Static Elastic Properties of the Left Coronary Circumflex Artery and the Common Carotid Artery in Dogs

By Dalil J. Patel, M.D., Ph.D., and Joseph S. Janicki, M.S.E.

ABSTRACT

Static elastic properties were studied in isolated segments of the left coronary circumflex artery (LCCA) and the right common carotid artery (CA) from 14 dogs at in-vivo values of pressure and length. A microscope with a Filar micrometer was used to measure radius and length. From these data it was possible to study the pressure-radius relationship and compute various anisotropic elastic moduli as well as the strain energy density (SED). Results indicate: (1) Both arteries exhibited hysteresis which could be minimized within two inflation-deflation cycles. (2) The pressure-radius relationship at constant length was essentially linear over a pressure range corresponding to the in-vivo pulse pressure. (3) The value of volume distensibility was $3.4 \times 10^{-4}$ cm$^2$/dyne for LCCA and $1.5 \times 10^{-4}$ cm$^2$/dyne for CA. (4) LCCA was more distensible than CA in the circumferential direction. (5) LCCA was stiffer in the longitudinal direction than in the circumferential direction; the reverse was true of CA. (6) SED values at physiologic dimensions were $314 \times 10^3$ dyne/cm$^2$ for LCCA and $226 \times 10^3$ dyne/cm$^2$ for CA. These values could be reproduced within 6% when the loading procedure was altered.

ADDITIONAL KEY WORDS pressure-radius relation coronary artery artery elasticity strain energy density

The aim of the present investigation was to study, in vitro, the static elastic properties of the left circumflex coronary artery and the right common carotid artery in the physiologic range of pressure and length. For each artery the study consisted of (a) obtaining the pressure-diameter relationship at constant in-vivo length around the in-vivo mean pressure, (b) estimating the static anisotropic elastic moduli of the vessel wall (1) around the in-vivo state of strain, and (c) calculating the amount of energy stored in the vessel wall during the process of restoring it from the unstressed state to the in-vivo state.

Gregg et al. (2) studied the pressure-volume relationship of the left anterior descending coronary artery and Bergel and others (3-5) studied the pressure-diameter relationship of the common carotid artery. Although useful, these studies fail to provide a detailed picture of vascular mechanics, which is required to answer some important questions concerning the role of mechanical factors in early vessel damage. For example, recent evidence indicates that the permeability and integrity of the endothelial surface is sensitive...
to an increase in strain energy density (6). The magnitude of the energy stored in the wall is determined, among other things, by the geometry and rheology of the vessel wall. The present study was undertaken to provide details of vascular mechanics necessary for further insight into this problem.

Methods

Fourteen dogs weighing 24 to 33.9 kg (average weight 27.7 kg) were studied under chloralose and urethane anesthesia (about 65 and 650 mg/kg). The left circumflex coronary artery (LCCA) and the right common carotid artery (CA) were sufficiently exposed so that a reasonably uniform segment could be measured and marked. During this time, the aortic pressure was continuously monitored. Care was taken to place the head and front legs of each dog in approximately the same way (the dog lying supine with the front legs tethered to the table near his head). Two plastic cylindrical plugs were selected to fit snugly into the lumen at each end of the vessel segment. The plugs were then introduced into the vessel lumen and coupled to the vessel wall by tying sutures externally around the vessel over the circumferential grooves in the plugs. The details of this procedure have been described elsewhere (1, 7). The segment was then removed from the dog and stripped of its surrounding tissues up to the adventitia.

Figure 1 depicts the experimental set-up. A hollow metal rod was screwed into a threaded hole through the center of the proximal plug so that the intravascular pressure could be changed; a solid metal plug containing a hook was screwed into the distal plug so that weights could be hung from it. The vessel segment (A) was mounted vertically on a ring stand so that its proximal end (upper end in Fig. 1) was fixed. The hollow metal rod (N) connected the vessel to a reservoir (Res) filled with 6% dextran solution kept at room temperature (26 to 28°C). The reservoir was made airtight and connected to a source of air pressure which could be adjusted by using the bleeder hose and a screw clamp (S). This pressure was monitored by a P-23 Dp Statham transducer (P) whose zero pressure reference was set at the midsegment level by adjusting the height of the reservoir. A centimeter scale (SC) divided in half millimeter increments was mounted on a stand behind the segment. The length (i.e., the distance between the ties coupling the blood vessel to the plugs) and midsegment external diameter of the blood vessel segment were measured using the microscope system (M), which was equipped with a Bausch and Lomb Filar micrometer (12.5X) eyepiece (E).

The microscope system was mounted on a stand which was adjustable in both the horizontal and vertical directions. The eyepiece had a movable hairline, controlled by a vernier knob, which could be used to average the vessel diameter over a length of approximately 4 mm. The combination of the eyepiece reticule and vernier made it possible to make measurements to within 0.006 cm. An insert of the view through the microscope is shown in the lower right-hand corner of Figure 1. The overall error in measurement of
diameter was estimated by repeatedly measuring the carotid and the coronary arteries at several given pressures. The average value for the error was \(<3\%\). To measure length, the microscope was used in conjunction with the scale.

In general, the experimental measurements consisted of adjusting the pressure to a given value, waiting 1 minute for stress-relaxation to occur (1) and then measuring length, \(L\), and diameter, \(D\). Throughout the experiment the vessel was kept moist, externally, by dripping dextran (at room temperature) over it. At the end of the experiment the blood vessel segment was slit open longitudinally and the unstressed length \(L_0\) and circumference were measured. The value of the unstressed radius \(R_0\) was obtained by dividing the value of unstressed circumference by \(2\pi\). The specimen was then weighed and the wall volume, \(V\), was calculated by

\[
V = \frac{\text{wt. in g}}{\text{tissue density in g/cm}^3}\]  

(1)

where the value of 1.06 g/cm\(^3\) was used for tissue density (8). Since the material is incompressible (9), \(V\) is a constant and the values of thickness, \(h\), could be calculated for different values of \(D\) and \(L\) from

\[
h = \frac{D}{2} - \sqrt{\frac{D^2}{4} - \frac{V}{\pi L}}.\]  

(2)

Subtracting the appropriate value of \(h\) or \(h/\delta\), respectively, from the measured external radius, \(R_e\), it was possible to calculate the values of internal radius, \(R_i\), or midwall radius, \(R_m\), as required.

EXPERIMENTS TO STUDY PRESSURE-RADIUS RELATIONSHIPS

The objective of the experiment was two-fold: (1) to minimize hysteresis effects in order to obtain reproducible results and (2) to collect necessary data for computation of various elastic parameters.

LCCA and CA are similar to the aorta in that under static distention the pressure-diameter curve does not follow the same path over an inflation and deflation cycle (i.e. a hysteresis loop is obtained). Prior to each experiment, this effect was minimized by inflating and deflating the vessel for two cycles over a pressure range of 0 to 200 \(\times 10^6\) dyne/cm\(^2\). During this time the vessel was held at a constant in-vivo length by an inextensible string (St in Fig 1) attached to the hook on the distal plug and the base of the stand.

With the length still held fixed a pressure diameter curve was then obtained. Here and in all subsequent measurements, data were collected from the ascending limb of the pressure-diameter curve as the pressure was raised to cover the physiologic range. Care was taken to include diameters obtained at end-diastolic, mean, and peak systolic pressures. From these data it was possible to estimate the percent change in radius around the mean value of the midwall radius, \(R_m\), from

\[
\% \text{ change in } R = \pm \frac{\Delta R}{R_m} \times 100.\]  

(3)

where \(\Delta R\) is the change in \(R\) corresponding to a given pulse pressure, \(\Delta P\). The volume distensibility was calculated from

\[
\text{Volume distensibility} = \frac{\Delta v}{v \times \Delta P}\]  

(4)

where \(\Delta v\) is the change in lumen volume from the diastolic value \(v\). The lumen volume \(v\) was calculated using the formula for a cylinder,

\[
v = \pi R_i^2 L.\]  

(5)

CALCULATION OF INCREMENTAL ELASTIC MODULI

Arteries are known to have anisotropic elastic properties, i.e., the elastic properties are different in different directions (1, 4, 10, 11). In addition, they demonstrate elastic symmetry (7) about the principal axes of stress, and the wall material is incompressible (9). For such a material it is possible to determine experimentally the incremental elastic moduli around a given state of strain from the following equations (12):

\[
\text{Elastic Properties, Vol. XXVII, August 1970}\
\]
where $E_r$, $E_g$, and $E_l$ are incremental Young's moduli in radial, $r$, circumferential, $\theta$, and longitudinal, $z$, directions; $e_r$ and $e_z$ are the incremental strains in circumferential and longitudinal directions; $P_r$, $P_e$, and $P_z$ are incremental stresses in the appropriate directions which can be calculated from intravascular pressure and longitudinal force measurements (1). $e^*$ and $P^*$ are incremental strain and stress for the special case when $P_e$ and $P_r$ are made zero by experimental manipulations.

The following experimental procedure was carried out in two steps to obtain the necessary data to solve equations 6 for $E_r$, $E_z$, and $E_l$:

1. At a constant in-vivo length, the pressure in the blood vessel segment was varied $\pm 20 \times 10^5$ dyne/cm$^2$ around the in-vivo mean pressure, $P_m$. The length was held constant during this procedure by adjusting the weights (W in Fig. 1) after the pressure was changed.

2. The length was then decreased about 5% by reducing the weight on the segment, holding $P_m$ essentially constant (within $5 \times 10^3$ dyne/cm$^2$ such that the value of $P_e$ and $P_r$ would be negligible (1)). During this maneuver, the values of pressure, external diameter and the weights hung on the segment were recorded. From these data, it was possible to solve equations 6.

The values of $e_r$, $e_z$, $P_r$, $P_e$, and $P_z$ were calculated using data obtained from step 1 of the experimental procedure:

$$e_r = \frac{1}{E_r} \left[ -\frac{P_r + P_e}{2} \right]$$

$$e_z = \frac{1}{E_z} \left[ -\frac{P_z + P_e}{2} \right]$$

$$e^* = \frac{1}{E^*} P^*$$

where $E_r$, $E_z$, and $E_l$ are incremental Young's moduli in radial, $r$, circumferential, $\theta$, and longitudinal, $z$, directions; $e_r$ and $e_z$ are the incremental strains in circumferential and longitudinal directions; $P_r$, $P_e$, and $P_z$ are incremental stresses in the appropriate directions which can be calculated from intravascular pressure and longitudinal force measurements (1). $e^*$ and $P^*$ are incremental strain and stress for the special case when $P_e$ and $P_r$ are made zero by experimental manipulations.

The values of $e^*$ and $P^*$ were calculated using data obtained from step 2 of the experimental procedure:

$$e^* = -\frac{\Delta L}{L}$$

where $\Delta L$ represents the change in length from the mean length, $L$,

$$P^* = \frac{F}{2\pi R_h} \left[ L - \frac{F}{2\pi R_h} \right]$$

where $F$ is the force in dynes due to the weight (W in Fig. 1) hanging on the vessel segment.

The value of $e_r$ for a constant length was zero. $P_r$ was approximated by

$$P_r = -\frac{\Delta P}{2}$$

and

$$P_e = \frac{P_e}{2} + \frac{F}{2\pi R_h} - \frac{F}{2\pi R_h}$$

where 1 and 2 refer to the evaluation of the bracketed quantities at the initial and final values of the pressure respectively.

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roundings, the strain energy (SE in dyne-cm) can be estimated from the external work performed on the blood vessel using the following formula:

\[ SE = \int_{v_i}^{v_f} P \, dv + \int_{L_o}^{L_f} F \, dL \]  

(13)

where \( P \) is the inflation pressure, in dyne/cm\(^2\), \( dv \) is the change in lumen volume in cm\(^3\), \( v_o \) is the unstressed lumen volume and \( v_f \) is the final lumen volume (i.e., the in-vivo volume). \( F \) is the force in dynes due to the weight (\( W \) in Fig. 1) hung on the vessel segment, \( dL \) is the change in length in cm, \( L_o \) is the initial length and \( L_f \) the final in-vivo length. It can be shown (13), under the conditions specified above, that this integral is independent of path, and is therefore not affected by the order in which the vessel is loaded.

The experimental procedure to evaluate SE consisted of hanging the vessel segment as in Figure 1 with the length unconstrained and varying the pressure in steps over the range 0 to the mean in-vivo value, \( P_m \), and recording the diameter and length at each step. Care was taken to vary the pressure slowly in order to approximate a quasi-static process. Once the value of \( P_m \) was reached, weights were added in steps, until the length was restored to its in-vivo value. Again at each step, the values of diameter, length and weight were recorded. This procedure was repeated in a slightly modified manner in that the order of loading was changed: after the segment was pressurized to about \( 50 \times 10^3 \) dyne/cm\(^2\), weights were added, in steps, until the total weight used above was reached, and then the vessel was pressurized in steps to \( P_m \).

The lumen volume, \( v \), of the vessel segment was computed from the values of \( R_t \) and \( L \). Plots of pressure vs. volume and force vs. length were then prepared as shown in Figures 2A and 2B and the areas under these curves were obtained using a planimeter.
These areas represent the values of the two definite integrals in equation 13, and their sum is equal to the strain energy (SE). The value of SED was then calculated by dividing SE by the wall volume, V.

Results and Discussion

HYSTERESIS

Results from a typical hysteresis experiment from LCCA are shown in Figure 3 as pressure-volume curves. Maximum hysteresis occurred during the first inflation-deflation cycle; the hysteresis loop for the second cycle was narrower. Similar observations have been reported by Bergel (3) and Remington (14) for other arteries. Hysteresis was quantified by comparing the width of the hysteresis loop measured as a change in lumen volume at midpressure, \( \delta V \), to the change in lumen volume over the range of pressure used, \( \Delta \).

The average values of this ratio \( \delta V / \Delta \) for LCCA from four dogs were 0.155 and 0.048, respectively, for the first and the second hysteresis loops; similar values for CA from four dogs were 0.110 and 0.052. From these data it was concluded that most of the hysteresis effect was removed by the first two inflation-deflation cycles.

PRESSURE-RADIUS RELATIONSHIP

Pressure-radius relationships covering the range of in-vivo pulse pressures were obtained at constant length both in LCCA and CA subsequent to the removal of hysteresis effects. Figure 4 illustrates this relationship from one dog. It is interesting to note that the relationship is essentially linear over this pressure range. Similar relationships were also obtained for other dogs. The data from all experiments are summarized in Table 1, Group 1, showing average values ±SE for various parameters. It is important to note that although the value of \( R_{\text{max}} \) in LCCA is...
smaller than $R_{max}$ in CA, $\Delta R / \Delta p$ in LCCA is greater than $\Delta R / \Delta p$ in CA, indicating that LCCA is more distensible than CA.

It was possible to compare our data from LCCA with the following studies: (1) Our value for volume distensibility from LCCA ($3.41 \times 10^{-6}$ cm$^3$/dyne evaluated around a mean pressure of $104 \times 10^3$ dyne/cm$^2$) is in reasonable agreement with a similar value of Douglas and Greenfield $^8$ ($4.39 \times 10^{-6}$ cm$^3$/dyne evaluated around a mean pressure of $126 \times 10^3$ dyne/cm$^2$) when one considers that, in general, arteries are more distensible at lower pressures. (2) The value of volume distensibility for the left anterior descending coronary artery of dogs calculated from the data of Gregg et al. ($2$) around $185 \times 10^3$ dyne/cm$^2$ pressure is $4.03 \times 10^{-6}$ cm$^3$/dyne which is $18\%$ higher than ours.

It was possible to compare our data from CA with similar data obtained by Attinger et al. ($15$), Bergel ($3$), Dobrin et al. ($16$), and Peterson et al. ($5$) using different techniques for measurement of radius (Table 2). For this purpose, the value of $E_{iso}$ was calculated using the formula (17):

$$E_{iso} = 0.75 \frac{R}{h} \frac{R^2}{1+R/h},$$

where $E_{iso}$ is the "isotropic incremental elastic modulus" as defined by Bergel. Although $E_{iso}$ does not describe true material properties of the anisotropic arterial wall ($1$), it was used here as a convenient means for comparison of data. It can be seen in Table 2, that our results agree reasonably well with those of Bergel, Dobrin and Attinger. Peterson's value for $E_{iso}$ on the other hand, is higher, i.e., he finds the arterial wall to be stiffer. The reason for this discrepancy may be as follows: Peterson et al. were evaluating dynamic elastic properties in the living dog, and in general the values of in-vivo dynamic elastic moduli are higher than the in-vitro static values ($1, 18$).

**ANISOTROPIC ELASTIC MODULI**

The values of the incremental elastic moduli in the circumferential, longitudinal and radial directions were estimated for LCCA and CA as explained in Methods. The pertinent results are shown in Table 1, Group 2. Since the overall properties of an artery are non-linear ($1, 4, 10$), a meaningful comparison of local properties, from different studies, can only be made at corresponding values of average state of strain, which can be inferred from the values of extension ratios, $\lambda_c = \frac{L}{L_0}$ and $\lambda_o = \frac{R}{R_0}$ included in the table.

It is important to point out that these estimates of elastic moduli are considered crude for the following reasons: (1) These arteries are viscoelastic, and the viscous effects, though minimized, were not completely eliminated. (2) Ratios $R/h$ tend to be small and could introduce errors in the calculation of wall stress based on thin-wall theory ($8$). (3) The computations are sensitive to errors in measurement of radius and to errors due to our inability to keep length exactly constant.
### Summary of Data

#### Group I

<table>
<thead>
<tr>
<th>Parameter</th>
<th>LCCA</th>
<th>CA</th>
<th>MDTA</th>
</tr>
</thead>
<tbody>
<tr>
<td>No. dogs</td>
<td>7</td>
<td>5</td>
<td></td>
</tr>
<tr>
<td>PM (dyne/cm²) × 10⁻³</td>
<td>164 ± 8</td>
<td>168 ± 15</td>
<td></td>
</tr>
<tr>
<td>L (cm)</td>
<td>1.97 ± .09</td>
<td>3.63 ± 11</td>
<td></td>
</tr>
<tr>
<td>PM (cm²)/dyne × 10⁻³</td>
<td>0.145 ± .01</td>
<td>0.223 ± .004</td>
<td></td>
</tr>
<tr>
<td>ΔR/ΔP (cm/dyne) × 10⁶</td>
<td>44 ± 5</td>
<td>49 ± 1</td>
<td></td>
</tr>
<tr>
<td>AR/AP (cm'/dyne) × 10⁺</td>
<td>0.192 ± .032</td>
<td>0.164 ± .017</td>
<td></td>
</tr>
<tr>
<td>% change in R</td>
<td>2.5 ± .38</td>
<td>14 ± 17</td>
<td></td>
</tr>
<tr>
<td>Volume distensibility (cm³/dyne) × 10⁸</td>
<td>3.41 ± .61</td>
<td>1.50 ± .20</td>
<td></td>
</tr>
</tbody>
</table>

#### Group II: Anisotropic Elastic Moduli

<table>
<thead>
<tr>
<th>Parameter</th>
<th>LCCA</th>
<th>CA</th>
<th>MDTA</th>
</tr>
</thead>
<tbody>
<tr>
<td>No. dogs</td>
<td>5</td>
<td>5</td>
<td></td>
</tr>
<tr>
<td>PM (dyne/cm³) × 10⁻³</td>
<td>164 ± 5</td>
<td>176 ± 9</td>
<td></td>
</tr>
<tr>
<td>R/Rh</td>
<td>9.71 ± .53</td>
<td>9.99 ± 1.18</td>
<td></td>
</tr>
<tr>
<td>λₚ</td>
<td>1.33 ± .95</td>
<td>1.53 ± .94</td>
<td></td>
</tr>
<tr>
<td>λₚ</td>
<td>1.58 ± .99</td>
<td>1.36 ± .67</td>
<td></td>
</tr>
<tr>
<td>Eₚ (dyne/cm³) × 10⁻⁴</td>
<td>7700 ± 2200</td>
<td>14600 ± 3100</td>
<td></td>
</tr>
<tr>
<td>Eₚ (dyne/cm³) × 10⁻⁴</td>
<td>35000 ± 10000</td>
<td>8700 ± 2000</td>
<td></td>
</tr>
</tbody>
</table>

#### Group III: Strain Energy Density (SED)

<table>
<thead>
<tr>
<th>Parameter</th>
<th>LCCA</th>
<th>CA</th>
<th>MDTA</th>
</tr>
</thead>
<tbody>
<tr>
<td>No. dogs</td>
<td>10</td>
<td>7</td>
<td>8</td>
</tr>
<tr>
<td>PM (dyne/cm³) × 10⁻³</td>
<td>165 ± 6</td>
<td>168 ± 11</td>
<td>157 ± 7</td>
</tr>
<tr>
<td>R/Rh</td>
<td>1.33 ± .94</td>
<td>1.55 ± .97</td>
<td>1.41 ± .04</td>
</tr>
<tr>
<td>λₚ</td>
<td>1.71 ± .10</td>
<td>1.44 ± .93</td>
<td>1.62 ± .50</td>
</tr>
<tr>
<td>λₚ</td>
<td>314 ± 50</td>
<td>226 ± 40</td>
<td>332 ± 25</td>
</tr>
</tbody>
</table>

Average values ± SE are shown. LCCA = left coronary circumflex artery; CA = right common carotid artery; MDTA = middle descending thoracic aorta; PM = mean pressure; L = length; Rₙ₉ₙ = mean value of midwall radius; ΔR is the change in midwall radius, ΔP for a corresponding change in pressure, ΔP; h = thickness. λₚ and λₚ are the extension ratios in longitudinal and circumferential directions. Eₚ and Eₚ are the incremental elastic moduli in circumferential and longitudinal directions.

Note: 1 cm H₂O pressure = 980 dyne/cm²

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During step 1 of the experimental procedure, this is reflected in the large scatter in the values of the elastic moduli shown in Table 1, Group 2. (4) It was not technically feasible to confirm independently the value of Eₚ for LCCA or CA in a direct manner on a test stand as was done for the thoracic aorta in a previous study (19), because of the relatively small size of these arteries. However, when a direct test was attempted on two specimens of CA, the value of Eₚ was grossly underestimated; Eₚ = 5500 × 10⁴ dyne/cm² by the direct method as compared to 11,100 dyne/cm² by the method used in the present study.

In spite of these reservations, it is reasonable to conclude that: (1) Both LCCA and CA demonstrate anisotropic elastic behavior in the physiologic range. (2) LCCA is stiffer in the longitudinal direction than in the circumferential direction (P < .05). (3) CA is stiffer in the circumferential direction than in the longitudinal direction (P < .05). A similar result has been recently reported by Dobrin et al. (20); it was also predicted by Anliker et al. (11) from their wave transmission studies. (4) In comparison with a previous study (1) LCCA was found to be as distensible as the thoracic aorta in the circumferential direction. As mentioned earlier, the incremental anisotropic elastic moduli are functions of initial stretch (1), which in turn is influenced by the position of the dog’s head as well as the front legs in the case of CA for the same intravascular pressure. All our CA studies were carried out with the dog lying supine.
with the front legs raised and tethered to the table near his head. In one dog, the stretch in CA increased 7% when the front legs were untethered from the table and raised to simulate the normal standing position. The value of the longitudinal modulus, $E_l$, increased as predicted, but not enough to alter the relationship $E_l > E_z$.

## STRAIN ENERGY DENSITY (SED)

SED was calculated as described in Methods for LCCA and CA. In addition, experiments were also carried out on the middle descending thoracic aorta (MDTA) of eight dogs to determine the value of SED for a large artery. The values for SED reported in Table 1, Group 3, could be reproduced within 5% for LCCA and CA and within 1.45 for MDTA even when the loading procedure was altered. Thus the theoretical prediction that the integrals in equation 13 are independent of path was confirmed experimentally. Moreover this finding justified the neglect of the kinetic energy and heat terms in calculation of SE from equation 13.

Since, for an elastic material, SED is a unique function of stress or strain (13), it may be considered a useful parameter to describe elastic properties of the artery. However, a valid comparison between various arteries can only be made at equal values of $\lambda$ and $\lambda_s$. SED may also play an important role in the mechanics of protein (e.g., lipoproteins) transport across the vascular interface as suggested by Fry (6).

In conclusion, we have studied the static elastic properties of the left circumflex coronary artery and the right common carotid artery in dogs at in-vivo values of pressure and length. The pressure-radius relationship at constant length, incremental static anisotropic elastic moduli, and the strain energy density have been evaluated. The initial dimensions are included so that one could locate precisely the point where the local properties were studied on the large nonlinear stress-strain curve. This is important for comparison of data from various sources.

### References


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