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Research paper

Polyvinyl alcohol cryogel: Optimizing the parameters of cryogenic treatment using hyperelastic models

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ABSTRACT

The PVA gels obtained by freezing/thawing cycles of PVA solutions, also called cryogels, exhibit non-linear elastic behavior and can mimic, within certain limits, the behavior of biological soft tissues such as arterial tissue. Several authors have investigated the effects of cryogenic processing parameters on the Young's modulus. However, an elastic modulus does not describe the non-linearity of the cryogel's stress–strain response. This study examines the non-linear elastic response of PVA cryogel under uniaxial tension and investigates how processing parameters such as the concentration, the number of thermal cycles, and the thawing rate affect this response. The relationship between the coefficients of the material model and the processing parameters was interpolated to find the set of parameters that would best approximate the elastic response of healthy porcine coronary arteries under uniaxial tension.

1. Introduction

Poly(vinyl alcohol) (PVA) cryogels have been studied for biomedical applications such as cartilage replacement (Kobayashi et al., 2005; Pan et al., 2007; Stammen et al., 2001; Swieszkowski et al., 2006), nucleus pulposus replacement (Allen et al., 2004; Joshi et al., 2006) and cardiac valve prostheses (Jiang et al., 2004) but also as a material for magnetic resonance and ultrasound imaging phantoms (Chu and Rutt, 1997; Surry et al., 2004).

PVA cryogels are produced by freeze/thaw processing of concentrated PVA solutions of water or dimethyl sulfoxide (DMSO). When using dimethyl sulfoxide solutions, the resulting gels become transparent (Hyon et al., 1989; Trieu and Qutubuddin, 1995). Lozinski et al. presented a series of papers (Lozinsky et al., 2008, 2007; Lozinsky and Damshkaln, 2000; Lozinsky et al., 1995, 1986, 2000) on the cryostructuration of PVA cryogels and detailed reviews (Lozinsky, 1998, 2002). The ability to modify the cryogel strength has been investigated by several authors for more than 20 years (Chu and Rutt, 1997; Domotenko et al., 1988; Lozinsky and Damshkaln, 2000; Lozinsky et al., 2000; Nagura et al., 1989; Peppas and Scott, 1992; Ricciardi et al., 2005; Stammen et al., 2001; Stauffer and Peppas, 1992; Watase and Nishinari, 1985b, 1988). The studies have shown that the properties of the cryogel

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depended on the PVA grade (molecular weight, degree of saponification), the polymer concentration (Ku et al., 1999) in the initial solution, the presence of additives and conditions of cryogenic treatment: temperature and duration of freezing, thawing rates (Lozinsky and Damshkaln, 2000; Wan et al., 2002) and number of cycles (Stauffer and Peppas, 1992). Watase et al. (Watase and Nishinari, 1985a,b, 1988, 1989; Watase et al., 1983) investigated the effect of freezing on the Young's modulus and concluded that increased elastic modulus resulted from increased crystallinity in the PVA hydrogel. Peppas and Stauffer (Peppas and Stauffer, 1991) discussed how repeated thermal cycling of PVA hydrogels can increase the elastic modulus.

However, when designing test specimens as substitutes for biological soft tissues, the processing parameters need to be optimized in order to obtain the desired response. The purpose of this paper is to model the elastic behavior of PVA cryogel under uniaxial tension, at moderate strain, with common "rubber-like" constitutive models used to simulate the behavior of biological soft tissues (Bedoya et al., 2006; Chau et al., 2004; Gerard et al., 2005; Hayashi and Imai, 1997; Kauer et al., 2002; Lally et al., 2005; Wu et al., 2007), and to propose a quantitative method to optimize the gel processing parameters to obtain the closest desired stress-response-to-stretch curve.

2. Material and methods

2.1. Gel preparation

All experiments used commercial grade PVA that was fully hydrolyzed and hot water soluble (P1763, Sigma). The viscosity average molecular weight ($\bar{M}_v$) was calculated using the equation (Beresniewicz, 1959):

$$[\eta] = 5.95 \times 10^{-4} \bar{M}_v^{0.63} \quad (1)$$

where the intrinsic viscosity $[\eta]$ is in dl g^{-1}. The viscosities of aqueous PVA solutions were measured using a rheometer (Bohlin Rheometer CVO120, Malvern Instruments Ltd) with double gap. The average viscosity molecular weight of the polymer was estimated to be 83 ± 5 kDa.

PVA solutions were prepared by mixing PVA powder in tap water at room temperature. The mixes were heated until entire dissolution of the powder. The container was kept sealed during the process to avoid drying. The resulting clear solution was cooled to room temperature, then slowly poured in a mold that consists of rectangular cavities in an aluminum (2024O grade) block. An air gap was maintained above the solution to allow for expansion during freezing. The mold was placed on a programmable chilling/heating plate (Peltier plate, IC25XT, TorreyPines Scientific). PVA cryogel samples were obtained by subjecting the polymer aqueous solutions to repeated thermal cycles. The specific processing parameters used for the study are described in Section 2.4.2. When completed, the samples were stored in water at ambient temperature until testing. During the thermal cycling, a temperature probe (Teflon insulated T probes with Teflon-coated junction) was placed in a dedicated hole of the aluminum block and another was plunged in a small volume of gel. The temperatures were recorded every ten seconds.

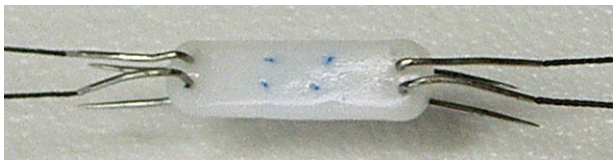

Fig. 1 – Tensile sample.

2.2. Coronary arteries

The heart of a healthy Yorkshire pig was harvested. The left and right coronary arteries were dissected. Any visible connective tissue was removed from the surface of the vessel before it was cut along its length, flattened gently, and sectioned to form rectangular strips in the longitudinal direction of the artery for uniaxial testing.

2.3. Mechanical testing

The gel and artery specimens were tested in uniaxial tension, using a mechanical tester (Enduratec ELF 3200, Bose) with a 1 kg load cell (Model 31, Sensotec Honeywell) and a black-and-white CCD-camera video-extensometer (Watec America LCL 902C camera with Computar TEC55 Lens) that allows automatic gauge mark tracking. The frequency of displacement and load data acquisition were set to 12.5 Hz and the video acquisition had a rate of 10 frames /s.

Before testing, the specimen dimensions were measured with a digital calliper (Litematic VL50A, Mitutoyo). Four markers were plotted with dye on the sample surface to track strain with video-extensometer. The samples were attached with curved needles and 2-0 polyester braided sutures at two insertion points, at each extremity (cf. Fig. 1). We selected a suture method rather than a clamping method because with the gripping method, the gel samples slid during stretching. As we immersed the specimen, emery paper was not an option and glue did not work on the gel. A template was used to limit the variability of insertion point location. The samples were preloaded to slightly tense the sutures. Preconditioning was achieved by executing six successive loading–unloading cycles of 4 mm displacement. The samples then underwent one maximal cyclic extension of 12 mm at a constant displacement rate of 0.1 mm/s. During the tests, the gel specimens were immersed in water at ambient temperature and the arterial specimens were immersed in a KREBS solution at 37 °C. Preliminary tests showed no significant difference for test at ambient temperature and at 37 °C with the gel. Watase and Nishinari (Watase and Nishinari, 1985b) also found that the dynamic Young's modulus of gels of PVA with high degree of saponification were independent of temperature up to 40 °C.

2.4. Data analysis

All the analyses were performed with the final extension data. The maximal stretch ratio ranged from 1.65 to 1.75. For consistency, and because a material is best characterized using larger deformation, all datasets were analyzed for

stretch up to 1.65. The arterial deformation between diastole and systole might be about 10% but it does not correspond to 1.1 stretch as the stretch is computed with respect to the load-free configuration, not to diastole. For example, Zhou and Fung reported, for canine thoracic aortas, that, in the physiological range the longitudinal stretch lay between 1.3–1.5 and the circumferential stretch lay between 1.4–1.8. In order to assess the effects of gel processing parameters, we first modeled the stress–stretch behavior then analyzed the model coefficients. In mechanical modeling of biological soft tissues, 'rubber-like' material models such as the Rivlin model are widely used.

Mooney and Rivlin proposed a strain energy function W as an infinite series in powers of $(I_1 - 3)$ and $(I_2 - 3)$ of the form:

$$W = \sum_{\substack{i+j=1 \\ i,j=0,\ldots,n}}^{n} C_{ij} (I_1 - 3)^i (I_2 - 3)^j \tag{2}$$

where C_{ij} are material constants, n is the order and I_1, I_2 the first and second invariants of the right Cauchy–Green deformation tensor C

$$\begin{aligned} I_1 &= \text{Trace}(C) = \lambda_1^2 + \lambda_2^2 + \lambda_3^2 \\ I_2 &= (\text{Trace}(C^2)\text{–Trace}(C)^2)/2 = \lambda_1^2\lambda_2^2 + \lambda_2^2\lambda_3^2 + \lambda_3^2\lambda_1^2 \end{aligned} \tag{3}$$

where $\lambda_{i=1,2,3}$ are the principal stretches. When all the C_{ij} are positive, the fundamental requirement of an always positive strain energy function in all ranges of deformation is fulfilled and the relation between stress and deformation is monotonic, leading to physically plausible solution (Hartmann, 2001). In the specific case where only the terms in I_1 ($j = 0$) are considered, the strain energy function is equivalent to the Yeoh model that generally provides a good fit for elastomers.

Assuming incompressibility ($\lambda_1\lambda_2\lambda_3 = 1$), the strain energy can be expressed as a function $\hat{W}$ of λ_1 and λ_2, or I_1 and I_2. The difference of Cauchy stresses in principal directions can be expressed as:

$$s_1 - s_3 = 2\frac{\partial \hat{W}}{\partial I_1}\left(\lambda_1^2 - \lambda_1^{-2}\lambda_2^{-2}\right) - 2\frac{\partial \hat{W}}{\partial I_2}\left(\lambda_1^{-2} - \lambda_1^2\lambda_2^2\right). \tag{4}$$

For uniaxial tension ($s_3 = 0$, $\lambda_1 = \lambda$, $\lambda_2 = \lambda^{-1/2}$), the stress expression becomes:

$$s_1 = 2\left(\lambda^2 - \lambda^{-1}\right)\left(\frac{\partial \hat{W}}{\partial I_1} + \lambda^{-1}\frac{\partial \hat{W}}{\partial I_2}\right). \tag{5}$$

2.4.1. *Optimization method*

To find values for the coefficients $p = [p_1 \ldots p_k]^T$ such that the model matches the experimental data as well as possible, the objective function to be minimized was defined as the squared 2-norm of the weighted residuals. Hence, the minimization problem was given by:

$$\min_p \chi^2 = \sum_i \left(\frac{\tilde{s}(\lambda_i, p) - s_i}{\sigma_i}\right)^2 \tag{6}$$

where (λ_i, s_i) is a set of stretch and Cauchy stress raw data, $\tilde{s}(\lambda_i, p)$ is the fitted value for the point λ_i and σ_i is an estimate of the typical error of measurement (*TEM*) for the Cauchy stress values s_i.

To estimate the typical error of measurement (due to experimental error and technological error), nine samples made with 10% in weight (wt%) PVA solution and submitted to nine thermal cycles at a thawing rate of 0.083 °C/min and freezing rate of 0.333 °C/min were tested in uniaxial tension. The typical error of measurement was computed as the standard deviation of the Cauchy stress values at uniformly spaced stretch values, for the nine repeated tests.

The problem was solved by the built-in Levenberg–Marquardt nonlinear least-square optimization algorithm in IGOR Pro (v6.01, WaveMetrics Inc.). The iterations stop either when the fractional decrease of χ^2 from one iteration to the next is less than 10^{-3} or when 40 iterations have been performed without converging.

To further analyze the effect of the processing parameters, we selected the best fit. As a measure of the goodness of fit, we observed the residuals and computed the root mean squared error, the adjusted coefficient of determination (R^2), and the 95% confidence bounds for the fitted parameters.

The Root Mean Squared Error (RMSE) is:

$$\text{RMSE} = \sqrt{\frac{\sum_{i=1}^{n}\left(\tilde{s}(\lambda_i, p) - s_i\right)^2}{n - k}} \tag{7}$$

where n is the number of observation points and k the number of fitted coefficients.

The coefficient of determination measures how successful the fit is in explaining the variation of the data. The adjusted coefficient of determination takes into account the residual degrees of freedom.

$$R^2 = 1 - \frac{\text{SSE}}{\text{SST}} \tag{8}$$

$$Adjusted\_R^2 = 1 - \frac{(n-1)\,\text{SSE}}{(n-k)\,\text{SST}} \tag{9}$$

where *SSE* is the sum of squared (non-weighted) residuals and SST/(n-1) the total variance.

The uncertainties in the coefficient estimates depend on the model. The confidence bounds for the fitted parameters are given by

$$cb = p \pm t_{(1-\alpha, n-k)}S \tag{10}$$

where p is the vector of the parameter estimates, t is the inverse of the Student's T cumulative distribution function and S the vector of the standard deviation of the parameter estimates (Rawlings et al., 1989).

2.4.2. *Analysis*

When molding small parts like the non-scaled artery model (Pazos et al., 2007), a lower viscosity solution is easier to handle and reduces air bubble entrapment. For all the prepared specimens, the initial PVA concentration was of 10 wt%, the minimal temperature was set to −20 °C and the maximal to +20 °C. Four series of tests were conducted:

- The first series consisted of nine samples produced and tested in the same conditions, five thermal cycles with a freezing rate set to 0.333 °C/min and a thawing rate set 0.083 °C/min, to estimate the typical error of measurement.

- The second series aimed to assess of the effect of freezing and thawing rates and consisted of four samples (2 factors with 2 values) processed with 5 thermal cycles processed with a freezing rate set to either −0.333 °C/min or −0.111 °C/min and a thawing rate set to either +0.333 °C/min, or +0.111 °C/min.
- The third and fourth series aimed to establish a relationship between the coefficients of the strain energy function and the gel processing parameters. The third series consisted of ten samples processed varying only one factor, the number of cycles, ranging from 2 to 13, with a freezing rate set to −0.333 °C/min and a thawing rate set to +0.111 °C/min.
- The fourth series consisted of twelve samples (one factor with 3 values and one factor with 4 values) processed with 2, 3 and 5 thermal cycles processed and a thawing rate of either +0.333 °C/min, +0.111 °C/min, +0.083 °C/min or +0.067 °C/min, the freezing rate being set to −0.333 °C/min.

Once a relationship was defined between the material coefficients and the gel processing parameters, the relation was applied to find the processing parameters that would minimize the sum of the squared differences between the modeled gel response and the curve that fitted the stress–stretch data of the coronary artery samples. The optimization method was the same as described before, except that the residuals were computed between two modeled curves instead of between the fitted curve and experimental data and that no weighting was applied.

3. Results

3.1. General

Fig. 2 illustrates a typical curve temperature obtained with a probe placed in PVA solution/gel during the gelation process. For set temperatures ranging from −20 °C to +20 °C, the temperatures measured in the mold were down to −16 °C and up to 23 °C. The effective thawing rates measured in the mold were 0.322 °C/min, 0.112 °C/min, 0.081 °C/min and 0.066 °C/min for rate set, respectively at 0.333 °C/min, 0.111 °C/min, 0.083 °C/min and 0.067 °C/min, confirming the efficiency of the temperature control.

Two phenomena were systematically observable on the temperature curves: the first, occurring during freezing, was a sudden temperature increase from about −8 °C to −1 °C then freezing resumed; the second, occurring during thawing, was a small slowing down of temperature increase around 0 °C. These observations concurred with the typical thermogram of cryostructure formation of the water–PVA system presented in the literature (Domotenko et al., 1988) where the first observation corresponds to crystallization and the second to ice melting.

Gels produced after one thermal cycle, or when the sample temperature did not reach about −10 °C due to technical error, fell under their own weight.

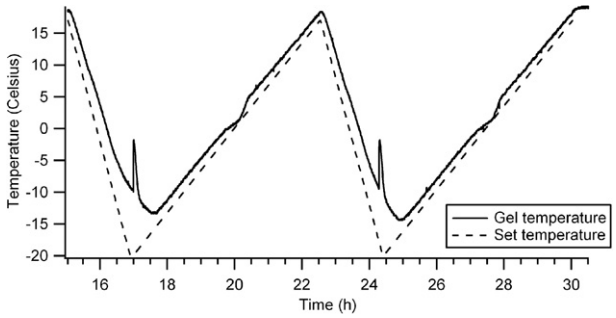

Fig. 2 – Temperature measured within a PVA sample during thermal cycles.

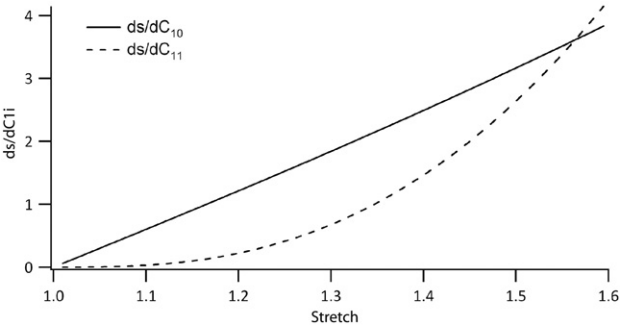

Fig. 3 – Contribution of C10 and C11 coefficients in the model for stretching between 1 and 1.6.

3.2. Data analysis

Based on uniaxial tension measurements of nine samples made with 10 wt% PVA solution submitted to nine thermal cycles at a thawing rate of 0.083 °C/min and freezing rate of 0.333 °C/min, the typical error of measurement (*TEM*) was estimated at 0.6 kPa + 3.5% of the Cauchy stress value. The weighting of the residuals for the curve fitting was defined based on this estimation.

When considering the Cauchy stress response to uniaxial stretch ranging between 1.0 and 1.6 of gel specimens, the strain energy density function that provided the best fit (coefficient of determination of 0.99) was:

$$\hat{W} = C_{10}(I_1 - 3) + C_{11}(I_1 - 3)(I_2 - 3). \quad (11)$$

The average RMSE over all the datasets were 1.2 kPa. Adding terms in the strain energy density function resulted in over-parameterization, and consequently, one or several parameters with large standard deviation. For this model, Fig. 3 presents the contribution of each coefficient to the stress response ($\frac{\partial s}{\partial C_{1i}}$ for $i = 0, 1$). For lower stretching, the contribution of C_{10} is predominant and the initial tangent modulus is proportional to C_{10}. For $\lambda \geq 1.56$, the contribution of C_{11} becomes the most important.

For the nine datasets of the first series, used to estimate the typical error of measurement, the optimized coefficients, based on weighted χ^2, were of 8.7 ± 0.9 kPa (mean ± standard deviation) for C_{10} and 18.9 ± 1.5 kPa for C_{11}, which correspond to variations of 10% and 8% respectively.

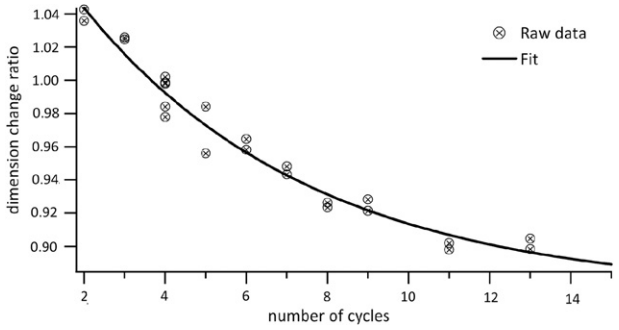

Fig. 4 – Dimension change ratio as a function of the number of thermal cycles.

The second series confirmed that the freezing rate had no significant effect; the difference was within the typical error of measurement.

The third series included ten gel samples obtained after 2 to 13 thermal cycles. The sample length and width changed, relatively to the mold size, with the number of cycles as shown in Fig. 4. The length and width of the samples shrank up to 10%. The gel stress response to stretching increased with the number of cycles (Fig. 5(a)). Considering the experimental error, the observed differences became non-significant after 11 thermal cycles. Similarly, the model coefficients increased, non-linearly, with the number of thermal cycles (cf. Fig. 5(b)).

The effect of thawing rate was studied for 2, 3, 5 and 13 thermal cycles. As illustrated in Fig. 6, slowing down the thawing rate increased the gel resistance to tension. It can be noticed that processing the gel with five thermal cycles at a thawing rate of 0.333 °C/min or with four thermal cycles at a thawing rate of 0.111 °C/min resulted in similar gels in terms of stress response to moderate uniaxial stretch. However, the processing time was shorter in the first case.

The relationship between the coefficients and the number of thermal cycles and the thawing rate was computed as:

$$C_{1j}(x,y) = c + A\left(1 - e^{-\tau x}\right) + B_1 y + B_2 y^2 \quad \text{for } j = 0, 1 \qquad (12)$$

where x, y are the normalized number of cycles and inverse thawing rate. For $j = 0$, $c = 0.24$ kPa, $A = 16.3$ kPa, $\tau = 1.248$, $B_1 = 4.67$ kPa, $B_2 = 0$, RMSE = 0.72 kPa, $R^2 = 0.949$ and adj$R^2 = 0.936$. For $j = 1$, $c = 0$, $A = 42.9$ kPa, $\tau = 1.286$, $B_1 = 0$, $B_2 = 8.99$ kPa, RMSE = 1.81 kPa, $R^2 = 0.999$ and adj$R^2 = 0.941$. The corresponding fitting curves for 2 to 5 cycles are shown Fig. 6(a, b).

The proposed fitting function (Eq. (12)) was used to estimate the optimum preparation parameters to obtain a gel that would mimic coronary arteries, with the optimum defined as the minimum difference between the modeled stress responses to uniaxial stretch. The estimated material coefficients (Eq. (11)) for the uniaxial tension stress–stretch data of porcine coronary arteries and the estimated coefficients for the closest mimicking gel are reported in Table 1 and showed in Fig. 7. The corresponding optimized gel processing parameters were six thermal cycles at a thawing rate of 0.082 °C/min and five cycles at a thawing rate of 0.093 °C/min, respectively. For the average curve, the parameters were estimated to 6 cycles with a rate thawing of 0.101 °C/min, for a root-mean squared difference of 4.5 kPa. The maximum variation in stress value was of 20%.

4. Discussion

Despite their limitations, "rubber-like" material models are commonly used in simulations to model the response of biological soft tissues. In this context, PVA cryogel appears as a good material to make test specimens with controlled properties. The cryogel has the advantages to be relatively easy to process and to form into relatively complex geometry.

During the freezing process, we observed a heat emission peak, occurring around −10 °C ± 2 °C for 10% to 15% PVA solution, similarly at every cycle. It corresponds to the crystallization of the solvent (Lozinsky, 1998) which results in increased solute concentration in unfrozen regions of the specimen, promoting interactions of the PVA chains, favoring intermolecular hydrogen bonding (Damshkaln et al., 1999; Domotenko et al., 1988; Lozinsky, 1998). When the gel did not reach this critical range of temperature, the resulting samples fell under their own weights and were translucid. The same results were obtained with gels produced by placing the mold at −20 °C during three days and with very

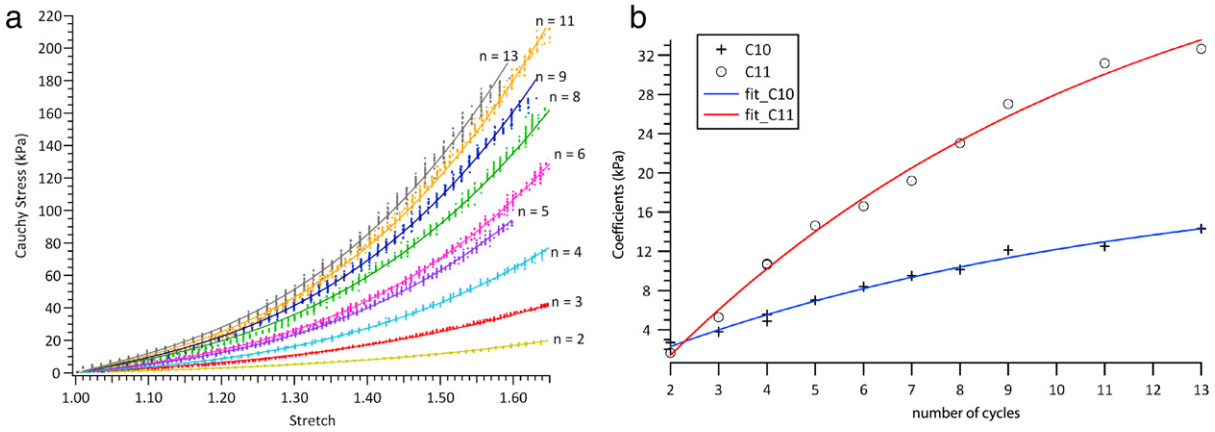

Fig. 5 – Effect of number of thermal cycles for 12% PVA solution and thawing rate of 1 °C/9 min (a) Stress–stretch data (b) Model coefficients.

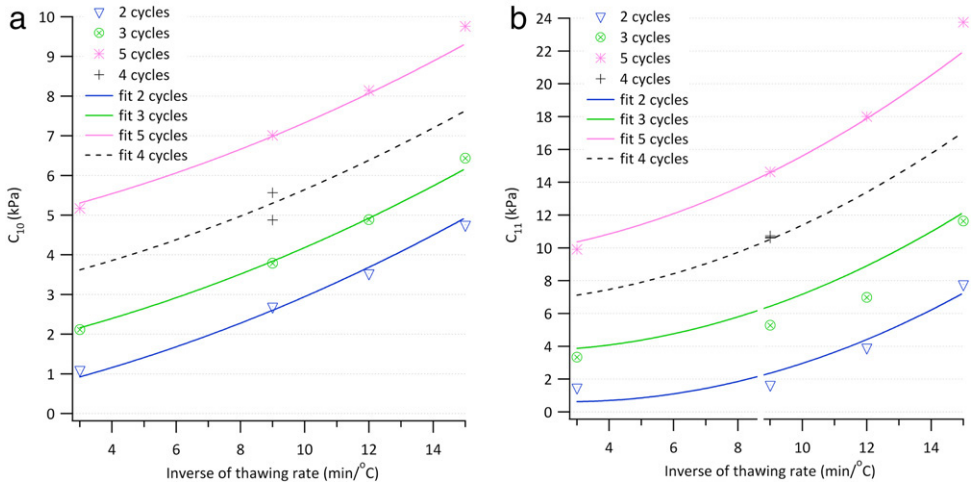

Fig. 6 – Effect of the thawing rate on the model coefficients, for 2 to 5 thermal cycles.

Table 1 – Material coefficients (95% confidence interval) for porcine coronary arteries and for the corresponding mimicking gel (RMSD: root mean squared difference between the modeled stress–stretch curves for the arterial tissue and for the gel).

	C_{10} (kPa)	C_{11} (kPa)	RMSE (kPa)	Adjusted R^2	Gel C_{10} (kPa)	Gel C_{11} (kPa)	RMSD (kPa)
Left artery	16.7 ± 0.1	12.7 ± 0.1	2.3	0.995	9.8	21.3	5.2
Right artery	13.0 ± 0.1	10.3 ± 0.1	2.9	0.988	8.0	16.5	3.8

rapid thawing, indicating that the formation of the cryogel takes place at the thawing stage (Lozinsky, 2002). During the melting of ice, the mobility of macromolecules and the efficiency of the intermolecular interactions increase. The slower the thawing, the longer the period the specimen is at the temperatures optimal for gel network formation (Lozinsky, 2002). No significant difference was observed when the gel was processed with a maximal set temperature of +10 °C, instead of +20 °C.

The number of intermolecular hydrogen bondings increases and molecular chains aggregate with increasing number of cycles (Hatakeyema et al., 2005) resulting in the formation of the strongest cryogels. The first coefficient of the model, C_{10}, is proportional to the initial Young's modulus. As reported in previous studies, the initial elastic modulus increased as the number of thermal cycles increased and when the thawing was slowed down. The second coefficient of the model, C_{11}, that contributes more to the non-linearity of the model, also increased as the number of thermal cycles increased and when the thawing was slowed down. The difference was that the thawing rate had a quadratic effect on C_{11}, whereas it had a linear effect on C_{10}. While the coefficients seemed to reach a plateau for higher numbers of cycles, the results suggest that they would continue to increase with slower thawing rates.

The selection criterion for the concentration of the PVA solution was the qualitative viscosity of the PVA solution and 10% appeared to be the most suitable for molding of small thickness. Our study results show that varying the number of cycles from to 2 to 5 or the thawing rate from 0.333 °C/min to 0.067 °C/min had about the same effect on the gel's elastic properties. However, the thawing rate variation provides a

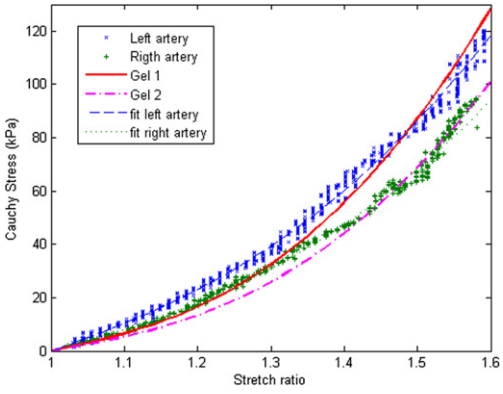

Fig. 7 – Uniaxial tension data for the arteries and gels presented in Table 1.

finer control of the desired elastic properties as the only constraint on the set value is related to the equipment used to freeze and thaw the gel. As shown in Fig. 6, it can process the gel with five thermal cycles at a thawing rate of 0.333 °C/min or with four thermal cycles at a thawing rate of 0.111 °C/min resulted in similar responses to moderate uniaxial stretch but the processing time was shorter with the former approach.

To define a specific quantitative selection criterion for the number of thermal cycles and thawing rate, we found it convenient to model the changes in the hyperelastic coefficients as a function of these two parameters using sets of experimental data to interpolate the coefficient estimates for any parameter values within the defined range. Our objective criterion was to minimize the squared differences between the estimated gel elastic response and

the target response obtained for uniaxial tensile testing of the tissue to be mimicked. If the likelihood of the curve shape were more important than the stress amplitude, then the objective criterion would have been to minimize the squared differences between the model coefficients rather that the resulting curves. The study was based on uniaxial tensile testing to be consistent with the data that were possible to obtain considering the size of the porcine coronary arteries. With the proposed method, we estimated that a gel processed with six thermal cycles and a thawing rate of 0.101 °C/min could approximate the uniaxial elastic response obtained for healthy porcine coronary arteries, for moderate strain.

The sensitivity to variations in processing parameters has some advantages when using the gel to create test specimens as a substitute for biological soft tissues, where sample-to-sample variations and changes in the elastic properties, within a certain range, could be desired.

5. Conclusions

- We presented a method to optimize cryogenic treatment parameters of PVA cryogels in order to obtain a desired non-linear elastic response.
- A Mooney–Rivlin type strain energy density function, with only C_{10} and C_{11} non-null coefficients, provided a good fit (coefficient of determination of 0.99) for uniaxial tension of PVA cryogels as well as for porcine coronary arteries.
- The number of cryogenic cycles had a similar effect on both material model coefficients. The second coefficient was relatively more sensitive to the thawing rate than the first one.

Acknowledgments

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