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Communication

New experimental protocols for tensile testing of abdominal aortic analogues

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ABSTRACT

This work proposes an *in vitro* tensile testing protocol that is able to characterize abdominal aortic (AA) analogues under physiologically inspired mechanical loadings. Kinematic parameters are defined in agreement with *in vivo* measurements of aortic dynamics. A specific focus is given to the choice of the applied loading rates, deriving from the knowledge of aortic Peterson modulus and blood pressure variations from diastolic to systolic instants. The influence of physiological elongation rates has been tested on both porcine AAs and a thermoplastic polyurethane (TPU) material used to elaborate AA analogues. The diastolic and systolic elongation rates estimates vary between orders of magnitude $\mathcal{O}(10^{-2})$ and $\mathcal{O}(10^{-1})\,s^{-1}$. Negligible differences are obtained when comparing stress–elongation responses between both physiological elongation rates. In contrast, a noticeable stiffening of the TPU mechanical response is observed compared to that obtained under the common low traction rate of $\mathcal{O}(10^{-3})\,s^{-1}$. This work shows how relevant physiological elongation rates can be evaluated as a function of age, gender and pathological context.

1. Introduction

Abdominal aortic aneurysm (AAA) is a permanent dilatation of the abdominal aorta (AA). Fundamental knowledge of AAAs dysfunctional biomechanics requires the mechanical characterization of aortic tissue under appropriate (patho-)physiological conditions. Alternatively to *in vivo* investigation of vascular mechanics, deformable AAA analogues have been made in the last decades. Most were placed into vascular flow simulators to investigate endovascular aneurysm repair [1–4] or fluid–structure interactions within AAAs [5,6]. Inflation testing was also conducted to identify AAAs deformation [7,8]. Yet, the intrinsic material properties of AAA analogues have been barely investigated and when they were [9,10], the testing kinematic conditions were not discussed in connection with previous protocols carried out on biological samples.

Numerous *in vitro* tensile tests have been reported to determine the mechanical behaviour of human AA/AAA [11–16] and porcine AA [17–20]. The typical protocol begins with a preconditioning phase (5–10 cycles) applied at a peak strain (5–10%) and constant elongation rate ($10^{-3}\,s^{-1}$), followed by a monotonic stretching to a chosen peak strain. Such measurements showed AA hyperelastic and anisotropic mechanical behaviour. Aortic wall's nonlinear viscoelastic properties were also demonstrated, albeit by very few studies [21,18,20]. Therefore, two factors are commonly discarded in experimental protocols, which make them unsuitable for mimicking physiological mechanical loadings:

- A single elongation rate is often considered during the characterization. So, periodic changes of tissue elongation rate occurring during the cardiac cycle are neglected.
- The relevance of the chosen elongation rate magnitude has been barely discussed regarding to *in vivo* mechanical loadings [22].

This study aims to propose a tensile-testing protocol able to characterize aortic analogues under mechanical loadings closer to *in vivo* loadings using suitable elongation rates, to test the influence of these elongation rates on both porcine AAs, and a polymer used in a recent vascular flow simulator [23,5].

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2. Materials and methods

2.1. Materials

2.1.1. Aortic analogue

An idealized AAA model was manufactured using a mixture of Estane® 5714 TPU [7]. Eight rectangular samples (S_i, $i\in[1\ldots8]$) were cut from tubular sections along orthoradial $\boldsymbol{e}_\theta$ and longitudinal $\boldsymbol{e}_l$ directions. Unloaded original length l_0, width w_0 and thickness t_0 were measured (±0.01 mm). Undeformed cross-sectional area was derived as $S_0 = w_0 \times t_0$.

2.1.2. Biological tissue

Three healthy AA tubular samples were excised from three 4-month old male pigs weighing 31 ± 4 kg. Experimental procedure was approved by the ethics board of the Surgical Center for Education and Research at Marseille's Nord Hospital. Five rectangular strips (B_i, $i\in[1\ldots5]$) were cut along $\boldsymbol{e}_l$. B_1 and B_2 (resp. B_4 and B_5) were extracted from the same aortic sample.

Strips' dimensions are summarized in Table 1.

2.2. Methods

S_i and B_i biomechanical behaviour was investigated using a uniaxial tensile-testing device [7]. The actual force f, and length l, in the stretch direction were measured. The corresponding Cauchy stress σ was calculated as the load f per unit of actual cross-section S. Prior to a kth test performed on a strip, initial length l_0^k and force f_0^k were measured (optional index $k=0$ refers to the undeformed configuration, *e.g.* $l_0^0 = l_0$). These values could differ from l_0 and null tension, when a previous load yielded to a residual elongation $\lambda^k = l/l_0^k$ and non-zero pre-stress $\sigma_0^k = f_0^k l_0^k/(S_0 l_0)$. Elongation rate during test k refers to the absolute time derivative $|\dot{\lambda}^k|$.

2.2.1. Preliminary stretch failure tests

Failure tests were performed on B_1 and B_2 at $|\dot{\lambda}^0| = 10^{-3}\,s^{-1}$, allowing comparison with published data on longitudinal porcine and human AA specimens. Fig. 1 shows the similar material properties of human AA wall and porcine tissue under test.

2.2.2. Physiologically inspired protocols

This part focuses on elaborating physiologically inspired protocols using more suitable kinematic parameters. According to above results, our approach was based on the human AA Peterson modulus value:

$$E_p = D_d \frac{P_s - P_d}{D_s - D_d}, \tag{1}$$

where P_s, P_d, D_s, D_d represent the systolic and diastolic pressure and diameter. Its *in vivo* evaluation derives from measurements of maximal luminal diameters and blood pressure variations from peak diastolic to systolic instants, referred as t_d and t_s [24,25]. Arterial cyclic motion occurs predominantly in the circumferential direction [26,27]. Therefore, assuming AA as an incompressible thin-walled cylindrical tube [12,25], AA tissue undergoes a periodic maximal elongation λ_m, assessed by:

$$\lambda_m = 1 + \frac{D_s - D_d}{D_d} = 1 + \frac{P_s - P_d}{E_p}, \tag{2}$$

Aortic tissue average elongation rates occurring during diastole and systole can be determined as $|\dot{\lambda}_d| = |\Delta\lambda|/\Delta t_d$ and $|\dot{\lambda}_s| = |\Delta\lambda|/\Delta t_s$, where $\Delta\lambda = \lambda_m - 1$ represents the maximal variation of tissue elongation between systolic and diastolic peaks, Δt_d, the diastole

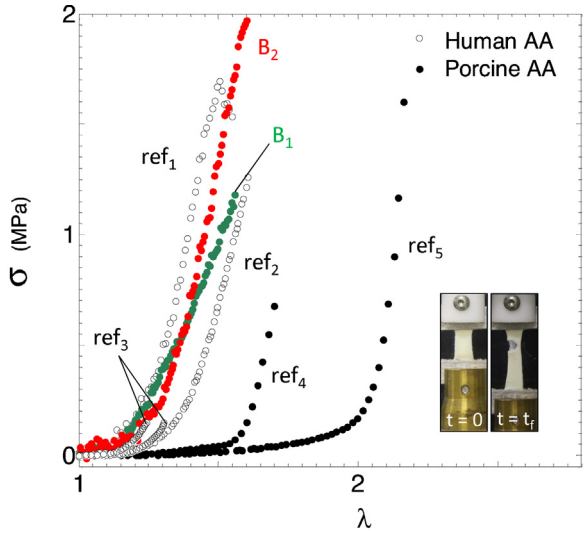

Fig. 1. Comparison between experimental data derived from monotonic tensile tests conducted on porcine and human AA specimens. Measurements of the present study carried out on porcine samples B_1 and B_2 are plotted in color, by contrast with previous data reported by [12] [ref$_1$], [16] [ref$_2$], [11] [ref$_3$], [20] [ref$_4$] and [17] [ref$_5$]. Pictures represent two typical configurations of a tested strip at initial ($t=0$) and failure ($t=t_f$) shot-instants.

Table 1
Geometrical and initial load parameters of the synthetic and biological samples tested in the present uniaxial loading measurements. Indice k in l_0^k and f_0^k refers to test number k performed at a constant specific elongate rate $|\dot{\lambda}^k|$ ($k\in[1\ldots3]$).

Sample	Axis	t_0 (mm)	w_0 (mm)	l_0 (mm)	$\alpha = l_0 : w_0$ (-)	l_0^1 (mm)	l_0^2 (mm)	l_0^3 (mm)	f_0^1 (N)	f_0^2 (N)	f_0^3 (N)
Synthetic AA											
S_1	$\boldsymbol{e}_l$	0.22	5	15	3.0	15.00	15.37	15.69	0.01	0.26	0.49
S_2	$\boldsymbol{e}_l$	0.26	5	15	3.0	15.50	15.91	16.21	0.01	0.25	0.48
S_3	$\boldsymbol{e}_l$	0.27	5	16	3.2	16.00	16.36	16.66	0.01	0.22	0.44
S_4	$\boldsymbol{e}_\theta$	0.22	5	15	3.0	15.00	15.40	15.76	0.01	0.27	0.52
S_5	$\boldsymbol{e}_\theta$	0.22	5	15	3.0	15.00	15.29	15.56	0.01	0.26	0.47
S_6	$\boldsymbol{e}_\theta$	0.23	5	16	3.2	16.00	16.43	16.75	0.01	0.27	0.46
S_7	$\boldsymbol{e}_l$	0.22	5	15	3.0	15.00	15.47	15.85	0.01	0.20	0.44
S_8	$\boldsymbol{e}_l$	0.28	5	16	3.2	16.00	16.31	16.85	0.01	0.26	0.45
Biological AA											
B_1	$\boldsymbol{e}_l$	1.30	9.90	19.40	1.9	–	–	–	0.02	–	-
B_2	$\boldsymbol{e}_l$	1.40	7.00	26.90	3.8	–	–	–	0.01	–	-
B_3	$\boldsymbol{e}_l$	1.49	5.66	15.52	2.7	22.38	22.73	26.10	0.01	0.02	0.08
B_4	$\boldsymbol{e}_l$	1.20	4.50	19.00	4.2	22.55	24.89	27.18	0.01	0.05	0.09
B_5	$\boldsymbol{e}_l$	1.10	8.00	21.50	2.7	25.03	27.43	29.71	0.09	0.16	0.16