the scaffold and, accordingly, in the newly-formed tissue [3,4]. If, on one hand, porosity is highly beneficial from a biological viewpoint, on the other hand it can decrease – even dramatically – the mechanical performance of the scaffold. This trend is common to all porous materials: in general, higher the scaffold porosity, lower the elastic modulus and mechanical strengths under compressive, tensile and flexural loads [5].

At present, the common criterion to assess the mechanical suitability of a porous scaffold for bone tissue engineering applications is based only on the compressive strength, which is relatively easy to assess in cellular ceramics: if the scaffold compressive strength is in the range of that of cancellous bone (2-12 MPa [6]), then it is considered potentially acceptable from a mechanical viewpoint. However, full understanding of the mechanical response of scaffolds to different loading conditions is very important considering that, once implanted *in vivo*, they are subjected to complex multiaxial solicitations.

Studies dealing with the assessment of tensile strength of bioceramic scaffolds are almost totally missing in the literature. Rehorek et al. [7] investigated the mechanical behaviour of 45S5 Bioglass®-derived glass-ceramic foams under tensile loading and showed that the deposition of a polymeric coating on the scaffold struts can increase the fracture strength. Chen et al. [8] determined the tensile strength of wollastonite-containing glass-ceramic scaffolds and used these foams to manufacture multilayer coatings on alumina for use in a novel type of hip joint prosthesis. In general, gaining reliable information on the tensile behaviour of cellular ceramics is challenging mainly due to

technological reasons. Indeed, the brittleness of porous ceramics brings some complications in fixating the specimen in the test machine since standard fixtures cannot be used [9]. In this regard, the only feasible option is to fix the scaffolds by applying carefully-selected adhesives or resins.

Finite element methods (FEMs) have been sometimes proposed to study the mechanical behaviour of bone tissue engineering scaffolds [10,11]. However FEM approaches, to be accurate, need to be combined with advanced imaging techniques (e.g. X-ray micro-computed tomography [12,13]) in order to have a precise geometrical reconstruction of the scaffold.

Information on the tensile strength of porous ceramics is needed for a complete interpretation of their mechanical behaviour and to validate models predicting their response in given biomedical applications. In the attempt to tackle this challenge and overcome the above-mentioned limitations, the present work proposes the use of a new model relating tensile strength, which is often difficult to determine experimentally for highly-porous ceramics, to total porosity, which is relatively easy to assess in ceramic scaffolds.

2. MATERIALS AND METHODS

2.1 Preparation of materials and scaffolds

The starting material used to fabricate the porous bioceramic scaffolds investigated in this work was a silicate glass with the molar composition $50SiO_2$-$6P_2O_5$-$2B_2O_3$-$35CaO$-$7Na_2O$. This glass, which was previously designed for biomedical applications in orthopaedics [14,15], was produced by using a standard melting procedure in a platinum crucible. The raw precursors (SiO_2, $Ca_3(PO_4)_2$, H_3BO_3, $CaCO_3$ and Na_2CO_3 analytical grade powders purchased from Sigma-Aldrich) were homogeneously mixed in a polyethylene bottle by using a roll mixer for 1 h. Then, the powder blend was introduced in

the crucible and melted at 1450 °C for 0.5 h in air inside an electrically-heated furnace. The melt was quenched in distilled water to produce a frit that was ball milled (Pulverisette 0, Frtisch, Germany) and sieved below 32 μm by a stainless steel sieve (Giuliani Technologies Srl, Torino, Italy). According to the authors' previous experience on other glass compositions, this particle size range is very suitable to produce porous glass-derived scaffolds by foam replication [16].

Macroporous cuboids (size 10 mm) of a commercial open-cell polyurethane sponge (45 ppi) were used as sacrificial templates during scaffold production. These small polymeric blocks were dipped into a water-based glass suspension and then compressed to squeeze the slurry out of the sponge pores. Poly(vinyl alcohol) was used as a binder so that the glass particles could more efficiently and homogeneously adhere onto the polymer walls, as suggested elsewhere [17]. Specifically, the binder was dissolved in water under magnetic stirring (200 r.p.m.) at 80 °C for 1 h; then, the glass powder was added to the batch that was stirred again for 20 min prior to the immersion of the sponge. Three sample batches were produced using different solid loads, i.e. 30, 35 and 40 wt.% of glass particles; the binder amount was fixed at 6 wt.% of the total slurry in all cases. The glass-coated foams were dried overnight at room temperature in air and then thermally treated at 900 °C for 3 h in an electrically-heated furnace (heating rate 5 °C/min) to burn off the polyurethane foam and sinter the glass particles, thereby obtaining a positive replica of the template.

2.2 Characterization

The glass-derived scaffolds were characterized by X-ray diffraction (XRD) analysis in order to assess the development of crystalline phases during the thermal treatment. The sintered foam was first ground into a fine powder and then underwent wide-angle (2θ within 10-70°) XRD by using a X'Pert Pro PW3040/60 diffractometer (PANalytical, Eindhoven, The Netherlands) operating at 40 kV and 30 mA

with Bragg-Brentano camera geometry, Cu Kα incident radiation (wavelength λ = 0.15405 nm), step size Δ(2θ) 0.02° and fixed counting time 1 s per step. Identification of crystalline phases was performed by using the X'Pert HighScore software (2.2b) equipped with the PCPDFWIN database (http://pcpdfwin.updatestar.com).

The pore-strut architecture of the scaffold was investigated by field-emission scanning electron microscopy (FESEM; Supra^{TM} 40, Zeiss, Oberkochen, Germany). The samples were sputter-coated with chromium prior to the analysis and inspected at an accelerating voltage of 15 kV.

The tensile strength of sintered scaffolds was assessed according to a previous experimental setup [8]. Before testing, each scaffold was polished with SiC grit paper (#1000 and glued to two loading fixtures, i.e. stainless steel bars (diameter 16 mm) that could be connected by to the testing machine by steel pins. Gluing was performed by using an epoxy resin (Araldite® AV 119, Ciba-Geigy), which is able to withstand a maximum tensile stress of above 40 MPa (as declared by the manufacturer). This adhesive was a gel at room temperature,; its polymerization was achieved by applying a heat treatment at 140 °C for 1 h in an oven. The tensile strength of the scaffolds was calculated as:

$$\sigma_t = \frac{F_t}{A_t} \tag{1}$$

where F_t (N) was the failure load and A_t (mm^2) was the resistant cross-sectional area.

The total porosity p of each scaffold was assessed before the mechanical test by mass-volume experimental measurements as follows:

$$p = 1 - \frac{\rho}{\rho_0} \tag{2}$$

where ρ is the density of the scaffold and ρ_0 is the density of the bulk material (i.e., the foam strut). Results were expressed as mean ± standard deviation on five measurements.

Tensile strength and porosity of five scaffolds belonging to each of the three batches (produced using different solid loads) were assessed. These experimental data were used as input values to the model developed in the section 3.

3. MODELLING

The tensile failure in porous ceramics is caused by the unstable crack propagation within the material. It is well known that the presence of pores causes a decrement of all mechanical properties of solids, including the tensile strength, compared to the bulk (i.e. non-porous) material of equal composition [5]. Hence, the development of reliable models describing the relationship between mechanical properties and porosity are of dramatic importance in materials science and technology as well as in industrial contexts.

Gibson and Ashby [18] derived the so-called "density-power-law model" which states that a given mechanical property (M) of a cellular solid (foam) is controlled by the relative density and the corresponding property of the bulk (M_0), i.e.:

$$\frac{M}{M_0} = C\left(\frac{\rho}{\rho_0}\right)^n \tag{3}$$

where C and n (> 0) are constants depending on the properties of constituent materials (e.g. composition, fine microstructure, morphological characteristics, loading and failure mechanisms).

Substituting Equation (2) in Equation (3), we obtain:

$$M = M_0 C(1-p)^n \tag{4}$$

If applied to determine the elastic modulus E of macroporous foams (such as the bioceramic scaffolds developed in this work), this simple equation can be written as follows [18]:

$$E = E_0(1-p)^2 \tag{5}$$

where C is assumed equal to unity and E_0 is the elastic modulus of the non-porous material.

The surface energy density of a porous solids is expressed as [19,20]:

$$\gamma = \gamma_0 e^{-\beta\left(1-\frac{\rho}{\rho_0}\right)} = \gamma_0 e^{-\beta p} \tag{6}$$

where γ_0 is the surface energy density of the bulk material and β is a constant exponent.

The Poisson's ratio ν depends on porosity p according to the following equation [21]:

$$\nu = 0.5 - \frac{E}{6K} \tag{7}$$

where K = K(p).

For highly-porous ceramics (i.e. if p > 0.40, which is recommended in scaffolds for bone repair), the parameter K can be expressed as a function of p, the bulk modulus (K_0) and the Poisson's ratio of the pore-free material (ν_0) as follows [21]:

$$K = K_0 \frac{2(1-2\nu_0)(1-p)}{3(1-\nu_0)} \tag{8}$$

where $K_0 = \frac{E_0}{3(1-2\nu_0)}$.

Thus, after substituting the expression of K_0 and Equation (8) in Equation (7), we obtain:

$$\nu = 0.5 - \frac{E}{6K} = 0.5 - \frac{3}{4}(1-\nu_0)(1-p) \tag{9}$$

We can assume that the shape of scaffold pores is equivalent to be elliptical along cross-sectional planes (2D geometry). Thus, considering the largest elliptical defect (pore) in the material [22], and according to the recently-developed concepts of quantized fracture mechanics [23], the tensile strength can be expressed as:

$$\sigma_t = \Psi \sqrt{\frac{2E\gamma}{\pi(1-\nu^2)(a+d)}} \tag{10}$$

with:

$$\Psi = \frac{\int_0^{\pi/2} \left(\sqrt{\sin^2 \psi + \left(\frac{a}{b}\right)^2 \cos^2 \psi} \right) d\psi}{\sqrt[4]{\sin^2 \psi + \left(\frac{a}{b}\right)^2 \cos^2 \psi}} \tag{11}$$

wherein *a* and *b* are equivalent to the minor and major axes of the elliptical pore, respectively, and *d* is the grain size. According to Pugno [24], *d* can be assumed as the "fracture quantum" and calculated as:

$$d = \frac{2E_0\gamma_0}{\pi\sigma_0^2} \tag{12}$$

where σ_0 is the tensile strength of the pore-free material.

After substituting Equations (5), (6), (9) and (12) in Equation (10), we obtain:

$$\sigma_t = \sigma_0 \Psi \sqrt{\frac{e^{-\beta p}}{\left\{1 - \left[0.5 - \frac{3}{4}(1-\nu_0)(1-p)^2\right]^2\right\}\left(1 + \frac{a}{d}\right)}} \cdot (1-p) \tag{13}$$

In porous materials, the volume of voids (pores) V_p, the volume of the solid fraction (grains) V_s and the total volume V_{tot} satisfy the following relationship:

$$V_{tot} = V_p + V_s \tag{14}$$

After some rearrangements, Equation (14) can be written as:

$$\frac{V_p}{V_s} = \frac{p}{1-p} \tag{15}$$

We have that $\frac{V_p}{V_s} \to 0$ if $p \to 0$ and $\frac{V_p}{V_s} \to \infty$ if $p \to 1$, which are consistent with physical reality.

Then, we can simply assume that:

$$\frac{V_p}{V_s} \propto \frac{a^3}{d^3} \tag{16}$$

Thus, indicating with ξ a constant shape factor related to pore geometry, Equation (16) can be transformed as follows:

$$\frac{a}{d} = \xi \left(\frac{V_p}{V_s}\right)^{\frac{1}{3}} = \xi \left(\frac{p}{1-p}\right)^{\frac{1}{3}} \tag{17}$$

Finally, substituting Equation (17) in Equation (12), we obtain:

$$\sigma_t = \sigma_0 \Psi \sqrt{\frac{e^{-\beta p}}{\left\{1 - \left[0.5 - \frac{3}{4}(1-\nu_0)(1-p)\right]^2\right\}\left[1 + \xi \left(\frac{p}{1-p}\right)^{\frac{1}{3}}\right]}} \cdot (1-p) \tag{18}$$

If $a = b$, i.e. when elliptical pores can be approximated with circles in 2D – and hence pores are assumed to be spherical in 3D –, we have $\Psi = \pi/2$ [22]. Thus, we obtain this final relationship between tensile strength and porosity in brittle porous solids:

$$\sigma_t = \sigma_0 \sqrt{\frac{\pi e^{-\beta p}}{2\left\{1 - \left[0.5 - \frac{3}{4}(1-\nu_0)(1-p)\right]^2\right\}\left[1 + \xi \left(\frac{p}{1-p}\right)^{\frac{1}{3}}\right]}} \cdot (1-p) \tag{19}$$

The parameter σ_0 was experimentally determined by performing tensile tests on sintered glass powder compacts according to the setup described in the section 2.2. The Poisson's ratio of the bulk material (ν_0) was estimated by means of the software SciGlass implementing the analytical method Priven 2000. The fitting of the experimental data to estimate the unknown parameters of the model (ξ and β) was carried out by using a proper code developed in MATLAB and based on the least squares interpolation.

4. RESULTS AND DISCUSSION

The sintered scaffolds exhibit a glass-ceramic nature, as shown in the XRD pattern reported in Figure 1. Specifically, three crystalline phases were identified, i.e. wollastonite ($CaSiO_3$, PDF code 00-027-0088), calcium borosilicate ($Ca_{11}Si_4B_2O_{22}$, PDF code: 00-045-0001), and cristobalite (SiO_2, PDF code: 01-077-1317). Of those, wollastonite is known to be highly biocompatible and suitable for

manufacturing mechanically-strong bone implants since the 1980s, when Kokubo and co-workers developed machinable and bioactive apatite/wollastonite glass-ceramics [25].

Figure 2a reveals that a positive replica of the porous template was successfully obtained after the thermal treatment, with a good reproduction of the trabecular architecture of the sacrificial polymer. The good densification of scaffold struts is shown in Figure 2b. The surface roughness of the scaffold walls can be attributable to the presence of crystalline phases (Figure 1) that develop during the sintering stage. Potentially, this feature is an interesting added value of the scaffold from a biological viewpoint: in fact, the textural properties of implant surfaces, such as morphology and roughness, are known to greatly influence cell responses *in vitro* and *in vivo* [26]. Early studies carried out in the 1990s have provided a first evidence that osteoblastic cells attach and proliferate preferably on surfaces that exhibit a diffused micrometric roughness [27,28]. Later, it was observed that the micrometric peaks and valleys of implant surfaces can affect the organization of cell cytoskeleton and, hence, the intracellular transduction signaling pathways [29], with an impact on tissue healing and regeneration. Preliminary in vitro biological tests on the glass-ceramic material investigated in this work revealed a good biocompatibility with mesenchymal stem cells and osteoblast-like cells (osteosarcoma Saos-2 line) [30].

Figure 2b also shows that the scaffold macropores have a diameter well above 100 μm and are interconnected, which are key properties to allow body fluids to flow into the scaffold and cells to migrate inside it [3].

The total porosity of the scaffolds varies in the range of 0.40 to 0.79 depending on the solid load used in the slurry during the fabrication process (see also Table 1): in general, higher the solid load, lower the porosity and higher the tensile strength. It is worth highlighting that this range fully covers the typical porosity range of human healthy cancellous bone – from about 0.40 of femur head to 0.80-0.85 of vertebral bodies in the spine [31-33].

The experimental values of tensile strength of the scaffolds, ranging from 2.3 to 7.4 MPa, are comparable to those reported for bovine trabecular bone (7.6 ± 2.2 MPa [34]) and human cancellous bone (10-20 MPa [35]), and are significantly higher than the results assessed for 45S5 Bioglass®-derived glass-ceramic scaffolds (0.011 MPa), even when these are reinforced with a polymeric coating (0.074 MPa) [7]. This high difference (about two orders of magnitude) is due to the different thermal behaviour of the glasses used: in fact, 45S5 Bioglass® exhibit a poor sinterability [36,37], which results in highly brittle scaffolds with hollow struts [17].

The low values of standard deviation reported in Table 1 suggests the good reproducibility of the method used for scaffold fabrication.

As $p > 0.40$, Equation (8) is actually applicable [21] and the theoretical validity of the model (Equation (19)) is confirmed for this case. The model was implemented by using $\nu_0 = 0.2520$ and $\sigma_0 = 45$ MPa as bulk characteristics, assessed as described in the section 2.2. The result of model fitting with experimental data is reported in Figure 3; the model parameters and the correlation coefficient R^2 are collected in Table 2. There is a very good agreement between model results and experimental data, as demonstrated by the high value of the coefficient R^2; this also suggests a good accuracy and reliable predictive capability of the model.

The experimental values were also interpolated by making use of the classical Gibson-Ashby density-power-law model, which is commonly used to describe the mechanical response of cellular foams [18]. When tensile loads are applied to brittle foams, Equation (4) becomes [38]:

$$\sigma_t = \sigma_0 C (1-p)^{\frac{3}{2}} \tag{20}$$

The results of model fitting through Equation (20) are reported in Figure 4. The comparison between the coefficients R^2 (Table 2) suggests a better accuracy and predictive capability of the new model.

The model developed in this work can be virtually applied to any kind of porous materials characterized by brittle mechanical behavior (not only biomedical ceramic scaffolds). If used at the

design stage, this model allows researchers to predict the tensile strength of highly-porous bioceramics – which is typically difficult to assess experimentally – as a function of total porosity and, hence, to optimize the scaffold performance by acting on the material composition and/or processing parameters (e.g. solid load in the slurry, as performed in this work).

In principle, the model proposed in Equation (19) is independent of the particular ceramic or glass composition considered and can be applied to a wide range of porous ceramics and glasses under certain assumptions, i.e. high porosity (above 40 vol.%) and spheroidal pores. If one of these hypotheses is not verified, the model becomes inaccurate and should be modified and refined accordingly to be fully valid again. This could be, for example, the case of 3D-printed scaffolds [39] that often exhibit a grid-like structure formed by parallel channels and struts.

The major advantage of this model is perhaps its relative simplicity, as the tensile strength is supposed to depend only on the total porosity of the material, provided that the bulk characteristics are known. Parameters related to the material morphology, such as surface roughness or size of starting particles, have not been taken into account; some of these aspects, e.g. the dependence of porosity (and hence tensile strength) on grain size were previously analysed by Rice in detail [40]. Furthermore, as already pointed out above, pores are supposed to be spherical, which leads to a dramatic simplification of Equation (11) but does not account for the "shape effects" due to irregular pore geometry. Finally, it is worth underlining that the model is valid for materials where unstable crack propagation occurs (brittle mechanical behaviour): therefore, Equation (19) cannot be applied to ductile materials or composites, e.g. polymer-matrix scaffolds with bioactive glass or ceramic inclusions.

5. CONCLUSIONS

The tensile strength of silicate glass-ceramic scaffolds, which were produced by sponge replication followed by sinter-crystallization, was experimentally assessed and found comparable to that of cancellous bone. The data were interpolated through a model, based on fracture mechanics, that relates tensile strength with total porosity and seems to exhibit better accuracy compared to the classical density-power-law approach. This model can be potentially used for a double purpose: (i) in a direct way, when the scaffold porosity is known, to reliably predict the tensile strength without the need for complicate experimental setups, or (ii) in an inverse way, if a strength target of the implant is fixed for a specific surgical need, to determine the required porosity the scaffold should have. Hence, the final porosity of the scaffold can be tailored by varying the fabrication parameters in a controlled way, so that the material fulfils the desired mechanical requirements. In general, implementing this model for predictive purposes can be useful to save time associated to the otherwise necessary trial-and-error optimization approach in several biomedical and industrial applications.

REFERENCES

[1] Baino F, Vitale-Brovarone C. Three-dimensional glass-derived scaffolds for bone tissue engineering: current trends and forecasts for the future. J Biomed Mater Res A 2011;97:514-535.

[2] El-Rashidy AA, Roether JA, Harhaus L, Kneser U, Boccaccini AR. Regenerating bone with bioactive glass scaffolds: A review of in vivo studies in bone defect models. Acta Biomater 2017;62:1-28

[3] Karageorgiou V, Kaplan D. Porosity of 3D biomaterial scaffolds and osteogenesis. Biomaterials 2005;26:5474-5491.

[4] Hong Y, Chen X, Jing X, Fan H, Gu Z, Zhang X. Fabrication and drug delivery of ultrathin mesoporous bioactive glass hollow fibers. Adv Funct Mate 2010;20:1503-1510.

[5] Fu Q, Saiz E, Rahaman MN, Tomsia AP. Bioactive glass scaffolds for bone tissue engineering: state of the art and future perspectives. Mater Sci Eng C 2011;31:1245-1256.

[6] Thompson JD, Hench LL. Mechanical properties of bioactive glasses, glass-ceramics and composites. Proc Inst Mech Eng H J Eng Med. 1998;212:127-136.

[7] Rehorek L, Chlup Z, Meng D, Yunos DM, Boccaccini AR, Dlouhy I. Response of 45S5 bioglass® foams to tensile loading. Ceram Int 2013;39:8015-8020.

[8] Chen Q, Baino F, Pugno NM, Vitale-Brovarone C. Bonding strength of glass-ceramic trabecular-like coatings to ceramic substrates for prosthetic applications. Mater Sci Eng C 2013;33:1530-1538.

[9] Brezny R, Green DJ. Uniaxial strength behaviour of brittle cellular materials. J Am Ceram Soc 1993;76:2185-2192.

[10] Lacroix D, Chateau A, Ginebra MP, Planell JA. Micro-finite element mod-els of bone tissue-engineering scaffolds. Biomaterials 2006;27:5326-5334.

[11] Chen Q, Huang S. Mechanical properties of a porous bioscaffold withhierarchy. Mater Lett 2013;95:89-92.

[12] Jones JR, Atwood RC, Poologasundarampillai G, Yue S, Lee PD. Quantifying the 3D macrostructure of tissue scaffolds. J Mater Sci Mater Med 2009;20:463-471.

[13] Tagliabue S, Rossi E, Baino F, Vitale-Brovarone C, Gastaldi D, Vena P. Micro-CT based finite element models for elastic properties of glass-ceramic scaffolds. J Mech Behavior Biomed Mater 2017;65:248-255.

[14] Baino F, Minguella J, Kirk N, Montealegre MA, Fiaschi C, Korkusuz F, Orlygsson G, Vitale-Brovarone C. Novel full-ceramic monoblock acetabular cup with a bioactive trabecular coating: design, fabrication and characterization. Ceram Int 2016;42:6833-6845.

[15] Baino F, Montealegre MA, Orlygsson G, Novajra G, Vitale-Brovarone C. Bioactive glass coatings fabricated by laser cladding on ceramic acetabular cups: a proof-of-concept study. J Mater Sci 2017;52:9115-9128.

[16] Baino F, Ferraris M, Bretcanu O, Verné E, Vitale-Brovarone C. Optimization of composition, structure and mechanical strength of bioactive 3-D glass-ceramic scaffolds for bone substitution. J Biomater Appl 2013;27:872-890.

[17] Chen QZ, Thompson ID, Boccaccini AR. 45S5 Bioglass®-derived glass-ceramic scaffolds for bone tissue engineering. Biomaterials 2006;27:2414-2425.

[18] Gibson LJ, Ashby F. The mechanics of three-dimensional cellular materials. Proc R Soc Lond A 1982;382:43-59.

[19] Rice RW. Grain size and porosity dependence of ceramic fracture energyand toughness at 22°C. J Mater Sci 1996;31:1969-1983.

[20] Sadowski T, Samborski S. Modeling of porous ceramics response to compressive loading. J Am Ceram Soc 2003;86:2218-2221.

[21] Arnold M, Boccaccini AR, Ondracek G. Prediction of the Poisson's ratio of porous materials. J Mater Sci 1996;31:1643-1646.

[22] Irwin GR. Crack-extension force for a part-through crack in a plate. J Appl Mech 1962;29:651-654.

[23] Pugno N, Ruoff R. Quantized fracture mechanics. Philos Mag 2004;84:2829-2845.

[24] Pugno N. Dynamic quantized fracture mechanics. Int J Fract 2006;140:159-168.

[25] Kokubo T, Ito S, Sakka S, Yamamuro T. Formation of a high strength bioactive glass-ceramic in the system $MgO–CaO–SiO_2–P_2O_5$. J Mater Sci 1986;21:536-540.

[26] Feller L, Jadwat Y, Khammissa RAG, Meyerov R, Schechter I, Lemmer J. Cellular responses evoked by different surface characteristics of intraosseous Ti implants. Biomed Res. Int 2015;2015: 171945.

[27] Schwartz Z, Boyan BD. Underlying mechanisms at the bone-biomaterial interface. J Cell Biochem 1994;56:340-347.

[28] Anselme K, Davidson P, Popa AM, Giazzon M, Liley M, Ploux L. The interactions of cells and bacteria with surfaces structured at the nanometer scale. Acta Biomater 2010;10:3824-12846.

[29] Spriano S, Yamaguchi S, Baino F, Ferraris S. A critical review of multifunctional titanium surfaces: New frontiers for improving osseointegration and host response, avoiding bacteria contamination. Acta Biomater 2018;79:1-22.

[30] Baino F, Minguella-Canela J, Korkusuz F, Korkusuz P, Kankılıç B, Montealegre MA, De los Santos-López MA, Vitale-Brovarone C. In vitro assessment of bioactive glass coatings on alumina/zirconia composite implants for potential use in prosthetic application. Int J Mol Sci 2019;20:722.

[31] Hildebrand T, Laib A, Muller R, Dequeker J, Ruegsegger P. Direct three dimensional morphometric analysis of human cancellous bone: microstructural data from spine, femur, iliac crest, and calcaneus. J Bone Miner Res 1999;14:1167-1174.

[32] Eckstein F, Matsuura M, Kuhn V, Priemel M, Muller R, Link TM, Lochmuller E. Sex differences of human trabecular bone microstructure in aging are site-dependent. J Bone Miner Res 2007;22:817-824.

[33] Greenwood C, Clement JG, Dicken AJ, Evans JPO, Lyburn ID, Martin RM, Rogers KD, Stone N, Adams G, Zioupos P. The micro-architecture of human cancellous bone from fracture neck of femur patients in relation to the structural integrity and fracture toughness of the tissue. Bone Rep 2015;3:67-75.

[34] Kaplan SJ, Hayes WC, Stone JL, Beaupré GS. Tensile strength of bovine trabecular bone. J Biomech 1985;18:723–727.

[35] Doblaré M, Garca JM, Gomez MJ. Modelling bone tissue fracture and healing: a review. Eng Fract Mech 2004;71:1809-1840.

[36] Golovchak R, Thapar P, Ingram A, Savytskii D, Jain H. Influence of phase separation on the devitrification of 45S5 bioglass. Acta Biomater 2014;10:4878-4886.

[37] Baino F, Fiume E. Quantifying the effect of particle size on the crystallization of 45S5 bioactive glass. Mater Lett 2018;224:54-58.

[38] Liu P, Fu C, Li T, Shi C. Relationship between tensile strength and porosity for high porosity metals. Sci China E 1999;42:100-107.

[39] Gmeiner R, Deisinger U, Schonherr J, Lechner B, Detsch R, Boccaccini AR, Stampfl J. Additive manufacturing of bioactive glasses and silicate bioceramics. J Ceram Sci Technol 2015;6:75-86.

[40] Rice RW. Relation of tensile strength-porosity effects in ceramics to porosity dependence of Young's modulus and fracture energy, porosity character and grain size. Mater Sci Eng A 1989;l12:215-224.

Figure legends

Figure 1. XRD patterns of powdered scaffolds after sintering at 900 °C for 3 h.

Figure 2. SEM micrographs showing the pore-strut structure of sintered scaffolds at different magnifications: (a) 50× and (b) 1300× .

Figure 3. Fitting of the tensile strength data by applying the model developed in this work (Equation (19)). black points – experimental values, blue line – model interpolation.

Figure 4. Fitting of the tensile strength data by applying the Gibson-Ashby model (Equation (20)). black points – experimental values, red line – model interpolation.

Tables

Table 1. Total porosity of the scaffolds.

Scaffold batch	p
Solid load = 30 wt.%	0. 75 ± 0.028
Solid load = 35 wt.%	0.63 ± 0.019
Solid load = 40 wt.%	0.42 ± 0.026

Table 2. Model parameters and coefficients of determination.

Model type	Model parameters	R^2
Present work (Equation (19))	$\xi = 44.4$; $\beta = 0.03529$	0.88
Density-power-law (Equation (20))	$C = 0.3385$	0.82