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Compressive Properties and Creep Resistance of a Novel, Porous, Semidegradable Poly(vinyl alcohol)/Poly(lactic-*co*-glycolic acid) Scaffold for Articular Cartilage Repair

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ABSTRACT: Tissue engineering for articular cartilage repair has shown success in ensuring the integration of neocartilage with surrounding natural tissue, but the rapid restoration of biomechanical functions remains a significant challenge. The poly(vinyl alcohol) (PVA) hydrogel is regarded as a potential articular cartilage replacement for its fair mechanical strength, whereas its lack of bioactivity limits its utility. To obtain a scaffold possessing expected bioactivity and initial mechanical properties, we herein report a novel salt-leaching technique to fabricate a porous PVA hydrogel simultaneously embedded with poly(lactic-*co*-glycolic acid) (PLGA) microspheres. Through the investigation of environmental scanning electron microscopy, we found that the porous PVA/PLGA scaffold was successfully manufactured. The compression and creep properties were also comprehensively studied before and after cell culturing. The relationship between the compressive modulus and strain ratio of the porous PVA/PLGA scaffold showed significant nonlinear behavior. The elastic compressive modulus was influenced a little by the porogen content, whereas it went higher with a higher PLGA microsphere content. The cell-cultured scaffolds presented higher compressive moduli than the initial ones. The creep resistance of the cell-cultured scaffolds was much better than that of the initial ones. In all, this new scaffold is a promising material for articular cartilage repair. © 2013 Wiley Periodicals, Inc. J. Appl. Polym. Sci. **2014**, *131*, 40311.

KEYWORDS: biodegradable; biomaterials; biomedical applications; crosslinking

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INTRODUCTION

Natural articular cartilage lacks the ability of self-repair and regeneration after injury, which makes it necessary to pursue new technologies to solve this problem.[1–3] With regard to repair materials, tissue engineering scaffolds fabricated with biodegradable materials and poly(vinyl alcohol) (PVA) hydrogel possessing a special three-dimensional structure have been proposed as two possible ways to achieve articular cartilage repair.[4–7] However, the poor mechanical properties of tissue engineering scaffolds and the nonbioactivity of PVA hydrogels limit their applications.[4,8]

Recently, a novel design for semidegradable scaffolds was proposed; it combined the bioactivity of tissue engineering materials and the adequate mechanical properties of PVA hydrogels with the aim of producing a regenerative tissue integrated with the surrounding natural cartilage in a load-bearing environment.[9,10] This scaffold involves several attractive features for biomedical application. Previous mechanical characterization of semidegradable scaffolds has mainly focused on compressive modulus testing.[11,12] In fact, as a replacement for articular cartilage, it is necessary to estimate the scaffolds' comprehensive compressive characteristics and creep resistance. There has been little investigation on this matter.

In this study, a novel, porous, semidegradable PVA/poly(lactic-*co*-glycolic acid) (PLGA) scaffold was prepared by the freezing–thawing and salt-leaching method. Meanwhile, PLGA microspheres were embedded into the matrices. The composite scaffold was hypothesized to have proper initial mechanical properties and to be beneficial for tissue engineering. Additionally, scaffolds were seeded with chondrocytes and cultured *in vitro* for 4 weeks; this was regarded as a further study of scaffolds for implantation. For both the initial scaffolds and the cell-cultured ones, the influences of the porogen and PLGA contents on the compressive properties and creep resistance were investigated. A comparison between the initial and cell-cultured scaffolds was also carried out to further estimate the possibility for cartilage repair.

EXPERIMENTAL

Materials

PLGA (50:50 lactide-to-glycolide ratio, $M_w = 65{,}000$) was purchased from Jinan Daigang Biomaterial Co., Ltd. (China). PVA (>99% hydrolyzed, $M_w = 89{,}000$–$98{,}000$, and low viscosity = 1788) was obtained separately from Sigma-Aldrich (St. Louis, MO) and Aladdin Reagent Co., Ltd. (China). Poly(vinyl pyrrolidone) (PVP; K-30), dichloromethane, acetone, sodium chloride powder, silver nitrate, and physiological saline solution were all procured from Sinopharm Chemical Reagent Co., Ltd. (China). Emulsifier octylphenol polyethoxylate (OP) was purchased from Shanghai Jiuyi Chemical Reagent Co., Ltd. (China). Bovine serum was gained from Nanjing Bioon Bio-Technology Co., Ltd. (China). Dulbecco's modified Eagle's medium, fetal bovine serum, and trypsin–ethylene diamine tetraacetic acid (EDTA) solution (0.05% trypsin and 0.02% EDTA) were purchased from Gibco Invitrogen (Carlsbad, CA).

Preparation of the PLGA Microspheres

PLGA microspheres were prepared by a modified emulsion–evaporation technique.[13] First, 480 mg of PLGA was added to 16 mL of an dichloromethane and acetone mixed solution (volume ratio = 3:1) to form the PLGA solution through sonication. Second, the solution was added dropwise to a PVA solution (2%w/w, low viscosity = 1788), which was stirred at a speed of 1000 rpm with a magnetic stirrer (IKA Big Squid) for about 10 min to form a homogeneous suspension. Then, the solution was subsequently stirred for 12 h to fully evaporate the dichloromethane and acetone. The resulting PLGA microspheres from centrifugation were washed with distilled water, filtered, lyophilized, and stored in desiccators for future use.

Preparation of the Semidegradable Scaffold

A semidegradable scaffold was fabricated through the freezing–thawing method. A detailed description of the process is given as follows. A 15 wt % homogeneous PVA solution (a little PVP was also added) was prepared by 10 h of stirring at 95°C, and it was then placed in air to cool it to room temperature for further use. A composite porogen prepared with an OP emulsifier and a proper weight of NaCl particles (mass ratio of OP to NaCl = 2:1), mixed with PLGA microspheres was added dropwise to the previous PVA solution under stirring conditions. After that, the mixture was continuously stirred for 2 h to form the homogeneous solution. Finally, the mixed solution was poured into a customized poly(methyl methacrylate) (PMMA) mold (a designed disk shape with a size of 4 mm in height and 12 mm in diameter), and this was subjected to seven freezing–thawing cycles consisting of 21 h at −20°C and followed by 3 h at 25°C. The resulting hydrogels were taken out from the mold and were washed in distilled water with thermostatic ultrasonic equipment. The porous, semidegradable scaffold was eventually gained until the NaCl particles were completely washed out.

In this study, the contents of the composite porogen and PLGA microsphere were two variable parameters. All of the specimens were labeled *A*/*B*, where *A* stands for the composite porogen content relative to the PVA/PVP solution and *B* stands for the PLGA microsphere content relative to those of dry PVA and the PVP powder.

Cell Isolation and *In Vitro* Culturing

Chondrocytes were isolated from the articular surface of a mature male New Zealand white rabbit (4 months old, 2.5 kg) under sterile conditions, as described by Iwasaki et al.,[14] and expanded in a culture medium containing 90% Dulbecco's modified Eagle's medium, 10% fetal bovine serum, 50 mg/mL ascorbic acid, 100 U/mL penicillin, and 100 μg/mL streptomycin at 37°C in a humidified incubator containing 5% CO_2. The medium was changed every 2 days. When the adherent cells reached subconfluence, they were passaged after trypsinization (0.05% trypsin and 0.02% EDTA). The second-passage cells were harvested by trypsinization followed by the addition of fresh culture medium to create a new suspension with a concentration of 2×10^7 cells/mL. The hydrogel samples prepared with different porogens and PLGA contents were inoculated in six-well, flat-bottom culture plates. A volume of 10 μL of cell suspension was added to each sample located at the center of the well. The cell-seeded hydrogel constructs were cultured at 37°C in 5% CO_2, and the medium was changed every 3 days. All of the samples were cultured for 4 weeks before further study.

Microstructure Study

The morphologies of the semidegradable scaffolds prepared with various composite porogens was observed with a Philips XL-30 environment scanning electron microscope at 20 kV. The pure PVA hydrogel was also studied as the blank sample. To maintain the real microstructure of the scaffold under wet conditions, all of the samples were observed directly without any pretreatment.

Elastic Compressive Modulus

The compressive modulus measurement was performed on an electromechanical material testing machine (model Instron E1000), and the samples were in unconfined compression between two stainless platens. During the whole compressive test, the samples were immersed in a phosphate buffer saline (PBS) solution bath and were compressed to 60% of their original height. The test strain rate of the sample was 4 mm/min. The slope of the stress–strain curve, taken as 14–16%, was similar to that of the line and was defined as the elastic compressive modulus. In this study, the initial scaffolds and the scaffolds cultured for 4 weeks were tested as a comparison. For each sample, three measurements were taken, and the average value was reported.

Effect of the Strain Ratio on the Compressive Modulus

The compressive process was performed as described in the previous section. The compressive moduli of the scaffolds at different compression strain levels were determined by the finite difference method and were expressed with the following equation:[15]

$$E_\varepsilon = \frac{\sigma_{\varepsilon+\Delta\varepsilon} - \sigma_{\varepsilon-\Delta\varepsilon}}{2\Delta_\varepsilon}$$

where E_ε is the compressive modulus of the scaffolds at a compression strain ratio value of ε, $\sigma_\varepsilon + \Delta\varepsilon$ is the stress value at $\varepsilon + \Delta\varepsilon$ strain ratio, $\sigma_\varepsilon - \Delta\varepsilon$ is the stress value at $\varepsilon - \Delta\varepsilon$ strain

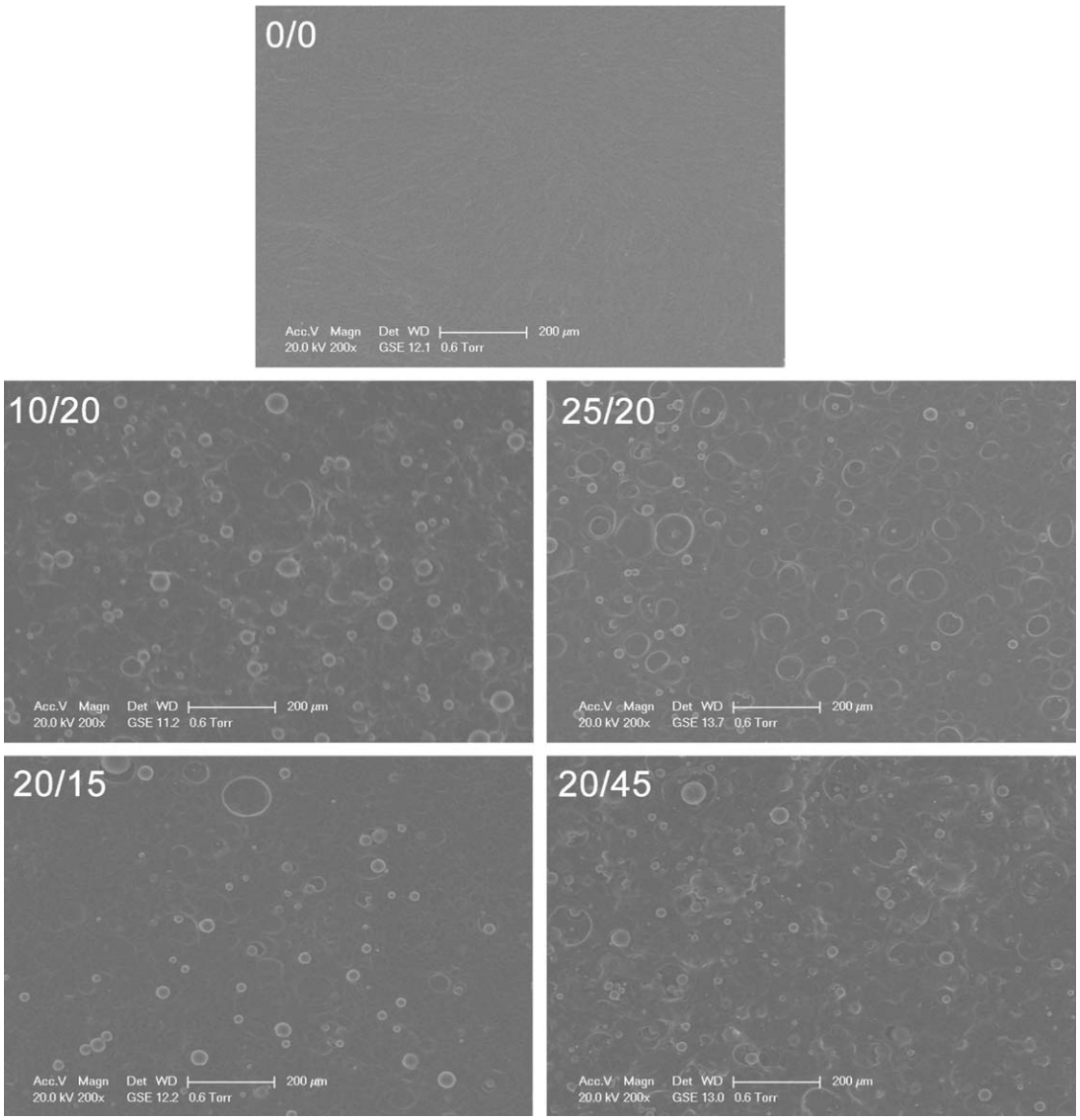

Figure 1. Microstructure of the PVA hydrogel and porous, semidegradable scaffolds.

ratio and $\Delta\varepsilon$ is the variable of strain value. The value of $\Delta\varepsilon$ was 1%. The measurement of each sample was repeated three times.

Creep Study

One of the several requirements of a hydrogel interposition device to maintain its functional performance in the long term is a high creep resistance.[16] The creep test was carried on in a PBS solution bath at room temperature on a HY-0230 electronic universal testing machine (Hengyi Co., Shanghai). The test samples were placed between two stainless steel compression plates, and the upper plate was lowered until it touched the top surface of the sample. The whole test process mimicked those reported in prior literature.[16] The compressive load was first ramped at a rate of 50 N/min to a load of 50 N; this was maintained constant for 100 min. The load was subsequently reduced at a rate of 50 N/min to 5 N, and this load was also kept constant for 100 min. The displacement and time were recorded to evaluate the creep properties.

RESULTS

Microstructure of the Porous, Semidegradable Scaffolds

Figure 1 demonstrates the microstructure of the pure PVA hydrogel (0/0). The surface structure of the hydrogel was homogeneous, and there were no obvious micropores on the surface; this was different from the previous study sample observed after drying.[17] As cartilage replacements are used in a physiological environment, hydrogels studied in wet conditions could reflect their true microstructure. Therefore, all of the samples in this study were observed in wet conditions by environmental scanning electron microscopy (ESEM). On one hand, it is shown in Figure 1 (10/20 and 25/20 samples) that the quantity of the pores significantly increased with increasing composite porogen, and the PLGA particles were distributed uniformly in the hydrogel matrix. The pores were vital for ensuring the cell culturing process, in which the cells needed to migrate and grow through pores.[18] On the other hand, the effect of the PLGA microsphere content on the morphology is shown in Figure 1

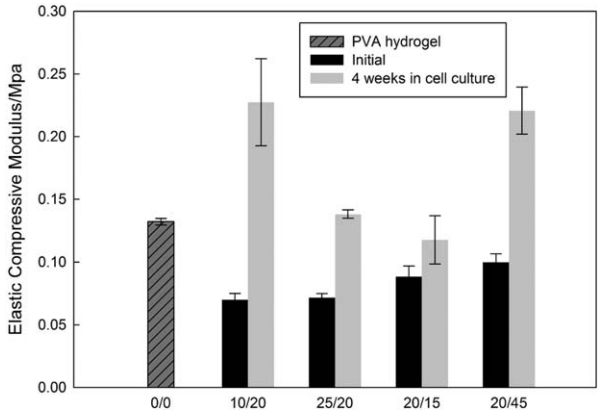

Figure 2. Elastic compressive moduli of the PVA hydrogel and porous, semidegradable scaffolds.

(20/15 and 20/45 samples). Except for the increasing microsphere amount in the hydrogel matrix, there was no obvious change in the microstructure. What we observed from the ESEM observation was that the composite porogen and PLGA microspheres, as two key parameters, were capable of adjusting the structure of the semidegradable scaffold to meet the requirements for cartilage replacement.

Elastic Compressive Moduli of the Porous, Semidegradable Scaffolds

The elastic compressive moduli of the scaffolds and PVA hydrogels are shown in Figure 2. Compared to the blank sample, the elastic compressive moduli of all of the initial scaffolds were relatively smaller. At the same time, the composite porogen did not have an obvious effect on the elastic compressive modulus; this, however, increased to a certain extent with increasing PLGA content. On the other hand, the elastic compressive modulus of the scaffolds seeded with chondrocytes and cultured for 4 weeks all significantly rose compared to initial ones. This was fully proven by both the 10/20 and 20/45 samples, whose elastic compressive modulus increased from 0.070 to 0.227 MPa and from 0.100 to 0.222 MPa, respectively.

Relationship Between the Compressive Modulus and the Strain Ratio

Figure 3(a,b) reveals the relationship between the compressive modulus and strain ratio of scaffolds manufactured with different amounts of composite porogen. We observed that the compressive modulus of the scaffolds increased exponentially with increasing compression; this indicated that the compressive modulus depended significantly on the compression ratio. Furthermore, for the initial samples, when the strain ratio was increased from 10 to 60%, the compressive moduli of specimens 10/20 and 25/20 rose from 0.05 to 1.63 MPa and from 0.05 to 1.90 MPa, respectively. The compressive modulus increased by more than 30 times. The composite porogen content did not obviously affect the compressive behavior, and the compressive modulus of the 10/20 and 25/20 samples trended to the same level at each strain ratio. For the cell-cultured samples, when the strain ratio increased from 10 to 60%, the compressive modulus of the cell-cultured specimens 10/20 and 25/20 rose from 0.20 to 5.33 MPa and 0.11 to 3.54 MPa, respectively. The cell-cultured scaffolds cultured for 4 weeks prepared with more composite porogen presented a low compressive modulus at the same strain ratio. On the other hand, for both samples 10/20 and 25/20, we found that the compressive modulus of the cell-cultured sample cultured for 4 weeks was significantly higher than that of the initial one at every compressive strain level. Moreover, with increasing compression ratio, the deviation extent of the initial and cell-cultured sample increased.

Figure 3(c,d) shows the relationship between the compressive modulus and the strain ratio of the porous scaffold prepared with different PLGA microsphere amounts. The compressive modulus also revealed a significant dependence on the compression ratio and increased exponentially with increasing compression. As the strain ratio increased from 10 to 60%, the compressive modulus of the initial 20/15 sample rose from 0.61 to 2.36 MPa; that of the initial 20/45 sample increased from 0.65 to 4.66 MPa. The compressive modulus increased by about four times on 20/15 and eight times on 20/45. For the initial scaffold, the sample prepared with more PLGA microspheres presented a higher compressive modulus. For example, under the 60% strain condition, the compressive modulus of 20/45 was nearly two times higher than that of 20/15. In a comparison of the initial and 4-week cell-cultured samples, for sample 20/15, the compressive modulus of the cell-cultured sample was higher than that of the initial one during the whole compression process. For sample 20/45, when the strain ratio was lower than 58%, the compressive modulus of the cell-cultured sample was higher than that of the initial one; however, the relationship reversed when the strain ratio exceeded 58%.

Creep Resistance Properties

Figure 4(a,b) demonstrates the creep properties of the porous, semidegradable scaffolds prepared with different composite porogen contents. We found that the total strain of the 10/20 initial sample under a load of 50 N for 100 min (TS_1) was 14.1% and the stain during the process of releasing the load from 50 to 5 N (TS_2) was 22.5%. In the end of the test, the whole strain (TS_3) was 45.5%. The TS_1, TS_2, and TS_3 values of the initial 25/20 specimen were 35.6, 15.7, and 79.1%, respectively. Through a comparison of all of the strain data between the initial 10/20 and 25/20 samples, it was easy to discover that sample 25/20 had poor creep resistance. For the samples cultured *in vitro* for 4 weeks, the TS_1, TS_2, and TS_3 values of sample 10/20 were 7.86, 20.4, and 42.3%, respectively, and the corresponding data of sample 25/20 were 24.5, 14.4, and 69.2%, respectively. After *in vitro* culturing, the sample prepared with more composite porogen still presented poor creep resistance. When we observed Figure 5(a,b) individually, the creep resistance of the cell-cultured scaffolds was found to be obviously superior to that of the initial ones. The TS_1 value of 10/20 decreased from 14.1 to 7.86%, and the TS_1 value of 25/20 decreased from 35.6 to 24.5%. The total strain also decreased significantly.

Figure 4(c,d) shows the creep resistance properties of the porous scaffolds with different amounts of PLGA microspheres.

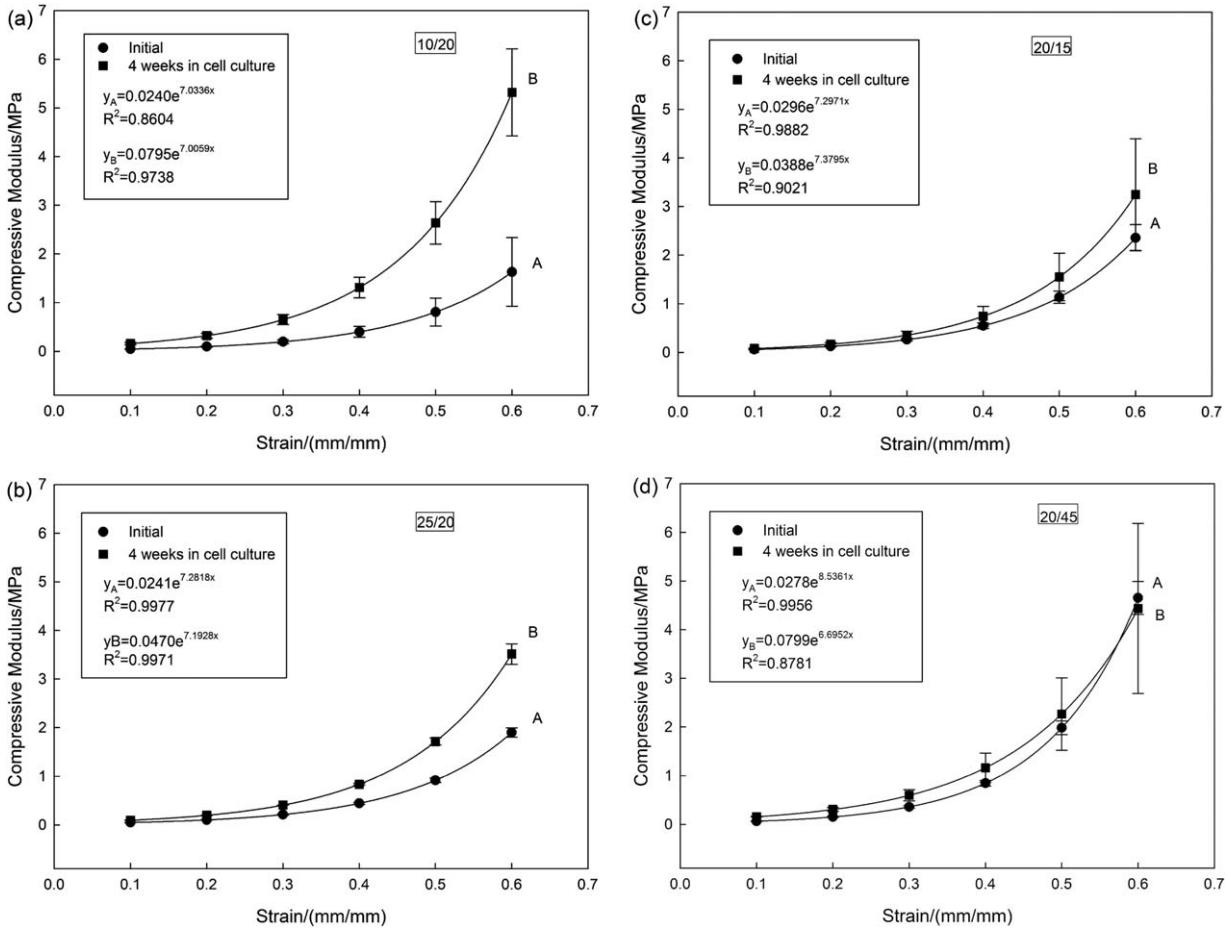

Figure 3. Effect of the strain ratio on the compressive modulus: (a,b) samples prepared with various porogen contents and (c,d) samples manufactured with different PLGA contents; $y = f(x)$ is the regression function of data point and R^2 is the degree of fitting.

In a comparison of Figure 4(c) with Figure 4(d), we observed that the initial sample 20/45 revealed better creep resistance than the initial sample 20/15. The TS_1, TS_2, and TS_3 values of 20/15 were 33.1, 17.6, and 69.4%, respectively, whereas the corresponding data of sample 20/45 were 26.4, 13.7, and 66.6%. However, for the 4-week cell-cultured samples, the result was different. The creep resistance of sample 20/15 was better than that of 20/45. The TS_1, TS_2, and TS_3 values of sample 20/15 were 7.4, 19.8, and 41.5%, respectively, whereas the sample 20/45 showed values of 17.0, 12.4, and 58.5%. Through the observation of Figure 4(c) and Figure 4(d) individually, we also concluded that the 4-week cell-cultured sample demonstrated better creep resistance than the initial one.

DISCUSSION

In this study, a novel scaffold that possessed bioactivity and some certain mechanical properties was prepared. Our primary goal was to observe the comprehensive mechanical properties, especially for the scaffolds cultured for 4 weeks *in vitro*. As a result, we could estimate the possibility of this scaffold for articular cartilage repair through a comparison between the initial and cell-cultured scaffolds. The hypothesis was that the compressive properties and creep resistance would improve after cell culturing. If the scaffold presented a prospective trend, it would potentially show better mechanical performance after implantation. According to our expectations, after a certain period of time, the regenerative composite containing the PVA hydrogel and cartilage tissue would successfully become a replacement for injured cartilage. What is more, the composite porogen and PLGA contents, as two vital factors in the preparation of the scaffolds, were also studied to estimate their effects on the mechanical behavior.

To explain the results better, we provided a schematic (Figure 5) about the process of scaffold formation and cell culturing on the basis of some reported studies.[8,10,11,17,19] Through the freezing–thawing method, the PVA hydrogel matrix formed. It is well known that a PVA hydrogel is a three-dimensional network structure. However, because of the existence of the composite porogen and PLGA microspheres, a different structure containing pores, PLGA microspheres, and crystalline PVA appeared (Figure 1). This porous structure impacted the mechanical behavior of the scaffold. Compared to that of the pure PVA hydrogel, the compressive modulus of all of the scaffolds decreased (Figure 2). Although the compressive properties were worse than those of the PVA hydrogel, the values were in the acceptable range (many tissue engineering scaffolds for cartilage repair present poor compressive properties[20–22]).

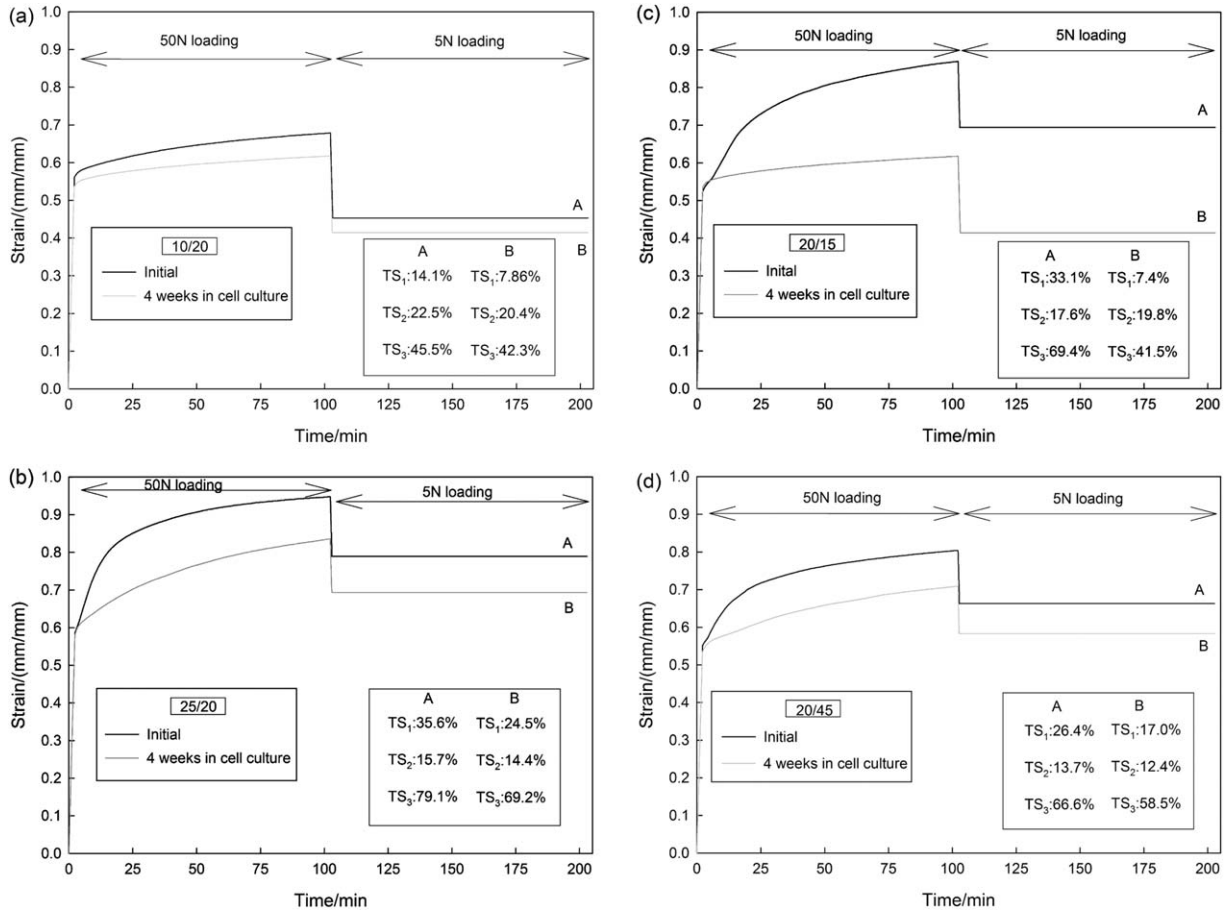

Figure 4. Creep resistance behavior of semidegradable scaffolds before and after cell culturing: (a,b) samples prepared with various porogen contents and (c,d) samples manufactured with different PLGA contents. TS_1 is the strain of the first compression to 50 N and with a 50 N load for 100 min, TS_2 is the strain of the releasing load from 50 to 5 N, and TS_3 is the stain of the whole creep test.

Our design of this porous, semidegradable scaffold was mainly for cell seeding and culturing. We expected that the cultured scaffolds would present better mechanical behavior than the initial ones. During the cell-seeding process, the chondrocytes adhered to the PLGA microspheres, which were a kind of bioactive material. After 4 weeks of cell culturing, the PLGA partly degraded, and the chondrocytes proliferated and secreted extracellular matrix in the pores. Therefore, the structure of the scaffold transformed to the structure described in Figure 5(e), which determined the compressive properties and creep resistance of the scaffolds cultured *in vitro* for 4 weeks.

The excellent mechanical behavior of natural cartilage under physiological conditions could be explained by biphasic theory (solid phase and interstitial fluid phase).[23] Articular cartilage, as a soft hydrated tissue, contains water in a range of 65–85% by weight.[24] It has been recognized that interstitial water pressurizes when loaded.[25] Moreover, the fluid pressurization has been hypothesized to be a major factor in the load-support mechanism, and some studies have shown that the load shared by the interstitial fluid may represent upward to 90%.[26] Previous works have shown that the PVA hydrogel is a three-dimensional structure containing a great amount of water.[27] The special structure of the PVA hydrogel is also composed of a PVA solid phase and an interstitial fluid phase. Therefore, it properly explains the compressive properties and creep resistance of the porous, semidegradable PVA/PLGA scaffold.

When the scaffolds were suffering from load, the change in structure could be explained as shown in Figure 6. In the compressive experiments, the interstitial fluid phase of the initial scaffolds was easily squeezed out because of the porous structure; this led to the effect in which the fluid pressurization decreased. Consequently, the initial scaffolds showed lower elastic compressive moduli compared to the pure PVA hydrogel. The porogen content did not affect the elastic compressive modulus significantly; this may have been due to the high porosity. The effect of the PLGA content on the elastic compressive modulus could be explained by the fact that PLGA was similar to a hard phase improving the mechanical properties of the matrix. After 4 weeks of *in vitro* cell culturing, the secreted extracellular matrix occupied the pores and changed the structure of the scaffolds. As a result, the fluid phase in the cell-cultured scaffold was not easily squeezed out under the loading condition compared to in the initial one. In some cases, the structure was compact enough to present a higher elastic compressive modulus than that of the pure PVA hydrogel (Figure 2).

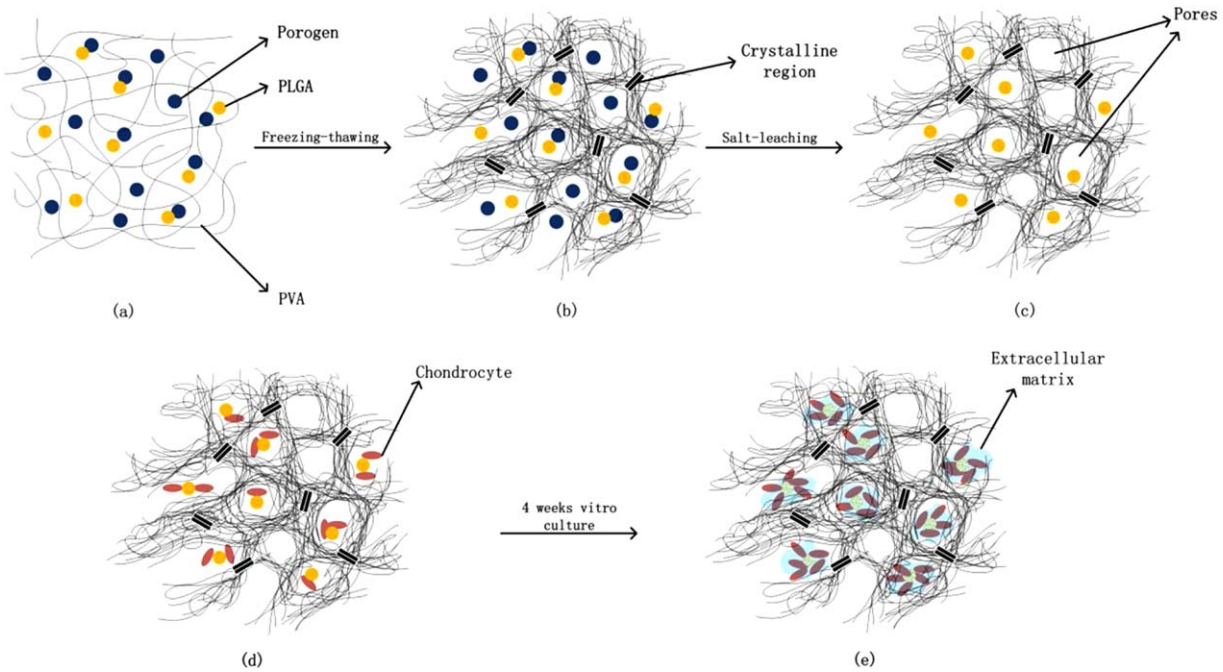

Figure 5. Schematic of scaffold formation and cell culturing. (a) Uniform mixture of the PLGA microspheres, porogen, and PVA solution at room temperature. (b) During the freezing–thawing cycles, phase separation began and forced PVA to become crystalline. (c) After the salt-leaching, on the one hand, the porogen was removed, and pores emerged. On the other hand, the PLGA microsphere was still in the matrix. (d) After cell seeding, the chondrocytes adhered to the PLGA microspheres. (e) During the 4 weeks of *in vitro* culturing, the chondrocytes proliferated and secreted extracellular matrix into the pores. [Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

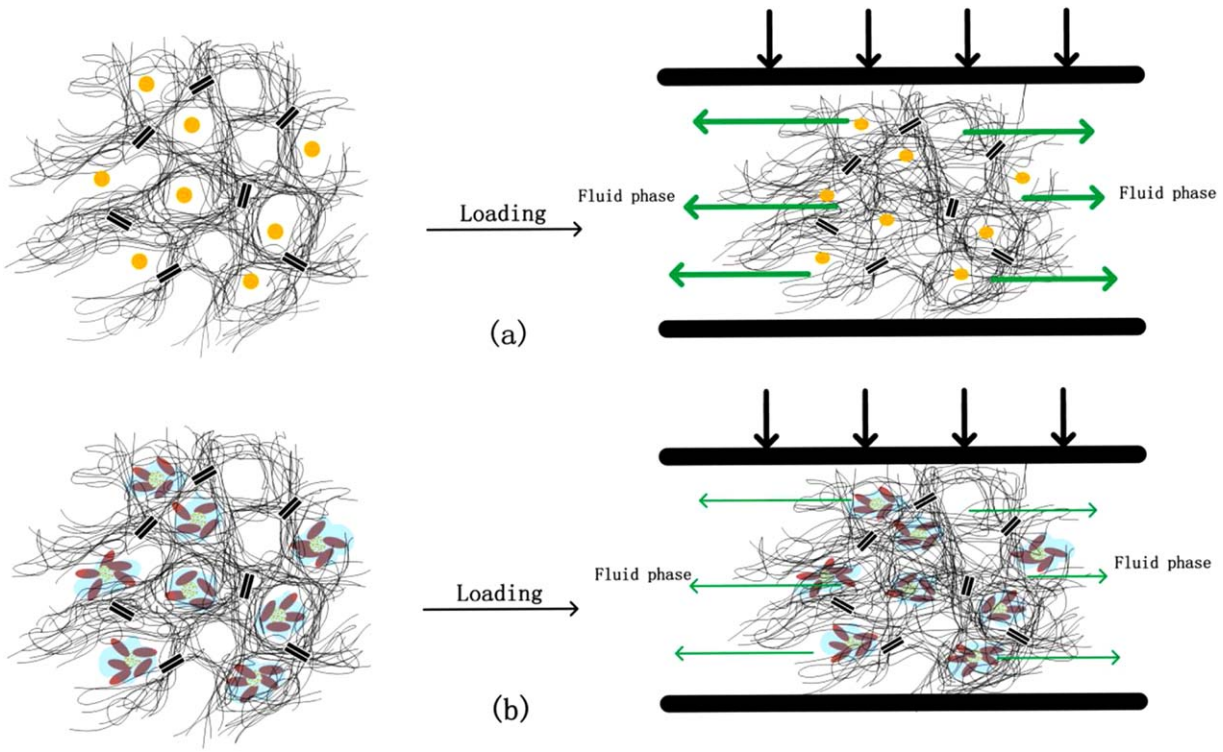

Figure 6. Schematic of scaffold suffering with an applied load: (a) the initial scaffold and (b) the scaffold after 4 weeks of cell culturing. [Color figure can be viewed in the online issue, which is available at wileyonlinelibrary.com.]

In the study of the effect of the strain ratio on the compressive modulus, all of the samples showed significant viscoelasticity, which is also a vital property of natural articular cartilage.[28] On this point, the semidegradable scaffold met the requirement of cartilage replacement. For samples 10/20 and 25/20, the latter one revealed a lower compressive modulus because it had relatively fewer extracellular matrices in more pores. However, in the initial conditions, the common porous structure resulted in similar compressive moduli. The difference was not magnified. For samples 20/15 and 20/45, as the PLGA microspheres distributed in the matrix were similar to a hard second phase particulate, playing a role in strengthening the hydrogel matrix. The compressive modulus of the initial 20/45 sample was higher than that of the initial 20/15 sample. After 4 weeks of *in vitro* culturing, the compressive modulus increased because of the effect of the extracellular matrix in the pores. However, for sample 20/45, there was more undegradable PLGA in the matrix, and this resulted in a higher compressive modulus when the strain ratio rose to a certain value.

For the part of creep study, the results could also be explained by the microstructure. On one hand, from the results of the microstructure (Figure 1), we concluded that the amount of porogen directly determined the quantity of pores. For the sample with a higher porosity, the liquid was more easily squeezed out from the network structure under applied loads. Under the same load level, creep easily occurred. Therefore, the 25/20 sample prepared with more composite porogen was inferior in creep resistance. The poorer creep resistance of the 25/20 sample after *in vitro* culturing may have resulted from the relatively higher porosity. On the other hand, the PLGA microsphere was similar to the hard-phase particles; this enhanced the strength of the matrix under the effect of an applied load. Thus, the sample prepared with more PLGA presented better creep resistance. During the process of *in vitro* culturing, the PLGA microspheres degraded gradually to lactic acid, leaving holes, so sample 20/45, containing more PLGA, presented more pores; this made the creep properties of the scaffold go bad. Although the extracellular matrix effectively improved the creep resistance of the scaffold, new pores may have had a greater impact on the performance of the creep resistance. Consequently, the creep properties of the 20/45 sample were poorer than those of the 20/15 sample after 4 weeks of culturing. For the cell-cultured specimens, the changed structure [Figure 5(e)] improved the ability of the sample to store the fluid phase[Figure 6(b)]; therefore, all of the cultured samples showed better creep resistance properties than the initial ones. Natural articular cartilage itself has excellent creep resistance.[29] In this study, the creep resistance of porous, semidegradable scaffolds improved after the process of chondrocyte seeding and *in* vitro culturing; this indicated that this technology for cartilage repair is promising.

The porous, semidegradable PVA/PLGA scaffold was prepared in this study, and the mechanical properties met our expectations. The composite porogen and PLGA microsphere contents could adjust the compressive properties and creep resistance. All of the cell-cultured samples showed better mechanical properties than the initial ones, so they have potential application after implantation. To understand the results well, we proposed some models to explain the reasons on the basis of some reported literatures. However, to verify the causes of this better, further work should be carried out on the histological analysis and simulation analysis of the microstructure. In a word, we successfully prepared a novel scaffold and proved its potential after implantation.

CONCLUSIONS

In this study, the compressive properties and creep resistance of novel, porous, semidegradable PVA/PLGA scaffolds prepared with various contents of composite porogen and PLGA were observed. In detail, we studied the elastic compressive modulus, the relationship between the compressive modulus and the strain ratio, and the creep resistance properties in certain loading conditions. The differences in these properties between the initial and 4-week cell-cultured scaffold were also explored comprehensively. With ESEM studies, we found that a porous structure with added PLGA microspheres was successfully manufactured. The mechanical properties could be summarized by the following conclusions:

1. Compared with the pure PVA hydrogel, the addition of the composite porogen and PLGA microspheres reduced the elastic compressive modulus. However, *in vitro* culturing for 4 weeks significantly increased the elastic compressive modulus.
2. The compressive modulus of the scaffolds increased exponentially with increasing compression ratio. Through *in vitro* culturing, the compressive modulus increased at nearly all of the strain ratio levels.
3. The increase in the composite porogen content reduced the creep resistance properties, and the same trend was observed after 4 weeks of culturing. The greater PLGA content increased the creep resistance, whereas the inverse result happened after *in vitro* culturing. Moreover, this scaffold was proper for cartilage repair because of the better creep resistance after *in vitro* culturing.

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