

Comparison between the mechanical behaviour of the human healthy AA and commercial prostheses under various mechanical loadings

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1. Introduction

Standard surgical treatment of abdominal aortic aneurysm involves the placement of tubular synthetic aortic prostheses. Most of these implants are made up of polyester textiles or porous expanded polytetrafluoroethylene. Normalised tests are dedicated to their assessment (ISO 7198:1998, cardiovascular implants and tubular vascular prostheses). However, such experiments are not sufficient to characterise the complete mechanical performance of these implants (Le Magnen et al. 2001) and to ensure their mechanical compatibility with the host artery. Thus, the design of mechanically compatible vascular prostheses still remains a challenge. Within this context, a full comparison of the mechanical behaviour of the human healthy abdominal aorta (AA) with commercial prostheses is proposed. An original numerical database on the mechanical behaviour of human AA subjected to various mechanical loadings is first built and then compared with experimental data obtained from mechanical tests performed on prostheses.

2. Methods

2.1 AA mechanical model

A two-layer model based on the ρ -model (Holzapfel and Ogden 2010) was chosen to describe the anisotropic hyperelastic AA behaviour. The volumetric strain–energy function is written as:

$$W = W_{\text{iso}} + W_{\text{m}} + W_{\text{a}}, \quad W_{\text{iso}} = 0.5c_1(I_1 - 3),$$

where W_{iso} is the isotropic contribution, c_1 being a material parameter and I_1 the first invariant of the right Cauchy–Green tensor. Functions W_{m} and W_{a} are the anisotropic contributions of the two-fibre families in the media and the adventitia:

$$W_i = \frac{k_1^i}{k_2^i} \left\{ e^{k_2^i [(1-\rho)(I_1-3)^2 + \rho(I_4-1)^2]} - 1 \right\}, \quad i \in \{m, a\},$$

where k_1^i and k_2^i denote the fibre material parameters, $\rho \in$

$[0, 1]$ is a measure of the AA anisotropy and I_4^i is an invariant depending on the angle between fibres in each layer θ_0^i . These eight material parameters adjusted (see Figure 1(a),(b)) with the biaxial tensile data reported by Vande Geest (2005) are reported in Table 1, for a group of donors aged < 30 years (Gr. 1), between 30 and 60 years (Gr. 2) and > 60 years (Gr. 3).

2.2 Numerical simulations

The AA model was implemented in a finite element code (Comsol Multiphysics™ 3.5a). Planar uniaxial or biaxial tests were simulated on planar AA square (20 mm width) specimens. Various tension ratios $T_{\theta_0}:T_{11}$ were used, i.e. 0:1, 0.5:1, 0.75:1, 1:1, 1:0.75, 1:0.5 and 1:0, the tests being stopped when the highest tension reached 120 N m^{-1} . Cantilever bending tests of slender strips of AAs under their own weight were simulated on similar samples. Cantilever bending tests were simulated on a representative AA geometry, i.e. a tube of 20 mm inner diameter and 50 mm long. Finally, inflations of the AA geometry with an internal pressure varying from 0 to 160 mmHg were simulated for several values of axial prestretch of the AA ($\lambda_{11} = 1-1.2$). The Peterson modulus E_P could thus be deduced from these simulations in the pressure range 80–120 mmHg.

2.3 Prostheses mechanics

Two aortic textile prostheses were selected, showing either a woven (P_1) or a knitted (P_2) structure. Both prostheses were characterised by a severe longitudinal crimping. Each tissue was solely subjected to stretch-controlled uniaxial tests, performed at a stretch rate of 10^{-2} s^{-1} , thanks to a tensile device (ElectroForce®, Bose Corporation, maximum load: 400 N). Both bending and inflation tests were conducted using mechanical conditions similar to those applied in numerical simulations. All tests were

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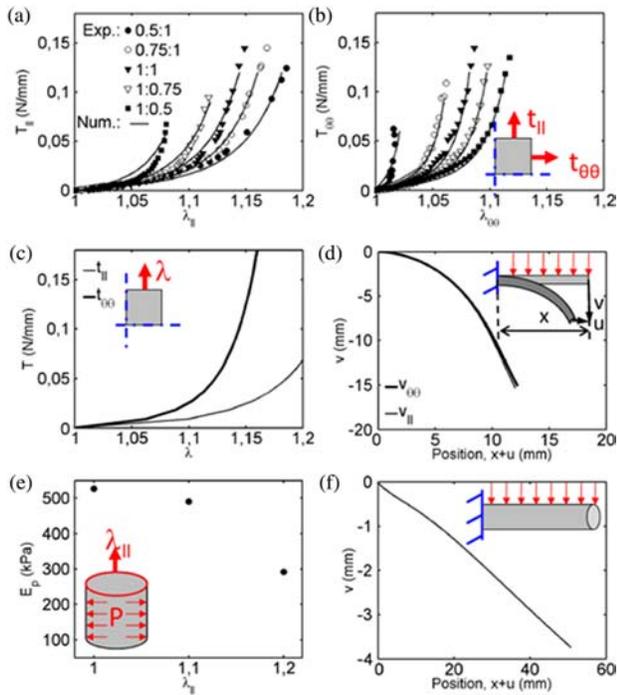


Figure 1. Numerical results for the Gr. 2 AAs (a and b): biaxial tensile tests versus Vande Geest's (2005) data (marks), (c) uniaxial tensile tests, (d) tissue sample bending test, (e) Peterson modulus and (f) tube bending test, displacement versus axial position.

combined to video recordings (Nikon CCD video camera 7360×4912 px) of the tube's initial and deformed shapes (Figure 2(a)). After bending tests, vertical and horizontal displacements of the tube centreline were extracted from the recorded images. During the inflation tests, the target axial prestretches were achieved by controlling the crimping deployment. Afterwards, the diameter variations of each prosthesis were detected from the images and the Peterson modulus E_P was estimated.

3. Results and discussion

Numerical results obtained for the Gr. 2 AAs are presented in Figure 1. Figure 1(a),(b) provides good agreement between the adjusted AA model and experimental results of Vande Geest (2005). The two simulated tensile tests plotted in Figure 1(a)–(c) clearly underline the anisotropic mechanical behaviour of the healthy aortic tissue. Figure 1 (d) depicts the bending of an AA slender strip under its own weight, in both longitudinal and orthoradial

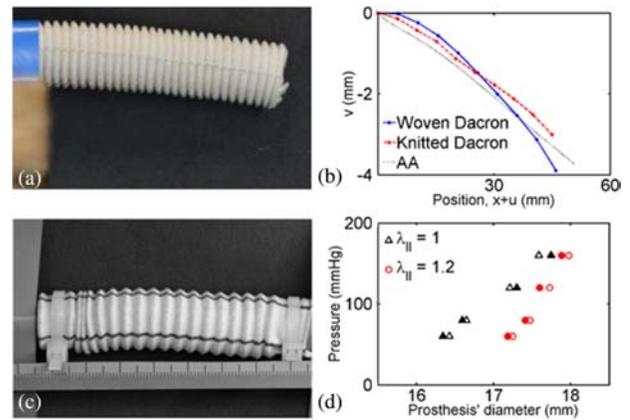


Figure 2. (a) Bending test on prosthesis P_1 . (b) Experimental and numerical results: displacement versus position. (c) Inflation test on P_2 . (d) Experimental pressure–diameter variations.

directions. In contrast to previous results, such mechanical response is weakly influenced by the anisotropy of the AA wall. Due to the anisotropy of the AA wall, E_P decreases while increasing the axial prestretch (see Figure 1(e)). Finally, Figure 1(f) shows the vertical displacement of the AA centreline during a cantilever bending test. Note that similar trends have been observed in the case of Gr. 1 and Gr. 3. However, if the mechanical responses of Gr. 2 and Gr. 3 AAs are quite similar, they are much stiffer than the mechanical responses for the Gr. 1 for planar biaxial and inflation tests. Surprisingly, our results also showed that this trend is reverted in the case of bending tests at both the tissue or aorta scale. These results must be confirmed by developing similar experiments. Figure 2(a) shows an example of a bending test on P_1 . Figure 2(b) depicts the evolution of the vertical displacement versus the axial position. The mean difference between the prostheses and the Gr. 2's AA results is 13.9% for P_1 and 8.8% for P_2 . Figure 2(c) illustrates an experimental inflation test performed on P_2 . Pressure–diameter curves are displayed in Figure 2(d) for axial prestretch $\lambda_{II} = 1$ and 1.2. The average E_P was assessed to 493 and 1406 kPa. In contrast to the AA, our experimental results showed that this modulus increases while increasing the axial stretch. The actual longitudinal crimping of vascular prostheses clearly increases their bending rigidities, which are similar to the bending rigidities of the AAs. However, this crimping may also lead to strong variations of prostheses' E_P with the axial prestretch.

Table 1. Optimised parameters of the AA model.

	c_1 (kPa)	k_1 (kPa)	k_2 (–)	θ_0 (°)	k_1 (kPa)	k_2 (–)	θ_0 (–)	θ_0 (°)	ρ (–)
1	13.6	3.3	0.2	0.5	3.9	0.5	88.8	0.7	
2	1.0	30.2	59.3	19.2	29.6	30.1	59.3	0.4	
3	3.1	41.1	32.0	4.0	26.8	49.1	77.7	0.2	

4. Conclusions

A comparison between the mechanical behaviour of the AA and prostheses under various mechanical loadings was performed. Bending tests and inflation tests have also been performed. The results showed that the prosthesis' crimping plays an important role on its mechanical behaviour; this crimping may or may not help at reproducing the mechanical behaviour of AA. Moreover, uniaxial tests on textiles, not presented here, have shown that they are much stiffer than AA tissue in both longitudinal and orthoradial directions and that their anisotropic behaviour remains very far from that of the host artery. All these results point out that the actual prostheses are not fully mechanically compatible with the host aorta.

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