Bending and pressurization test of the human aortic arch: Experiments, modelling and simulation of a patient-specific case

Claudio M. García-Herrera\textsuperscript{a}\textsuperscript{*}, Diego J. Celentano\textsuperscript{b}, Marcela A. Cruchaga\textsuperscript{a}

\textsuperscript{a}: Departamento de Ingeniería Mecánica, Universidad de Santiago de Chile, USACH
Av. Bernardo O’Higgins 3363, Santiago de Chile, Chile
\textsuperscript{b}: Departamento de Ingeniería Mecánica y Metalúrgica, Pontificia Universidad Católica de Chile
Av. Vicuña Mackenna 4860, Santiago de Chile, Chile

Abstract

This work presents experiments, modelling and simulation aimed at describing the mechanical behaviour of the human aortic arch during the bending and pressurization test. The main motivation is to describe the material response of this artery when it is subjected to large quasi-static deformations in three different stages: bending, axial stretching and internal pressurization. The sample corresponds to a young artery without cardiovascular pathologies and the pressure levels are within the normal and hypertension physiological ranges. Two are the principal findings of this work: firstly, the material characterization performed via tensile test measurements that serve to derive the material parameters of a hyperelastic isotropic constitutive model and, secondly, the assessment of these material parameters in the simulation of the bending and pressurization test. Overall, the reported material characterization was found to provide a realistic description of the mechanical behaviour of the aortic arch under severe complex loading conditions considered in the bending and pressurization test.

1 INTRODUCTION

Clinical reports on complex anomalies of the human aortic arch have been extensively published in the past (e.g., double aortic arch and right aortic arch with left ductus/ligamentum arteriosus; see Kocis et al. (1997) for a complete review on this subject). Other relevant anomalies closely related to the human aortic arch have been recently studied, e.g., the so-called bovine aortic arch (Bizzarri et al., 2008a) and the aberrant right subclavian artery (Tochii et al., 2008). On the other hand, some of the genetical pathologies usually deriving in aneurysms that affect the mechanical response of the aortic arch are the Marfan (Nollen et al., 2004), DiGeorge (Momma et al., 1999) and MAGIC (Caceres et al., 2006) syndromes.

\textsuperscript{*}Corresponding author: claudio.garcia@usach.cl
Surgical repairs for aortic arch aneurysms have been carried out in different ways: complete replacement (Bednarkiewicz et al., 2002) or reconstruction consisting of an off-pump distal or proximal reimplantation of the aortic arch vessels combined with an endovascular large stent graft to exclude the entire aortic arch (Dambrin et al., 2005; Al Shammari et al., 2000). These surgical techniques are not only associated with considerable mortality and morbidity but also have undoubtedly mechanical consequences since they involve the application of loading (typically pressure) and the generation of strain during the repairing or replacement of sick vessels. This fact justifies the need of achieving a better understanding of the mechanical response of the human aortic arch. One possible way to achieve this goal is by using numerical simulations that may provide useful information for medical therapies of the related pathologies. In this context, one of the major challenges is the definition of realistic and reliable stress-strain relationships of the vessel (Holzapfel et al., 2007).

The aortic arch may be subjected to extreme loading conditions in situations such as automobile crashes. In this context, traumatic aortic arch false aneurysms after blunt chest trauma can be developed (Bizzarri et al., 2008b). Three types of mechanical actions causing the blunt traumatic aortic rupture have been identified: stretching, intravascular pressure and water-hammer effect (Richens et al., 2002; Field et al., 2007). In particular, hypertension at rest or during effort in patients with aortic arch coarctation has been studied in cases with successful repair or mild degree of obstruction (De Caro et al., 2007). Moreover, impact-sled tests with human cadaver thoraces have been carried out to investigate the aortic injury mechanism caused by the effect of acceleration that induces a differential motion of the aortic arch relative to the heart and its neighboring vessels (Forman et al., 2008).

Numerical simulations have been recently performed to predict the mechanical response of the human aortic arch. Beller et al. (2004) and Beller et al. (2005) studied aortic arches under physiological conditions (in patients with and without aortic insufficiency) by means of a linear elastic isotropic constitutive model (i.e., the tissue stiffness variation with increasing strain was not taken into account) and boundary conditions that accounted for experimentally-measured aortic root displacements during the cardiac cycle. Both aortic root displacement and hypertension were found to significantly increase the longitudinal stress in the ascending aorta. Gao et al. (2006) performed a study of the stress distribution in a layered aortic arch model (also using a linear elastic isotropic law) with interaction between a pulsatile flow and the wall of the blood vessel. This work indicates that the circumferential stress in the aortic wall is directly associated with blood pressure, supporting the clinical importance of blood pressure control. However, it should be noted that the numerical simulation and experimental validation of the mechanical response of the human aortic arch under severe loading conditions is nowadays a research subject to be explored.

The aim of this work is to analyze the mechanical response of the human aortic arch during the bending and pressurization test through in-vitro experiments, constitutive modelling and numerical simulation. This task involves both the characterization of the structural behaviour of the aortic arch under severe loading conditions and the numerical validation of a material constitutive model. The large quasi-static deformations considered in the reported test are achieved in three different stages: bending, axial stretching and internal pressurization. The high stretching and pressure levels involved in these experiments pertain to mimic those de-
developed in a real-life situation (e.g., an automobile crash and stents implants). The sample corresponds to a young artery without cardiovascular pathologies. The mechanical behaviour reported in the present investigation deals with strong deformations corresponding to pressure levels within the normal and hypertension physiological ranges. The material and methods considered in this study are presented in Section 2. In particular, the experimental procedure described in Section 2.1 encompasses two different *in-vitro* tests: uniaxial tensile for material characterization together with bending and pressurization. Moreover, Section 2.2 briefly describes the Demiray constitutive model adopted in this work to describe the material response while Section 2.3 presents the procedure to fit the related material parameters. The obtained experimental, analytical and numerical results included in Section 3 are discussed in Section 4. Specifically, the material characterization via the tensile tests using the Demiray constitutive model is detailed in Section 3.1. The procedure to fit the material parameters for such model is particularly analyzed. Although other material models may be used to characterize the behaviour of the aortic arch tissue, this relatively simple model was chosen here in order to evaluate its capabilities and limitations in describing the material response in complex loading conditions. Finally, this material characterization in the modelling of the bending and pressurization test is assessed in Section 3.2 where the numerical results are satisfactorily validated with the experimental measurements.

2 MATERIAL AND METHODS

2.1 EXPERIMENTAL PROCEDURE

2.1.1 MATERIAL

The human aortic arch sample tested in this work has been provided by the Hospital Puerta de Hierro at Madrid. This sample, obtained according to well-established protocols of the Ethical Committee of such Hospital (Goicolea et al., 2006), came from a cardiac transplant donor without previous arterial risk factors (i.e., tissue with low cholesterol levels, normal physiological pressure and absence of arterial pathologies of a donor with neither smoking nor diabetes records) whose death was not related to cardiovascular problems. The aortic arch considered in this study is shown in Figure 1. This corresponds to a 44 years-old woman of 65 kg in weight and 1.60 m tall. All the *in-vitro* mechanical tests described in this work have been performed in the same day using samples obtained immediately after excision (i.e., one day from the time of death to testing). Although for experimental purposes the use of more samples would have been desirable, it should be noted that healthy and young aortic arches are not easily available. However, taking into account the low dispersion observed in the mechanical response of healthy and young human thoracic descending aortas in tensile and pressurization tests (García-Herrera, 2008), the mechanical characterization described below can be assumed to provide a representative behaviour of aortic arches belonging to young donors without cardiovascular pathologies.
2.1.2 TENSILE TEST

One of the most common procedures to characterize the passive mechanical behaviour of the human aortic wall is the tensile test. Specifically, the aim of this test is to obtain a stress-strain relationship of the material under a uniform deformation pattern. In this study, the strip samples were obtained from the arch wall previously used in the bending and pressurization test to be described below.

The tensile test assembly together with the clamp used to fix the sample to the jaws is shown in Figure 2. The tests were carried out with the specimens permanently submerged in physiological serum (standard Phosphate Buffered Saline, PBS) at a temperature of 37 ±0.5 °C. In order to achieve uniform conditions for each sample, a time interval of 10 minutes was considered between the end of the assembly and the beginning of the test. To precondition the samples, ten successive loading cycles were executed up to a stress value of 300 kPa. The load cell velocity considered in the tests up to the rupture of the sample was 0.03 mm/s (which results in a deformation rate of 15%/min approximately). The chosen sample length and width are plotted in Figure 2. As usual, the samples were cut along the longitudinal and circumferential directions in order to characterize the degree of anisotropy in the material response (Mohan and Melvin, 1982; Okamoto et al., 2002; García-Herrera, 2008). Six samples were tested for each direction. Only tests exhibiting rupture at approximately the center of the sample have been considered. Since the artery wall is composed of three different layers (Fung, 1993), valid results are assumed up to the rupture instant of any of such layers.

Axial load and axial jaws displacement were recorded during the whole test (with precisions of 0.01 N and 0.001 mm, respectively; the maximum load cell limit was 10 N). The Cauchy axial stress $\sigma_1$ was computed as $F/A$, where $F$ is the axial load and $A$ is the current transversal area. The axial stretch $\lambda_1$ was calculated as $L/L_0$, with $L$ and $L_0$ being the current and initial sample lengths, respectively. The current transversal area $A$ is evaluated through the incompressibility condition that leads to $A = A_0/\lambda_1$, where $A_0$ is the initial transversal area of the sample. Moreover, assuming a uniform strain distribution along the sample, the stresses
2.1.3 BENDING AND PRESSURIZATION TEST

This test is aimed at assessing the mechanical response of the human aortic arch when it is subjected to large quasi-static deformations given by severe bending, axial stretching and internal pressurization. The initial configuration of the artery (shown in Figure 1) is approximately a $90^\circ$ circular arch with medium radius of 28.5 mm where the average internal diameter and thickness of the transversal section of the arch are 18 mm and 1.3 mm, respectively.

The test was carried out with the specimen permanently submerged in physiological serum (PBS) at a temperature of $37 \pm 0.5$ °C. In order to achieve uniform temperature conditions in the sample, a time interval of 10 minutes was considered between the end of the assembly and the beginning of the test.

In this test, the bending stage is performed to place the specimen in the tensile machine jaws as shown in Figure 3. The self-contact wrinkles that can be seen in the front view are due to the local buckling that develops in this zone during the bending. In contrast, the tissue observed in the back view exhibits a tensile axial stress state.

The next loading stage is achieved by axially deforming the arch whereas the subsequent pressurization stage consists in the application of internal pressure by means of an external compressor that injects PBS into the artery (further details of the experimental setup can be found in Guinea et al. (2005)). The full loading sequence is schematically depicted in Figure 4.

The axial stretching stage is accomplished via a prescribed displacement that corresponds
Figure 3: Aortic arch sample mounted and clamped in the tensile machine jaws (end of bending stage); A) front view and B) back view

Figure 4: Loading sequence: bending, axial stretching and internal pressurization

to an average longitudinal stretch, defined as the ratio between the final and initial lengths of the artery axis, of 1.7. The initial length of the artery axis is considered as the average of the internal and external arc lengths (measured via image postprocessing with an error of ±2.0 mm) while the final length of the artery axis corresponds to the final distance between the jaws. This displacement value was selected to remove the wrinkles and, thus, to allow more accurate measurements of the external diameter of the artery during the pressurization stage. Moreover, the load cell velocity considered in the tests up to the rupture of the sample was 0.03 mm/s.

Internal pressure and external diameter of the sample were recorded during the pressurization stage. An optical extensometer was used to acquire the external diameter evolution (with a precision of 0.001 mm) along the height of the sample. Although the deformed cross-sections were no longer circular, the diameter values measured from the optical extensometer located at different angular positions varied around 10% with respect to the maximum diameter. Moreover, ten successive loading cycles up to a pressure value of 200 mmHg were executed to precondition the samples. This pressure level is beyond the normal physiological
range (i.e., 80 − 120 mmHg). The pressure rate considered in this stage was 1 mmHg/s.

2.2 CONSTITUTIVE MODELING

According to the measurements to be presented in Section 3, an elastic, isotropic and rate-independent material response is considered for the artery analyzed in the present work. Moreover, its behaviour is taken as incompressible due to the large amount of water present in it (Oijen, 2003). To this end, hyperelastic constitutive models can be used to describe its mechanical response (Ogden, 1984; Fung, 1993; Holzapfel, 2000; Raghavan and Vorp, 2000; Prendergast et al., 2003; Masson et al., 2008; Kroon and Holzapfel, 2009). In this context, a deformation energy function \( W \), assumed to describe the isothermal material behaviour under any loading conditions, is usually defined in terms of the right Cauchy deformation tensor \( \mathbf{C} = \mathbf{F}^T \cdot \mathbf{F} \), where \( \mathbf{F} \) is the deformation gradient tensor and \( \mathbf{T} \) is the transpose symbol (note that \( \det \mathbf{F} = 1 \) in this case). Invoking classical arguments of continuum mechanics, the Cauchy stress tensor \( \mathbf{\sigma} \) is defined as \( \mathbf{\sigma} = 2 \mathbf{F} \cdot \frac{\partial W}{\partial \mathbf{C}} \cdot \mathbf{F}^T \).

In particular, the energy function proposed by Demiray (1972) is expressed as:

\[
W = \frac{a}{b} \left[ \exp \left( \frac{b}{2} (I_1 - 3) \right) - 1 \right]
\]

where \( I_1 \) is the first invariant of \( \mathbf{C} \) (\( I_1 = \text{tr}(\mathbf{C}) \), \( \text{tr} \) being the trace symbol). Although this isotropic model is relatively simple (i.e., only depends on \( I_1 \)), reasonably good responses at high levels of deformation can be predicted with it (Delfino et al., 1997). Only two constants, \( a \) and \( b \), are needed for the material characterization where the parameter \( a \) has a clear physical meaning given by the slope at the origin of the Cauchy stress versus stretch tensile test curve.

The constitutive model expressed by equation 1 is adopted in this work to assess its capabilities in the prediction of the material response in the bending and pressurization test to be presented in Section 3.2 by using the material parameters obtained from the mechanical characterization described below.

Moreover, this constitutive model is implemented in an in-house finite element code extensively validated in many engineering applications where isoparametric elements including a \( B-bar \) technique are used to avoid numerical locking due to material incompressibility (see Celentano (2001) and references therein).

2.3 MATERIAL CHARACTERIZATION VIA THE TENSILE TEST

The aim of this section is to determine the material parameters of the Demiray constitutive model briefly presented in Section 2.2 from the uniaxial test measurements to be reported in Section 3.1.

For the Demiray constitutive model, the Cauchy stress associated to the loading direction 1 can be exclusively written in terms of the related stretch \( \lambda_1 \) since the incompressibility constraint for an isotropic behaviour reads as \( \lambda_2 = \lambda_3 = \frac{1}{\sqrt{\lambda_1}} \) (Ogden, 1984). Thus:

\[
\sigma_1 = a \left( \lambda_1^2 - \frac{1}{\lambda_1} \right) \exp \left[ \frac{b}{2} \left( \lambda_1^2 + \frac{2}{\lambda_1} - 3 \right) \right]
\]
In this case, the logarithmic version of equation 2 results in a linear least-squares fitting procedure of the material parameters $a$ and $b$. Moreover, equal weights for both the longitudinal and circumferential responses were simultaneously considered.

3 RESULTS

3.1 TENSILE TEST

The average stress-stretch curves for both the longitudinal and circumferential directions are plotted in Figure 5 (the vertical bars denote the standard error, i.e., the ratio between the standard deviation and the square root of the number of specimens). It should be noted that similar material responses were observed at different positions around the circumference and along the length of the artery. Therefore, the good repeatability achieved in the experiments justifies the assumption of homogeneity in the constitutive model described in Section 2.2. Moreover, the high rupture stress value obtained for this aortic arch can be attributable to the fact that this tissue was young and healthy.

The stress-stretch curve obtained by applying a least-squares fitting of the resulting $\sigma_1(\lambda_1)$ relationship 2 to the corresponding experimental data for both the longitudinal and circumferential directions is plotted in Figure 5. According to the measurements reported above, the whole stretch range (1.0 – 2.3) was chosen for the present material characterization. The resulting material parameters derived with this procedure are $a = 107.19 \pm 5.30$ kPa and $b = 1.40 \pm 0.08$.

Figure 5: Experimental and computed results of Cauchy stress versus stretch
3.2 BENDING AND PRESSURIZATION TEST

The experimentally measured evolution of the internal pressure (during the loading and unloading phases) in terms of the maximum external diameter for the axial stretch achieved in the previous loading stage is shown in Figure 6. In this case, the external diameter rate was found to be 0.05 mm/s approximately (which results in a circumferential deformation rate of 45%/min approximately). The average circumferential stretch at the end of the test, defined as the ratio between the final and initial external diameters of the artery, was approximately 1.5.

Figure 6: Experimental and computed results of internal pressure versus external diameter of the artery

The mechanical response of the human aortic arch during the bending and pressurization test already described in Section 2.1.3 is analyzed in this work by using the non-linear constitutive model and the corresponding material parameters respectively presented in Sections 2.2 and 3.1. To this end, a numerical simulation of the stages involved in such test (i.e., bending including axial stretching and pressurization) is carried out. The aortic arch was geometrically discretized as a 90° elbow with constant internal diameter and thickness (18 mm and 1.3 mm, respectively). A convergence study of the numerical response to different discretizations was performed. The resulting finite element mesh shown in Figure 7A is composed of 1280 eight-noded isoparametric elements. It is seen that only one half of the arch was considered in the computations owing to the symmetry of the problem. A finer discretization was chosen for both ends since large strain and stress gradients are expected in those zones.

The boundary conditions adopted in the simulation, which are essentially the same as those imposed in the experiment, are schematically depicted in Figure 4. The lower end was clamped during the whole test. The bending stage (which also involves axial stretching in the present analysis) was carried out by applying a prescribed displacement at the upper end of the arch (as in the experiment, with an average longitudinal stretch of 1.7) in order to straighten it. In this way, the numerical simulation of local buckling and self-contact developed during the
purely bending phase shown in Figure 3 was avoided. As the mechanical behaviour of the arch is considered elastic, it can be assumed that the simplified boundary conditions adopted in the simulation do not affect the predictions of the material response (this load-history independent response was verified by performing an additional simulation with exactly the same boundary conditions as those of the experiment where, in this case, self-contact is developed at the end of the bending stage). Then, the upper end was clamped to subsequently perform the pressurization stage up to a final pressure value of 200 mmHg (it should be mentioned that the additional non-linearity that resulted from this follower load was taken into account in the simulation). Moreover, due to the large deformations considered in this test, the effect of the residual stress is neglected in the present analysis (according to García-Herrera (2008), the residual stresses can be estimated as 20% of the stresses in an artery loaded within the physiological range).

Figure 7 depicts the computed deformed configurations of the aortic arch during the whole loading sequence: bending (B,C) and pressurization (D-F). Figure 8 plots maximum principal stress contours at the end of the bending and pressurization stages.

The computed curve of the internal pressure versus the maximum external diameter of the artery is plotted in Figure 6. Moreover, qualitative and quantitative experimental-numerical comparisons of the final deformed configuration of the aortic arch are respectively shown in Figures 9 and 10.

4 DISCUSSION

Two zones with different stiffnesses can clearly be identified in the experimental curves of Figure 5. At low deformations, the curves show a flexible response with a nearly constant slope. The first zone ranges up to $\lambda_1 \simeq 1.4$ for both the longitudinal and circumferential samples. In this first zone the material behaviour is clearly isotropic, i.e., its stiffness is mainly provided by the elastin component of the tissue. For larger deformations, on the other hand, the slopes of the curves start to increase up to the rupture stage. The material anisotropy in this second elongation zone, reflected in the largest differences between the responses corresponding to both sample directions, is apparent at the very end of the test. This is due to the significant action of the collagen fibers that occurs at high elongation levels. However, in the stretching range $(1.0 - 1.8)$ the behaviour can be assumed as practically isotropic. Thus, an adequate description of the material behaviour in this deformation range (which in turn exhibits a low stress dispersion) can be simply tackled by means of isotropic constitutive models.

It is also seen in Figure 5 that the Demiray model provides, due to its isotropic nature, an average response that lies between those of the two analyzed sample orientations. An excellent adjustment is clearly seen within the stretching range $(1.0 - 1.8)$. Although an approximate fitting is achieved at high deformations, the stiffness increase in these stretching levels is reasonably well captured.

The experimental pressure-diameter curves shown in Figure 6 exhibit a nearly linear response for pressure values less than 130 mmHg. This is presumably due to the fact that mainly the elastin is active in this deformation interval. The effect of the collagen fibers is apparent for pressure values higher than 150 mmHg where the material becomes stiffer. This
Figure 7: Deformed configurations of the aortic arch for different steps of the numerical simulation: initial (A), bending and axial stretching (B,C) and pressurization (D-F)

Late elastin-collagen transition can be attributable to the fact that, due to the complex pattern that develops after the two first deformation stages (i.e., bending and axial stretching), the collagen fibers play a relevant role at the very end of the pressurization stage. Finally, it should be noted that the loading-unloading sequences for both the axial stretching and pressurization stages exhibited nearly elastic and rate-independent material responses in this case. In particular, this effect is apparent in the loading-unloading curves corresponding to the pressurization stage shown in Figure 8 (note that the maximum difference between the diameters resulting from these two curves is, for a given pressure, less than 5%).

It is seen in Figure 8 that the unfolding of the vessel causes a bending stress pattern, i.e., axial compression in zone I (that exhibits, in addition, local buckling at the ends of the
arch), axial tension in zone III and a neutral axis in the vicinity of zone II. As shown in Figure 7C, this last zone experiences large rotations with low stretching levels. Once the final pressure value is applied, tensile stresses with lower values than that recorded at the rupture stage in the tensile test (see Figure 5) develop in region V. Owing to the previous bending and axial deformation, the stress distribution at the central region of the sample (zones I, II

Figure 8: Maximum principal stress contours [Pa]

Figure 9: Final deformed configuration of the aortic arch: experiment (left) and simulation (right)
Figure 10: Maximum external diameter distribution along the artery length

and III) is not uniform. It can also be observed that the severe deformation developed at zone II of the artery is mainly due to the effect of the internal pressure (the ratios of the final to initial average external diameter and thickness values are 1.55 and 0.55, respectively) where the circumferential stress reaches 590 kPa. Using the average dimensions resulting from the simulation (i.e., external diameter of 32 mm and thickness of 0.71 mm), the analytical expression corresponding to thin-walled tubes gives a stress value of 600 kPa. These similar stress values confirms that the bending effect does not substantially affect the circumferential response in zone II. Moreover, the barrelling formation in zone IV is due to the compression stress state promoted by the initial bending of the vessel.

As already mentioned, the residual stresses can be estimated as 20% of the stresses in an artery loaded within the physiological range (80-120 mmHg), i.e., their values are bounded to 30 kPa (García-Herrera, 2008). The stresses developed in the bending and pressurization test reached 1000 kPa, hence the effect of the residual stresses is practically negligible.

The numerical results shown in Figure 6 reasonably adjust the experimental measurements (the maximum experimental-numerical discrepancy in the diameter is, for a given pressure level, lower than 4%). The nearly linear response predicted by the model is due to the low levels of circumferential stretching developed during the pressurization stage. This is consistent with the material behaviour shown in Figure 5. In addition, it is seen that the numerical results do not properly capture the very different stiffness regime observed in the experiments at the end of the test. However, the material characterization performed in Section 3.1 was found to provide an overall realistic response of the aortic arch when it is subjected to internal pressure.

An overall good agreement between the experimental and computed final deformed configu-
ration of the aortic arch can be clearly appreciated in Figure 9. In particular, the unsymmetric deformation pattern observed in the experiments is adequately captured by the simulation. The experimental observation that the deformed cross-sections were no longer circular was also confirmed by the numerical predictions. Finally, a reasonable good agreement between experimental and computed values for the maximum external diameter distribution along the artery length at the final deformed configuration can be appreciated in Figure 10.

The Demiray model is known to be ill-posed for experimental characterization of the material response (see Criscione (2003)). This drawback could be tackled, as proposed by Criscione (2004), by using an alternative set of invariants such that the resulting energy function minimizes the experimental error. It should be noted, however, that the methodology used in this work was found to provide an overall good fitting between the experimental and numerical results.

5 CONCLUSIONS

Experiments, constitutive modelling and numerical simulation aimed at analyzing the mechanical response of the human aortic arch during the bending and pressurization test has been presented. The experiments carried out in this work were designed to achieve high stretching and pressure levels as those developed in a real-life situation (e.g., an automobile crash or specific surgical treatments such as stents implants). The realization of this in-vitro test is a novel contribution of the present work. Overall, the material characterization carried out in this study was found to be consistent since it provided a reasonable and realistic description of the mechanical behaviour of the aortic arch under the large quasi-static deformations considered in this test. In particular, a good agreement between the experimental and computed results of the internal pressure versus the external diameter of the artery have been obtained. The description of the material response during the bending and pressurization test applying a model previously characterized via tensile testing is also an original aspect of this study.

Future research on this area will be focused on the limitations of the present analysis, i.e., further validation including more results than those considered in this work, assessment of other constitutive models (e.g., those defined within a well-posed theoretical framework for experimental determination of the material response), influence of residual stresses and material inhomogeneity and, in addition, experiments, modelling, inverse material parameter estimation and simulation of arteries subjected to extreme dynamic loading conditions.

6 CONFLICTS OF INTEREST

The authors have no conflicting interests regarding this paper.

7 ACKNOWLEDGEMENTS

The authors wish to express their appreciation to Dr. R. Burgos and C. García-Montero of the Hospital de Puerta de Hierro at Madrid for the provision of arterial tissues analyzed in this
work. The supports provided by the Postdoctoral Program at the Universidad de Santiago de Chile (USACH) and the FONDECYT Project No. 11090266 of the Chilean Council of Research and Technology (CONICYT) are all gratefully acknowledged.

8 REFERENCES


