Rapidly Self-Expandable Polymeric Stents with a Shape-Memory Property

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A novel biodegradable stent, made of chitosan films cross-linked with an epoxy compound, with a shape-memory property was developed. To reduce their crystallinity, glycerol and poly(ethylene oxide) were blended in the chitosan films. The mechanical properties of the prepared stent were studied using a commercially available metallic stent as a control. After blending, the ductility of the chitosan films was improved, and the compressive strength of the stent was significantly enhanced. The metallic stent could tolerate elastic deformations of 10% before becoming irreversibly deformed, while the polymeric stent was able to withstand deformations up to 30% and still regain its original configuration. The developed stent could rapidly expand (~150 s) from its crimped (temporary) to fully expanded (permanent) states stimulated by hydration, which is advantageous considering avoiding its migration during in vivo deployment. In the preliminary animal study, the implanted stent was found to be intact, and no thrombus formation was seen in the stent-implanted vessel. This degradable stent can be an attractive alternative to metallic stents and may serve as a useful vehicle for local drug delivery.

Introduction

Over the past decade, the use of endoluminal metallic stents during percutaneous transluminal coronary angioplasty has become a common practice for treating coronary arterial stenosis.[1] However, these metallic stents may induce various long-term side effects such as a varying degrees of thrombogenesis and a significant intimal hyperplasia.[2] It was reported that the restenosis rates in patients who receive metallic stents are still 20−40% at 6 months postprocedure, and it rarely occurs thereafter.[3,4] Considering the short-term need and the potential for long-term complications with metallic stents, stents made of biodegradable polymeric materials may be an ideal alternative.[5]

Tamai et al. developed a poly-L-lactic acid (PLLA) stent (the Igaki−Tamai stent) with a self-expandable property. Their preliminary experiment suggested that this stent is feasible and effective in humans.[5] Although it is biodegradable, the self-expansion of the Igaki−Tamai stent has to be achieved by a heated balloon during deployment, which may cause vessel injury.[6] Venkatraman et al. developed a bilayered biodegradable stent composed of PLLA and poly-D-L-lactide-glycolide (PLGA) that can self-expand at 37 °C.[6] The minimum time of full expansion of this bilayered stent in an aqueous environment was about 8 min. To prevent it from migration during deployment, it is desired that the stent be expanded rapidly against the vessel wall after insertion.[7]

In the study, a novel stent made of chitosan-based films that can rapidly self-expand in an aqueous medium with a shape-memory property was developed. It was reported that chitosan is biodegradable, nontoxic, and soft-tissue compatible and has been used extensively for various biomedical applications.[8−11] Additionally, chitosan exhibits a relatively high mechanical strength due to its polysaccharide structure.[12] However, one of the key concerns about the use of chitosan films is their fragile nature.[13] To improve their mechanical properties, glycerol and poly(ethylene glycol) (PEG) [or poly(ethylene oxide) (PEO)] were blended in the chitosan films. The blended chitosan films were cut into strips and wound onto a mandrel and subsequently cross-linked by an epoxy compound (ethylene glycol diglycidyl ether) in an aqueous environment to form the helical stents.

The obtained helical stents possess the ability to memorize their permanent shapes on the mandrel (i.e., a property of shape memory) because of the covalent cross-links formed in the chitosan stripes during fixation by epoxy compounds. Shape-memory materials and stimuli-responsive materials have the capability of changing their shape upon application of an external stimulus.[14,15] In our case, the shape-switching process is controlled by hydration of the cross-linked stents.

The blended chitosan films were characterized by an X-ray diffractometer (XRD) and an Instron material testing machine. Effects of the pH environment during fixation on the cross-linking characteristics of the stents were evaluated. The prepared stents were then studied for their mechanical properties. A commercially available metallic stent was used as a control. Also, the self-expansion time, shape memory, and degradability of the stents were investigated in vitro. Finally, the feasibility of deploying the developed stent in the abdominal aorta of a rabbit model was evaluated.

Experimental Section

Materials. Chitosan (viscosity ≥ 400 mPa·s, 1% in 1% acetic acid at 20 °C, MW ~700 kDa) with a degree of deacetylation of approximately 85% was purchased from Fluka Chemical Co. (Switzerland). PEG (MW 10 kDa) and PEO [MW 200 kDa (PEO200) or MW 400 kDa (PEO400)] were obtained from Sigma-Aldrich (St. Louis, MO). Ethylene glycol diglycidyl ether (Denacol EX-810, epoxy

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compound) was acquired from Nagase Chemicals, Ltd. (Osaka, Japan). All other chemicals and reagents used were of analytical grade.

Preparation of the Blended Chitosan Films. A solution casting method was used in making the blended chitosan films. Chitosan (2% by w/v) was first dissolved in deionized (DI) water containing 1% (v/v) of acetic acid. The jelly-like chitosan solution was dialyzed (MWCO: 12000−14000, Spectrum Laboratories, Inc., Laguna Hills, CA) against DI water for 48 h with water exchanges several times to remove excess acetic acid (final pH ~6.0). Subsequently, a clear chitosan solution was obtained. An aqueous glycerol and PEG (or PEO) solution was then added to the dialyzed chitosan solution and thoroughly blended for 10 min. The composition of the blend was 0.1/0.2/1.0 by weight of glycerol, PEG (or PEO), and chitosan, respectively. The blend was subsequently cast onto a plastic plate and allowed to dry in air at room temperature for 3 days. The obtained film had a uniform thickness of 0.10 ± 0.01 mm.

Characteristics of the Blended Chitosan Films. Film XRD (MXP18, MAC Science, Japan) patterns were obtained to analyze the crystalline characteristics of each blended chitosan film. The diffraction patterns were determined over a range of diffraction angles $2\theta = 5-40°$. To understand how the ductility of chitosan could be improved by blending, an Instron material testing machine (Mini 44, Canton, MA) was employed to perform the tensile test for all the blended chitosan films. The standard dumbbell-shaped specimens were cut from these films. Stress−strain curves of test samples were then determined by uniaxial measurements at a constant speed of 10 mm/min ($n = 5$). The strain-at-fracture value was taken as the percent strain at the point of fracture, while the ultimate-tensile-strength value was taken as the force at which fracture occurred divided by the initial cross-sectional area of the test sample.[16]

Effects of pH on the Cross-Linking Characteristics of Test Stents. The blended chitosan films were cut into strips with a length of 100 mm and a width of 2 mm for fabrication of the helical stents. The strips were wound onto a mandrel (diameter = 3 mm, the ends of the strips were fixed on the mandrel by an adhesive tape) and then chemically cross-linked in the 10% epoxy compound solution at 37 °C for 12 h. To elucidate the effects of pH on the cross-linking characteristics of the stents, the epoxy compound solution was buffered with potassium hydrogen phthalate (0.05 M, pH 4.0), phosphate-buffered saline (PBS, 0.01 M, pH 7.4), or sodium carbonate/sodium bicarbonate (0.21 M/0.02 M, pH 10.5). After fixation, the stents that remained on the mandrel (the permanent shape of the stent in an aqueous environment, outside diameter ~3.3 mm) were rinsed several times with PBS to remove the remaining cross-linking agent and allowed to dry in air. Subsequently, the stents were removed from the mandrel. The degree of cross-linking of the stent, measured by the ninhydrin assay,[17] was defined as the percentage of the free amino groups in the blended chitosan film reacted with epoxy compound subsequent to fixation ($n = 5$).

Mechanical Properties of the Stents. The mechanical properties (the compression load and the collapsed pressure) of the stents with different compositions were evaluated. To study the compression load, test stents were first swollen in PBS and then loaded in a compression platen (Figure 1a) equipped with a miniature force recorder mounted on an Instron material testing machine. The test stent was compressed along its radial direction, and deformation at various loads was recorded ($n = 5$). The time courses of the stent deformation before, during, and after loading were recorded by photography.

The collapsed pressure of the stent was measured using a custom-made pressure chamber similar to that reported by Venkatraman et al.[18] (Figure 1b). The system consisted of a pressure chamber and a flow loop flowing PBS at 37 °C. The test stent was mounted in flexible Tygon tubing connected to the flow loop within the pressure chamber. To simulate the physiological conditions in the human coronary artery, the volumetric flow rate of PBS and the luminal pressure were controlled in the ranges of 100−110 cm^3/min and 110−120 mmHg, respectively.[19]

In the study, compressed air was introduced into the pressure chamber to compress the tubing with or without test stent mounted. At the first observation of tubing collapse, with or without the stent mounted, the pressures on the pressure gauge were read and recorded.

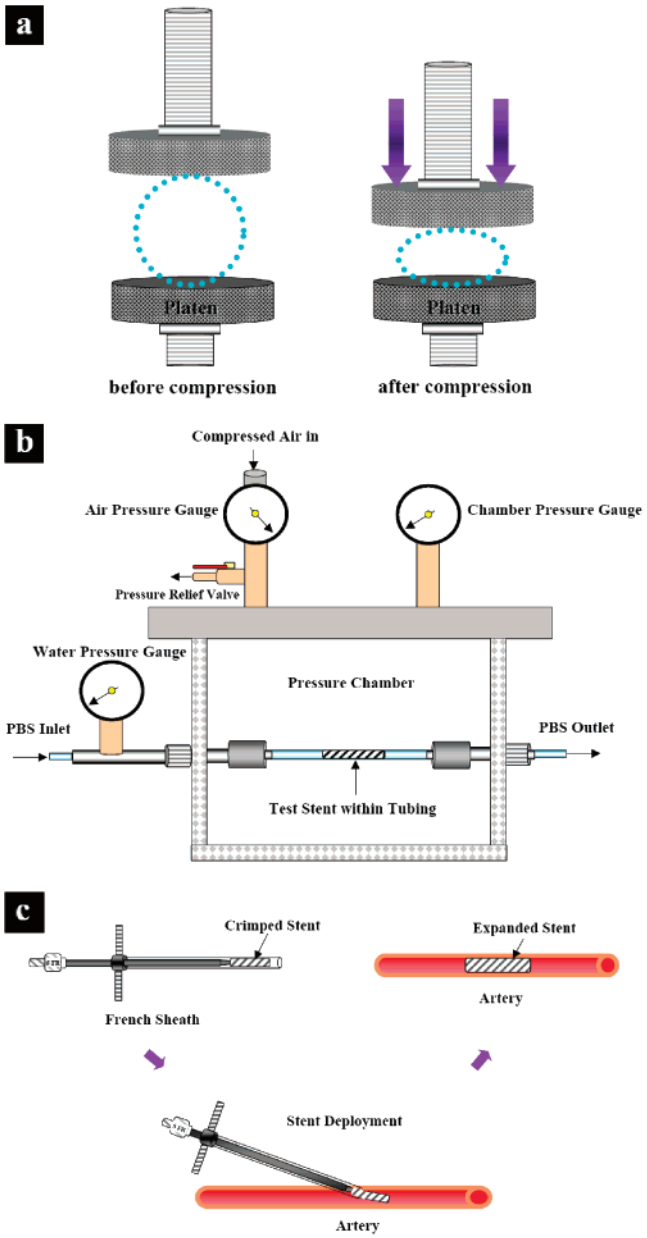

Figure 1. Schematic illustrations of the apparatus used in (a) the compression test and (b) the collapsed pressure test. (c) Schematic illustration of the crimped state of the stent that can be deployed into an artery via a French sheath for the animal study.

The collapsed pressure of the test stent was defined as the pressure difference between the two (with or without the stent mounted, $n = 5$). A commercially available metallic stent (Multi-Link PIXEL, Guidant, St. Paul, MN) was used as a control.

Self-Expansion Time and Shape Memory of Test Stents. The prepared stent was thoroughly soaked in PBS and then allowed to dry and shrink in air (outside diameter ~2.0 mm, the crimped shape of the stent that can be deployed into an artery via a French sheath or a balloon catheter for our animal study, Figure 1c). The time courses of self-expansion of the crimped stent immersed in PBS were observed and recorded by photography. The time taken for the crimped stent to be fully expanded (self-expansion time) in an aqueous environment was measured with a timer ($n = 5$).

In Vitro Enzymatic Degradability of Test Stents. The degradability of the stents with distinct degrees of cross-linking (without cross-linking, 75%, 85%, and 90%) was evaluated in vitro using lysozyme (from chicken egg white with an activity of 50 000 U/mg protein, EC 3.2.1.17, Sigma-Aldrich). Samples of a known weight of test stents were well immersed in a 4 mg/mL lysozyme solution in PBS at pH 7.4 and incubated at 37 °C for 10 weeks. The immersed test samples were taken

out every week ($n = 5$ for each studied group), and the lysozyme solution was replaced with a fresh one at the same time. After enzymatic degradation, the formation of oligomers containing *N*-glucosamine units, due to the cleaved β-glycosidic bonds of chitosan, induced an increment in the free-amino-group content in the incubation medium, which can be determined by the ninhydrin assay.[17] Additionally, test samples were examined by a scanning electron microscope (SEM, model JSM-5600, JEOL Technics, Tokyo, Japan).

Preliminary Animal Study. Animal care and use was performed in compliance with the "Guide for the Care and Use of Laboratory Animals" prepared by the Institute of Laboratory Animal Resources, National Research Council, and published by the National Academy Press (revised in 1996) and approved by the Institutional Review Board (IRB). The feasibility of deploying the developed stent into an artery (abdominal aorta) was evaluated in a rabbit model (New Zealand White rabbit, 2.9−3.3 kg, $n = 8$). To fit the size of the rabbit abdominal aorta (2.2−2.3 mm), test stents were fabricated in a smaller size (the length and outer diameter of the fully expanded stent were approximately 15 mm and 2.4−2.5 mm, respectively). The stent-to-artery diameter ratio at full expansion was about 1.1−1.2 to prevent the migration of the deployed stent. Before deployment, test stents were coated with heparin (55.1 ± 6.2 μg/cm^2, Duraflo, Edwards Life Science, Irvine, CA)[20] to reduce a possible thrombogenicity. The heparinized stents were then sterilized by ethylene oxide.

During the surgical procedure, rabbits were anesthetized with isoflurane inhalant. After systemic heparinization (Heparin LEO, LEO Pharma, Inc., Plantation, FL) with an intravenous dose of 150 IU/kg,[21] a transverse aortotomy was made. Test stents in the crimped state were then deployed into the abdominal aorta of rabbits via a 5.0 French sheath (Figure 1c). Subsequently, the aortotomy was closed with a continuous 8-0 non-resorbable suture (Prolene). Twenty-four hours after the procedure, all animals were euthanized, and the abdominal aorta was removed. The hemocompatibility of the implanted stent and the patency of the artery were grossly examined.

Statistical Analysis. Statistical analysis for the determination of differences in the measured properties between groups was accomplished using one-way analysis of variance and determination of confidence intervals, performed with a computer statistical program (Statistical Analysis System, version 6.08, SAS Institute, Inc., Cary, NC). All data are presented as a mean value with its standard deviation indicated (mean $\pm$ SD). Differences were considered to be statistically significant when the p values were less than 0.05.

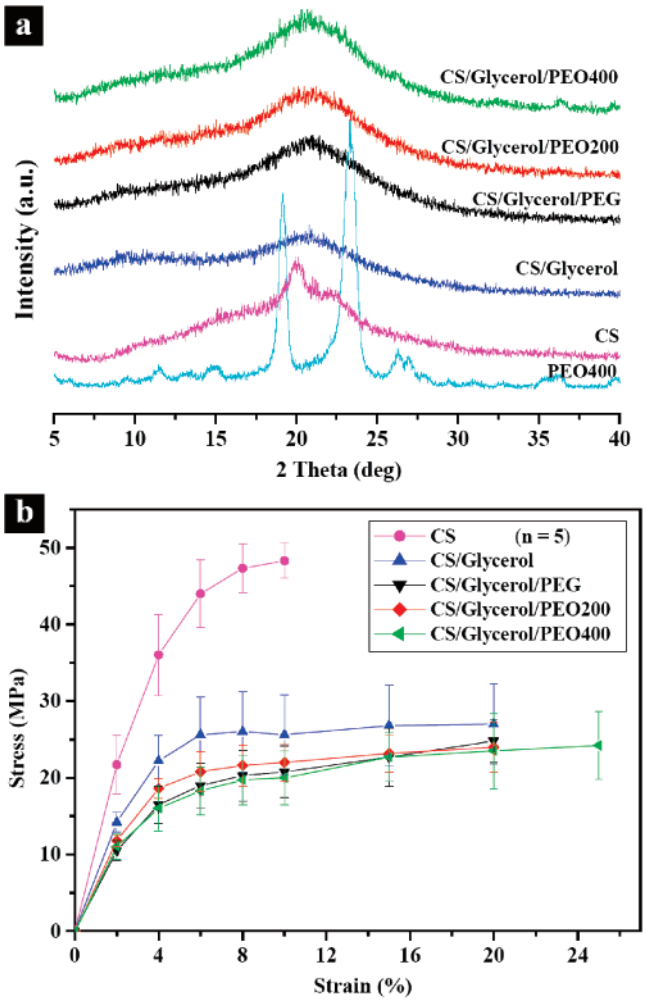

Figure 2. (a) X-ray diffractograms and (b) results of tensile test of test films made of chitosan and those blended with glycerol and PEG or PEO. CS: chitosan; PEG: poly(ethylene glycol), MW 10 kDa; PEO200: poly(ethylene oxide), MW 200 kDa; PEO400: poly(ethylene oxide), MW 400 kDa.

Results and Discussion

Characteristics of the Blended Chitosan Films. Chitosan is an important biomaterial that has been widely used in biomedical applications.[22−25] However, chitosan-made films are generally fragile because of their crystallinity.[13] To reduce their crystallinity, glycerol and PEG (or PEO) were blended in the preparation of chitosan films in the study. Glycerol incorporation is known to reduce the membrane crystallinity.[26] Additionally, Kolhe et al. reported that blending PEG in chitosan is an efficient means to improve its ductility.[13]

Figure 2a shows the diffraction patterns of the films made of PEO400, chitosan, or chitosan blended with glycerol and PEG (or PEO200 or PEO400). As shown, strong reflections were observed in the diffraction patterns for the films made of PEO400 or chitosan, which were associated with their crystalline regions. After blending with glycerol and PEG (or PEO), the reflection in the diffraction pattern for chitosan was significantly diminished and appeared broader, indicating that its crystalline region had become amorphous.

The mechanical properties of the blended chitosan films were characterized and compared with that made of chitosan alone (Figure 2b and Table 1). As shown, the film made of chitosan alone was stiff with an ultimate-tensile-strength value of 48 MPa and a strain-at-fracture value of 10%. It is speculated that materials with a stiff nature would make the stent fabrication very difficult and its in vivo deployment unfeasible. It was found that the blended chitosan films had a significantly lower ultimate tensile strength with a greater strain at fracture, suggesting that the ductility of chitosan was considerably improved after blending. The ultimate-tensile-strength values for the blended chitosan films were all comparable, while the film made of chitosan blended with glycerol and PEO400 (CS/glycerol/PEO400) exhibited a significantly greater strain-at-fracture value or a better ductility than the other test samples.

Effects of pH on the Cross-Linking Characteristics of Test Stents. The pH environment used during the fixation of stents played an important role in affecting their cross-linking characteristics. As shown in Figure 3a, after immersing in the fixation solution, the stents on the mandrel at pH 4.0 and 7.4 became deformed with time. It is known that the pKa value of the amino groups on chitosan is about 6.5.[27] At pH 4.0, a relatively large amount of the amino groups on chitosan were protonated ($-NH_3^+$) and became hydrophilic. Thus, the stent on the mandrel was over swollen in an aqueous environment and therefore was not able to maintain its helical configuration during and after fixation (Figure 3a and 3b).

At pH 7.4, it was found that the stent on the mandrel was swollen to some extent when first immersing in the fixation solution (Figure 3a). This may be attributed to the fact that the buffer at pH 7.4 is not basic enough to neutralize all the protonated amino groups on chitosan immediately. Meanwhile,

Table 1. Ultimate-Tensile-Strength and Strain-at-Fracture Values of Test Films Made of Chitosan and Those Blended with Glycerol and PEG or PEO ($n = 5$)

test film	CS[a]	CS/glycerol[a]	CS/glycerol/ PEG[a]	CS/glycerol/ PEO200[a]	CS/glycerol/ PEO400[a]
ultimate tensile strength (MPa)	48.3 ± 2.3	27.4 ± 5.2	24.8 ± 2.8	24.2 ± 3.5	23.8 ± 4.4
strain at fracture (%)	11.3 ± 1.5	19.8 ± 3.2	20.0 ± 2.7	21.4 ± 3.3	25.3 ± 2.9

[a] CS: chitosan; PEG: poly(ethylene glycol), MW 10 kDa; PEO200: poly(ethylene oxide), MW 200 kDa; PEO400: poly(ethylene oxide), MW 400 kDa.

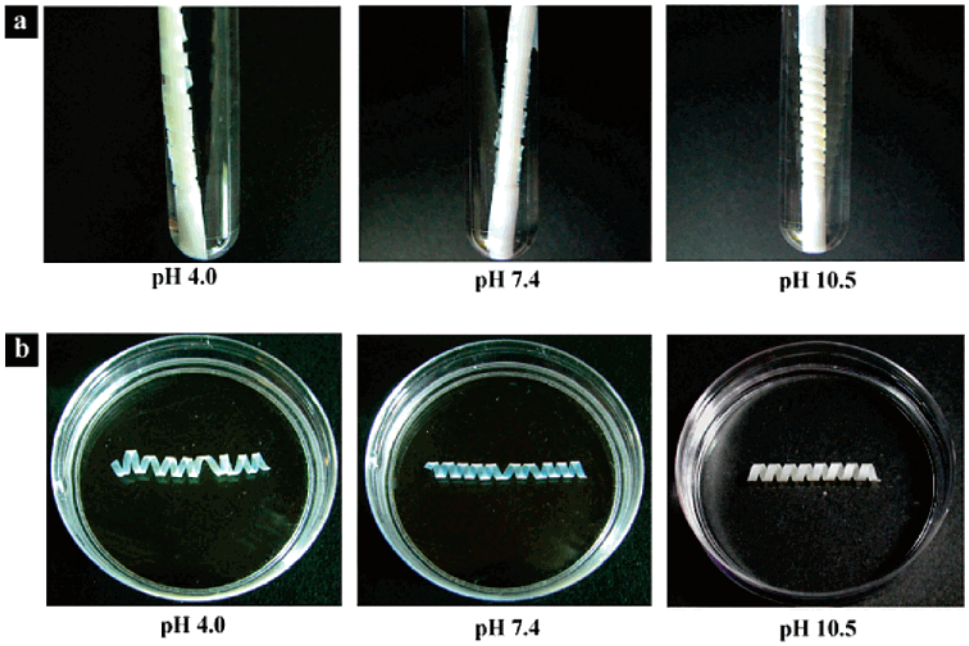

Figure 3. (a) Configurations of the polymeric (CS/glycerol/PEO400) stents on the mandrel during fixation at distinct pHs. (b) Configurations of the stents after fixation; the stents were immersed in a PBS solution.

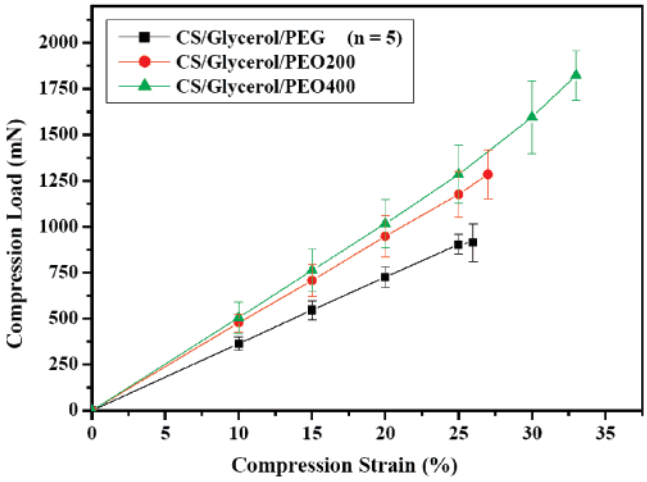

Figure 4. Results obtained in the compression test for the chitosan stents blended with glycerol and PEG or PEO. CS: chitosan; PEG: poly(ethylene glycol), MW 10 kDa; PEO200: poly(ethylene oxide), MW 200 kDa; PEO400: poly(ethylene oxide), MW 400 kDa.

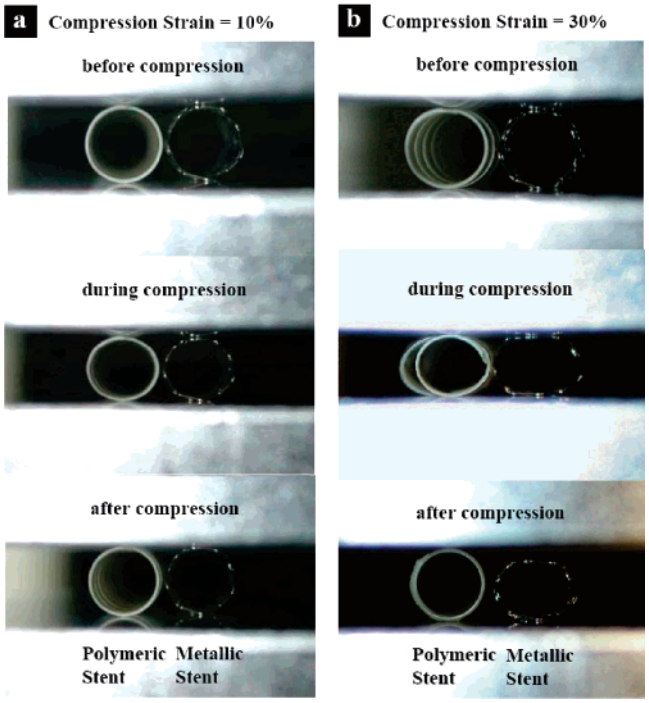

Figure 5. Elastic deformations of the polymeric (CS/glycerol/PEO400) stent and a metallic stent before, during, and after compression: (a) under a compression strain of 10% and (b) under a compression strain of 30%.

the reaction of cross-linking by epoxy compound within the stent was ongoing. Therefore, an undesirable configuration of the stent was observed after fixation (Figure 3b), due to its initial swelling on the mandrel. It was noted that swelling of the stent on the mandrel was not evident at pH 10.5, and it maintained the helical configuration during and after fixation (Figure 3a,b). For that reason, the stent fixed at pH 10.5 was used for the following studies.

After fixation for 12 h, the stents fixed at pH 7.4 (90.3 ± 6.0%) and 10.5 (86.7 ± 7.3%) had a significantly greater degree of cross-linking than that fixed at pH 4.0 (65.1 ± 4.7%). The epoxy compound used in the study may react with the amino groups on chitosan under acidic or basic conditions.[28] However, both the acid- and base-catalyzed reactions require a nucleophilic reagent, the unprotonated amino groups on chitosan. As a result,

Table 2. Collapsed Pressures (Absolute Pressure) of Test Stents Made of Chitosan Blended with Glycerol and PEG or PEO ($n = 5$)

test stents	CS/glycerol/PEG[a]	CS/glycerol/PEO200[a]	CS/glycerol/PEO400[a]	metallic stent
collapsed pressure (bars)	1.78 ± 0.08	1.95 ± 0.10	2.18 ± 0.10	2.47 ± 0.16

[a] CS: chitosan; PEG: poly(ethylene glycol), MW 10 kDa; PEO200: poly(ethylene oxide), MW 200 kDa; PEO400: poly(ethylene oxide), MW 400 kDa.

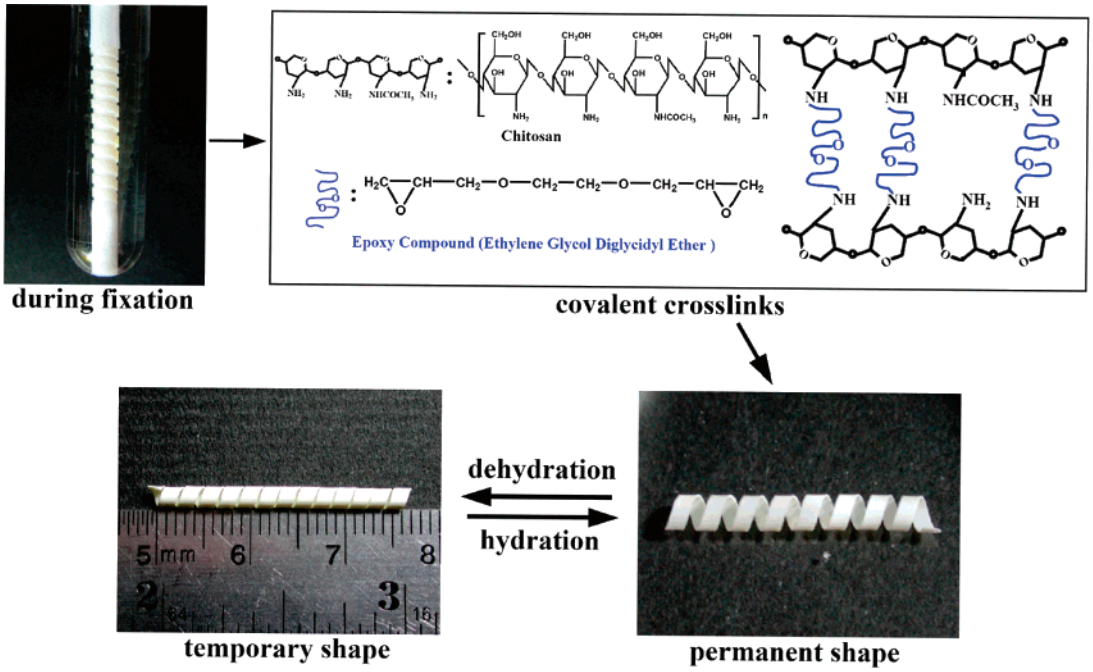

Figure 6. Cross-linking structures and photographs of the permanent and temporary shapes of the polymeric (CS/glycerol/PEO400) stent. The shape-switching process is reversible and controlled by hydration or dehydration of the cross-linked stent.

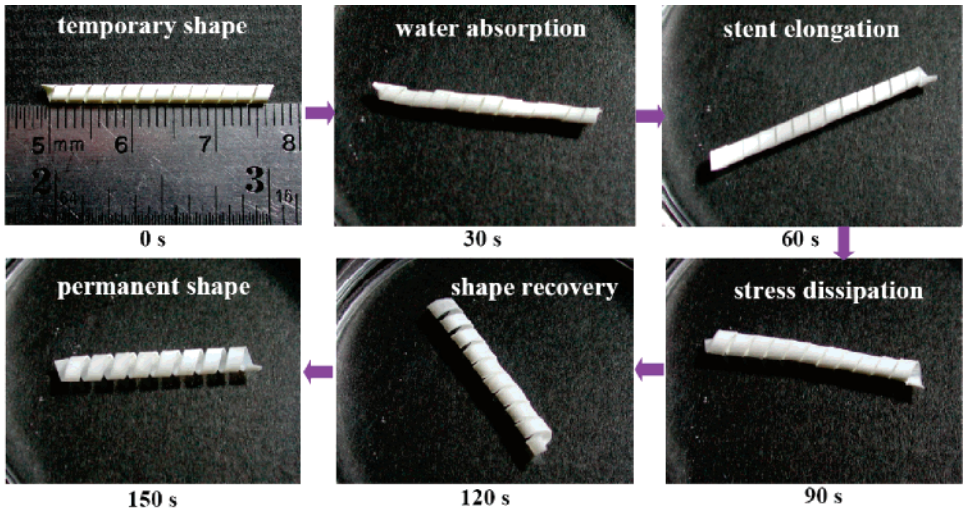

Figure 7. Photographs of the time courses of self-expansion of the polymeric (CS/glycerol/PEO400) stent, immersed in PBS solution, stimulated by hydration.

the degree of cross-linking for the stent fixed at pH 4.0 was the lowest among all studied groups. It was found that the chitosan stent blended with PEG, PEO200 or PEO400 did not significantly affect its configuration and the degree of cross-linking.

Mechanical Properties of the Fixed Stents. The molecular weight of PEG (or PEO) blending in chitosan significantly influenced the mechanical properties of test stents. Figure 4 shows the results obtained in the compression test for the chitosan stents blended with PEG (MW 10 kDa), PEO200 (MW 200 kDa) or PEO400 (MW 400 kDa). A compression test determines the behavior of test stents under crushing loads. It is useful for measurement of the compressive fracture and elastic properties of test stents. As shown, the stent containing PEO400 had the highest compression load (1823 ± 135 mN) and the greatest value of strain-at-fracture (33.0 ± 2.8%) among all studied groups. This indicated that the compressive strength of the stent might be improved by blending a high molecular weight of PEO in chitosan, as a result of a better molecular entanglement during film formation. The elastic deformation of this polymeric (CS/glycerol/PEO400) stent under compression was compared with that of a commercially available metallic stent.

Elastic deformation refers to changes in shape that disappear completely after the release of external forces.[29] When external forces exceed the limit of elastic deformation, test stents become plastically deformed (or fractured), resulting in a permanent

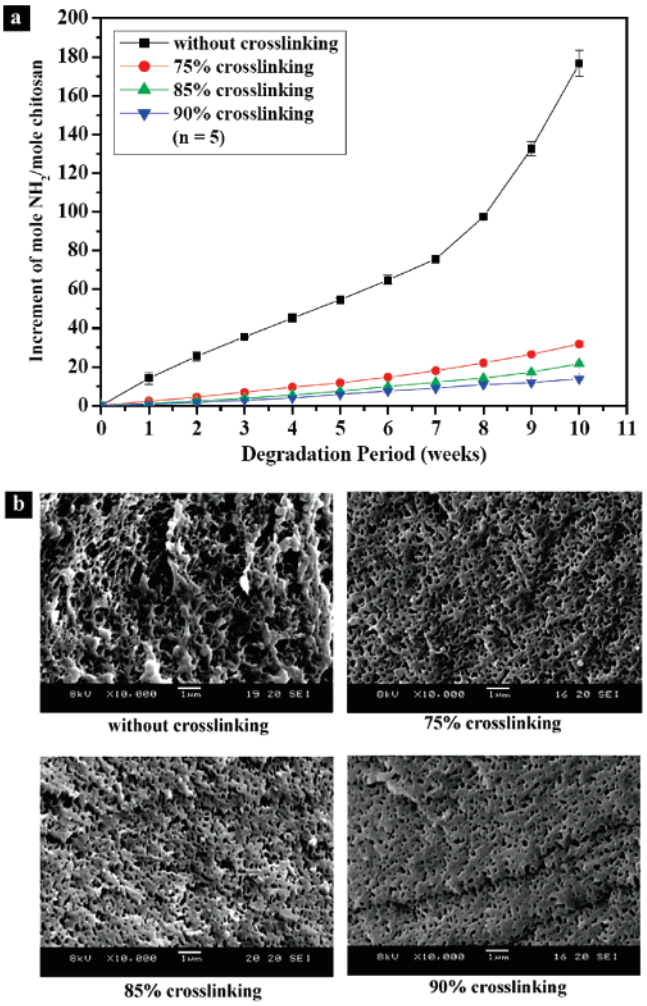

Figure 8. (a) Increments in the free-amino-group content of the polymeric (CS/glycerol/PEO400) stent with distinct degrees of cross-linking after degradation in a lysozyme solution for up to 10 weeks. (b) SEM micrographs of test samples retrieved at 10 weeks after degradation.

change of shape. As shown in Figure 5, the metallic stent could tolerate deformations of 10% before becoming irreversibly deformed. The polymeric stent, however, was able to withstand deformations up to 30% and still regain its original configuration after the release of external forces. It is known that materials with a high degree of elasticity are advantageous in the design of devices that require high kink resistance and preservation of shape.[30]

The collapsed pressures of the polymeric stents blended with distinct molecular weights of PEG (or PEO) were conducted in a flow loop mimicking the physiological flow rate and pressure within a human coronary artery. The results are given in Table 2. It was found that the collapsed pressures (absolute pressure) of the studied polymeric stents were all significantly greater than the physiological pressure within the human body (~1 bar absolute pressure). Among the studied polymeric stents, the stent blended with PEO400 had the greatest collapsed pressure.

The aforementioned results indicated that the CS/glycerol/PEO400 stent had the best mechanical properties among all studied polymeric stents. Therefore, it was chosen for the subsequent studies.

Self-Expansion Time and Shape Memory of the Polymeric Stent. The polymeric stent developed in the study can rapidly self-expand in an aqueous medium with a shape-memory property. Shape-memory materials possess the ability to memorize a permanent shape, which differs from their initial temporary shape.[30] The permanent shape of our stent was fixed on the mandrel by covalent cross-links within the blended chitosan films by means of an epoxy compound (Figure 6). There are a couple of ether bonds (−O−) in the epoxy compound used in the study, which may serve as flexible joints in the cross-linking bridges. Lohre et al. showed that the cytotoxicity of epoxy compounds is significantly lower than that of glutaraldehyde.[31] They suggested that epoxy compound is an acceptable fixative for processing implantable devices, provided that its residual remains below the level found to be cytotoxic.

The chemically cross-linked polymers form insoluble materials that swell in an aqueous environment. As shown in Figure

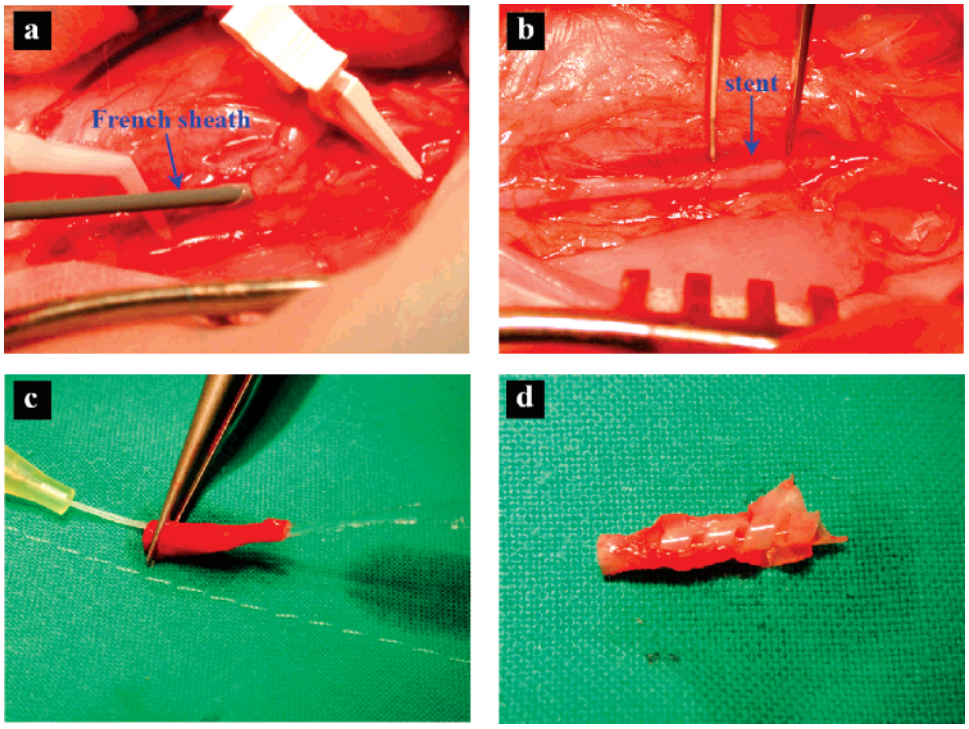

Figure 9. Photographs of a test stent (a) during and (b) after deployment in the rabbit abdominal aorta. Photographs of the stent-implanted vessel retrieved at 24 h postprocedure: (c) patency test; (d) after incision of the aorta.

7, the dried (crimped) stent recovered its permanent shape with time upon hydration to some equilibrium value. After recovering it permanent shape (about 150 s later), the slipping or flow of the polymer chains under hydration was stopped because of the cross-linkage points within the chains. The cross-linkage points act as anchors or "permanent entanglements" and prevent the chains from slipping from each other. Therefore, the transition from the temporary shape of the stent (the dried and crimped stent) to its permanent shape (the expanded stent) can be initiated by an external stimulus: hydration of the stent. Additionally, it was found that this process was reversible (Figure 6). Thus, the developed polymeric stent could potentially be brought into an artery using minimally invasive surgery in a crimped (temporary) shape and then, on demand, be expanded to its permanent shape as required (Figure 1c).

As shown in Figure 7, the self-expansion time of the developed stent from its crimped to permanent shape was 150 ± 10 s (at 37 °C), which was significantly faster than that of polymeric stents made of PLLA (~20 min at 37 °C for the Igaki−Tamai stent)[5] or PLLA/PLGA (~8 min at 37 °C for the stent developed by Venkatraman et al.).[6] The self-expansion of the developed stent is stimulated by hydration, while those made of PLLA or PLLA/PLGA are thermally induced. The raw materials (chitosan and PEO400) used in the fabrication of the developed stent are relatively hydrophilic, and therefore their response to hydration is rapid. A rapid self-expandability of the stent is advantageous, considering avoiding migration of the stent during its in vivo deployment.[7]

In Vitro Enzymatic Degradability of Test Stents. Figure 8a presents the results of degradation of the stents with different degrees of cross-linking in a lysozyme solution for up to 10 weeks. It was reported that the active site of lysozyme consists of six subsites, which bind the *N*-acetylglucosamine residues on chitosan.[32,33] The degradability of each studied group was examined by analyzing the increased *N*-glucosamine units in the incubation medium.[34] As shown, the increments in the free-amino-group content for the cross-linked stents were significantly less than that without cross-linking. This fact was also confirmed by our SEM examination of test samples retrieved at 10 weeks after degradation (Figure 8b). These results indicated that the resistance of the chitosan stent against enzymatic degradation increased considerably, after cross-linking with epoxy compound. This observation may be attributed to the fact that the cleavage sites of chitosan molecules are hidden by the action of cross-linking, resulting in the inhibition of enzyme−substrate interaction.

Preliminary Animal Study. Figure 9a and 9b presents photographs of a test stent during and after deployment in the rabbit abdominal aorta. Twenty-four hours after the procedure, no migration of the implanted stent was observed, and the abdominal aorta was patent for all test animals (i.e., the patency rate was 100%, Figure 9c). After incision of the aorta, the implanted stent was found to be intact. Additionally, no thrombus formation was seen in the stent-implanted vessel (Figure 9d). These results indicated that deployment of the developed stent in an artery using a French sheath was feasible.

Conclusions

In the study, a biodegradable stent, made of blended chitosan films cross-linked with an epoxy compound, with a shape-memory property was developed. The developed stent can rapidly expand from a crimped state to its expanded state, stimulated by hydration in an aqueous environment. The stent could potentially be brought into an artery using minimally invasive surgery. In the preliminary animal study, the implanted stent was found to be intact, and no thrombus formation was observed in the stent-implanted vessel.

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