

Fabrication of elastomer arterial models with specified compliance

M. STEVANOV, J. BARUTHIO, AND B. ECLANCHER

Université Louis Pasteur, Faculté de Médecine, Institut de Physique
Biologique, 67085 Strasbourg Cedex, France

Stevanov, M., J. Baruthio, and B. Eclancher. Fabrication of elastomer arterial models with specified compliance. *J Appl Physiol* 88: 1291–1294, 2000.—A simple way of making elastic tubes using a mechanical lathe for precise control of the wall thickness is proposed in this study. These tubes are particularly useful for modeling properties of large arteries. Tubes with different geometric parameters and hence different elastic behavior have been made with a silicon elastomer (Rhodorsil RTV 1556). They have been created to be used for compliance measurements in hemodynamics research. Within a limited range of pressures, depending on the wall thickness, such tubes can be used to study models in which the compliance value is assumed to be constant.

wave velocity; hemodynamics; vessel wall elasticity

THE MECHANICAL PROPRIETIES of large blood vessels are an important aspect of cardiovascular function. Vascular compliance (C) is a useful physical parameter to measure these properties because it reflects the elastic nature of blood vessel walls. It is defined as the fractional change in vessel cross section (A) per unit change in pressure (P) normalized by the mean vessel cross section (A_0), as given in Eq. 1

$$C = \frac{\partial A / \partial P}{A_0} \quad (1)$$

It is known to be inversely related to the pulse-wave velocity (c_0) via

$$C = \frac{1}{\sqrt{\rho c_0}} \quad (2)$$

where ρ denotes the fluid density.

Changes in aortic compliance appear to correlate with age and several artery diseases such as hypertension or atherosclerosis (1, 2). In vitro hemodynamics research requires arterial models with physiological values of compliance. The dependence of compliance on distending pressure is complex because a vessel may exhibit a nonlinear and spatially nonuniform elastic behavior. However, to establish a tractable relationship between pressure, compliance, and vessel area, in some theoretical models (4) it is assumed that compliance

values are constant over the range of pressures encountered during the distension of a vessel segment. In this study, we propose a simple way of making arterial models with a constant compliance value within the limited physiological range of pressure. The compliance of the tubes can be predicted before they are produced by using the relationship between the elastic properties of the silicon elastomer and the geometric characteristics of the tubes. Different values of compliance, covering the physiological range, were obtained by varying the wall thickness while keeping the tube diameter constant. The precise control of the wall thickness during the fabrication process allows us to choose the most suitable properties for the type of vessels studied.

MATERIALS AND METHODS

Choice of elastic characteristics. The rate at which the flow and pressure waves propagate in an elastic vessel is determined from the elastic properties of the vessel wall and blood. The Moens-Korteweg equation gives the pulse-wave velocity as a function of the vessel and fluid characteristics

$$c_0 = \frac{1}{\sqrt{\rho \left(\frac{1}{\epsilon} + \frac{2R}{Eh} \right)}} \cong \frac{1}{\sqrt{\rho 2R/Eh}} \quad (3)$$

where ϵ denotes the modulus of fluid elasticity, R is the inner radius of the vessel, E is the modulus of wall elasticity (Young's modulus), and h is the vessel thickness. For blood, as for water, ϵ has a considerably higher value than for vessel walls (ratio of about 10^4), so that the term $1/\epsilon$ in Eq. 3 can be neglected (3). Equation 3 supposes a perfectly elastic tube with a small wall thickness in comparison to the tube diameter. These conditions have been respected in the manufactured tubes because they represent large arterial models. By choosing the geometric parameters of the tubes (diameter and wall thickness), we determined their compliance and pulse-wave velocity using Eqs. 2 and 3.

Tube fabrication. The rubber tubes were made out of a two-component silicon elastomer (Rhodorsil RTV 1556 A and B, Rhône-Poulenc Silicones), which cross-links at room temperature through polyaddition reaction. Rhodorsil RTV 1556 A and B is a viscous liquid, which after cross-linking becomes a strong elastic material. After preparing the mixture and degasifying it under vacuum (about 10 mmHg), we poured the mixture into a disposable syringe. The elastomer was then applied on a Vaseline-coated rigid brass bar of circular cross section, spinning at low speed (5 rpm) in the horizontal position (Fig. 1). The Vaseline allowed us to remove the tube. The bar was spun on a mechanical lathe at a speed that enabled the viscous mixture to run down and spread uni-

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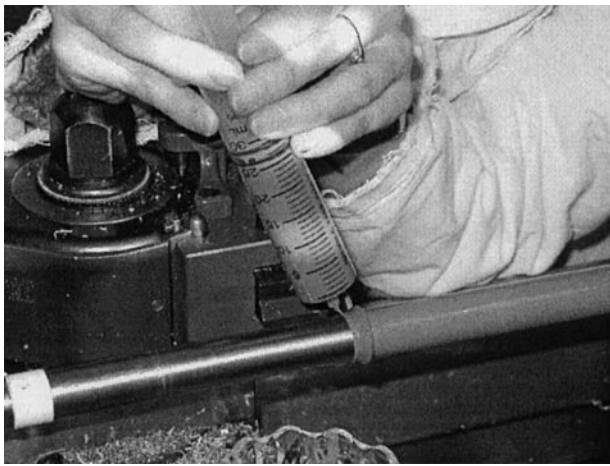


Fig. 1. Fabrication of a tube. Elastomer in liquid state was applied on a rigid metallic bar of circular cross section, spinning at low speed. Elastomer thickness was controlled with a mechanical lathe tool.

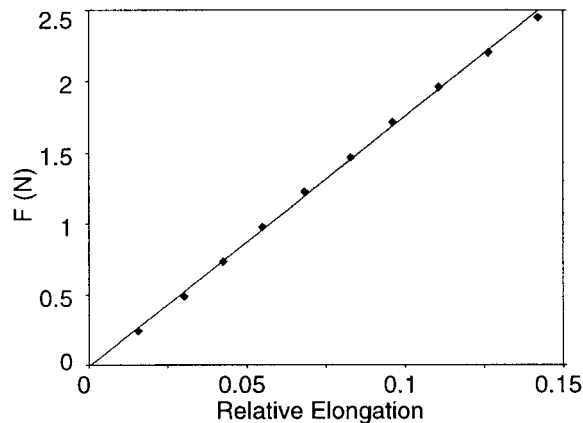
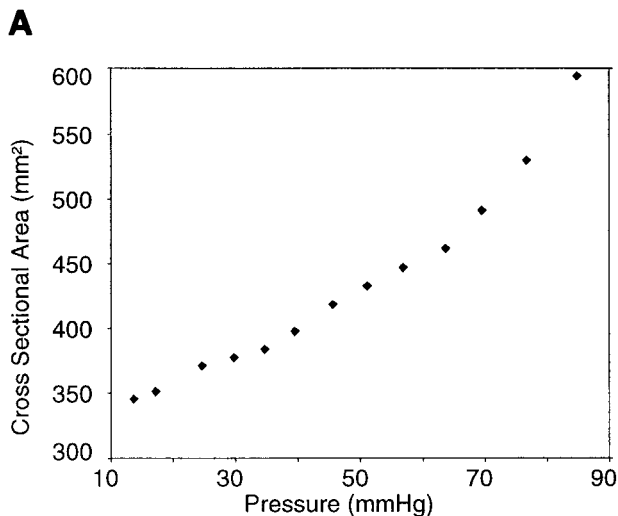


Fig. 2. Relationship between force (F) and relative elongation obtained on elastomer sample. Relationship is remarkably linear ($r^2 > 0.99$). Young's modulus, calculated as ratio of curve slope to cross section of sample, was $6.68 \times 10^5 \text{ N/m}^2$.



formly through centrifugal action. The syringe was used to deliver a thin thread of mixture, a few millimeters above the bar, at a constant rate, while the thickness of the material on the bar was controlled with a lathe tool with an accuracy of 0.1 mm. Cross-linking was accelerated by heating up the material on the spinning bar at $\sim 60^\circ\text{C}$ for 1 h. The tube was finally removed from the bar after having rested at room temperature for 1 day.

Before the fabrication of the tubes, a cylindrical elastomer sample 5.8 mm in diameter was made to determine the value of Young's modulus. The mixture was prepared as for the fabrication of tubes, and then it was introduced into a test tube. After cross-linking, the sample was released by breaking the test tube. Young's modulus was determined by measuring the relative elongation as a function of the applied force. From the simplified Moens-Korteweg equation (Eq. 3) and the relationship between the pulse-wave velocity and the compliance (Eq. 2), the geometric parameters were chosen for the tubes to meet the expected compliance values. More accurate compliance measurements were finally obtained from area-pressure curves for each manufactured tube.

RESULTS AND DISCUSSION

To determine the wall thickness corresponding to different compliance values within the physiological range, measurements of Young's modulus were performed before the tubes were produced. We found Young's modulus to be constant within the range of applied forces (Fig. 2), which characterizes the elastic properties of the silicon material. Young's modulus computed from this curve was $6.68 \times 10^5 \text{ N/m}^2$. With this value, we determined approximate wall thickness values for different tubes with the required compliance. Seven 35-cm tubes with inner diameters of 19 mm and wall thicknesses of 0.24, 0.31, 0.5, 0.58, 0.61, 0.74, and 1.04 mm were made and tested. The wall thickness of samples sliced off the tubes was measured with a profile projector (Nikon, 6C) with an accuracy of 0.01 mm.

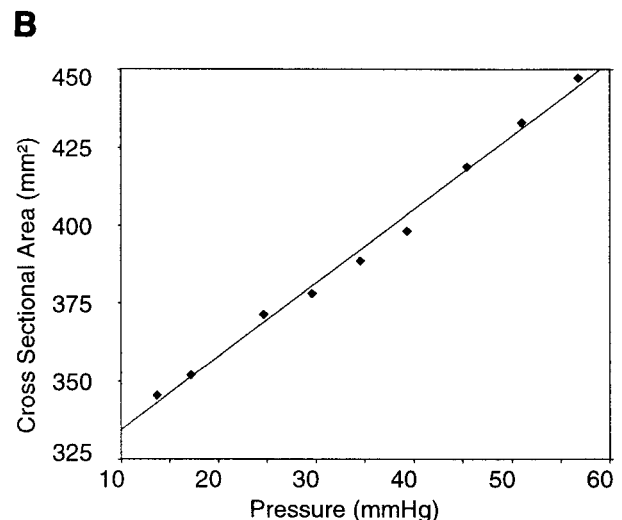


Fig. 3. *A*: cross-sectional area vs. pressure data plot for tube with a wall thickness of 0.61 mm. *B*: curve plotted from data in *A*, for pressures below 60 mmHg representing the part of the data set for which a linear relationship ($r^2 > 0.99$) has been established. Compliance was calculated from correlation between area and pressure.

Table 1. Predicted and measured compliance values with relative errors for the 7 tubes with different wall thickness and ranges of pressure in which measurements were performed

Wall Thickness, mm	Linear Range of Pressure, mmHg	Predicted Compliance, %/mmHg	Measured Compliance, %/mmHg	Relative Error, %
0.24	0–20	1.58	1.63	3.5
0.31	0–25	1.26	1.33	5.2
0.5	0–45	0.76	0.80	5.5
0.58	0–50	0.65	0.64	2.2
0.61	0–60	0.62	0.60	3.8
0.74	0–70	0.51	0.57	11.4
1.04	0–80	0.36	0.32	12.3

Predicted compliance was computed from Eqs. 2 and 3 with $R = 9.5$ mm, $\rho = 1.006 \times 10^3$ kg/cm³ and $E = 6.68 \times 10^5$ N/m². R , inner radius of tube; ρ , fluid density; E , modulus of wall elasticity.

The compliance of the tubes was determined from quasistatic area-pressure curves. To obtain those curves, the tubes were mounted inside a water-filled tank and connected to a water column. The pressure in the tubes was made to vary by changing the height of the water column and was measured with a pressure catheter (Millar microtip catheter pressure transducer; tip PC-350). The diameter was measured at the position where the pressure was measured by using an echo-tracking ultrasonic apparatus. Area-pressure measurement curves were obtained by varying the pressure from 10 to 85 mmHg in steps of ~ 5 mmHg. The curves show different elastic behavior for tubes with different wall thicknesses. A linear relationship between area and pressure has been established up to a limit value of pressure, depending on the wall thickness. This value increases as a function of the thickness of tube wall. For pressures exceeding the limit value, the area-pressure relationship becomes nonlinear. An example of an area-pressure curve, for the tube with 0.61-mm wall

thickness, is shown on Fig. 3A. The compliance was always calculated from the linear part [coefficient of correlation (r^2) > 0.98 for all the tubes] of the area-pressure curves using Eq. 1. In the case of the tube with 0.61-mm wall thickness, shown in Fig. 3B, the compliance was 0.60%/mmHg for pressures between 0 and 60 mmHg. The results of compliance measurements compared with the compliance values predicted from Eqs. 2 and 3 for the seven tubes are given in Table 1. Figure 4A shows the good linear correlation between the predicted and measured compliance values. The linear regression line is $y = 1.07x - 0.04$ (%/mmHg) with $r^2 = 0.99$. In Fig. 4B the same measurements have been used to plot the wave velocity calculated from the measured compliance values using Eq. 2 vs. the predicted compliance, together with a reference curve, presenting an ideal case in which measured and predicted values would be equal.

The values of the compliance measurements agree with the values calculated from Eqs. 2 and 3 for all the tubes, with an error of 5.5% or less, except for the two least flexible tubes, for which the error increases to 11.4 and 12.3% (Table 1). We suppose that the error does not arise from the calculated values, because the assumption underlying the Moens-Korteweg equation that the tubes are thin walled ($ht/2R < 0.1$) has been quite respected for all the tubes. A disagreement between the computed and measured values could, however, be explained by less accurate compliance measurements for more rigid tubes, as relative changes in the tube diameter decrease for “thick” tubes. Measurements could be improved by choosing a more precise technique for diameter measurements.

In conclusion, the technique proposed in this study offers a simple way of producing arterial models with linear elastic behavior within given pressure ranges, depending on wall thickness. Choosing the geometric parameters allows considerable freedom in producing

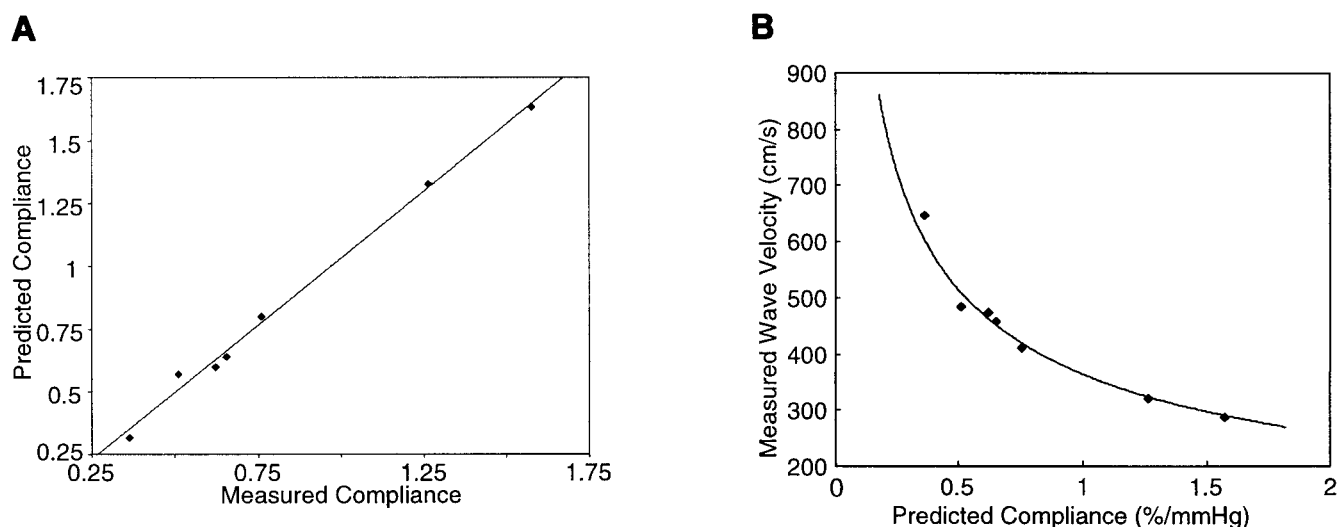


Fig. 4. A: measured compliance values vs. compliance values predicted from Moens-Korteweg equation for 7 different tubes are given by points. Linear regression line (solid line) is $y = 1.07x - 0.04$ (%/mmHg). B: wave velocities calculated from measured compliance values by Eq. 2 vs. predicted compliance values for 7 different tubes are given by points and compared with reference curve on which predicted wave velocity values have been plotted against predicted compliance values (solid line).

tubes with different compliance values. Tubes made in this way can be used in hemodynamics research, for instance in the development of new techniques for in vitro compliance measurements such as magnetic resonance flow imaging or various Doppler methods.

Address for reprint requests and other correspondence: M. Stevanov, Institut de Physique Biologique, Faculté de Médecine, 4, rue Kirschleger, 67085 Strasbourg Cedex, France (E-mail: stevanov@ipb.u-strasbg.fr or baruthio@ipb.u-strasbg.fr).

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