Variability of the Viscoelastic Constants Along the Aortic Axis of the Dog

By Richard W. Lawton, M.D.

Measurements of the elasticity of isolated aortic specimens are complicated by the nonlinear mechanical behavior of such tissues. Nonlinearity occurs in both static and dynamic measurements. Krafka, in a series of studies on the static elasticity of isolated aortic strips, computed elastic moduli, each at a different elongation. These moduli differed from each other and varied independently under different conditions. Under dynamic conditions, elastic moduli may be several times greater than static elastic moduli at the same mean elongation. Damping in such specimens appears to be frequency dependent.

Nonlinearity of the mechanical properties of blood vessel walls has been attributed to a complex interaction among the various fiber types present. King, on the other hand, has interpreted the static mechanical behavior of the aortic wall in terms of modern high polymer theory. Recently, this latter interpretation has been extended to include dynamic data. The justification for the application of the polymer theory rests, in part, upon experimental observations that the elastic force in isolated aortic specimens is entropic over a considerable range of elongation. Anomalous thermoelasticity, nonlinear stress-strain curves and long range reversible extensibility are characteristics usually associated with rubber-like materials. Many biological tissues show qualitatively these same characteristics.

This report presents a study of 47 aortic strips from 34 mongrel dogs, using a polymeric model. A statistical analysis of the variations in dynamic elasticity and damping which occur with location along the arterial axis and during stretch and retraction of aortic strips is presented. The analysis of the mechanical behavior involves rather complex statistical mechanical equations (Appendix). Solution of these equations for a wide range of values of the parameters involved have been obtained on an IBM card calculator and reduced to graphic form. From such graphs, the value of the elastomeric modulus and the damping coefficient can be obtained quickly from the experimental data of each aortic strip.

Method and Results

The experimental method has been presented in detail elsewhere. In brief, a specimen aortic strip is hung from an electromechanical vibrator of variable frequency in a chamber where the environment can be maintained at constant temperature and humidity. With a load on the specimen, the frequency of the vibrator is adjusted until maximum amplitude at resonance is observed by means of an oscilloscope. If the amplitude of vibration is less than 5 per cent of the specimen length, the method yields satisfactory data. For the observations reported here, an aortic strip usually had been freshly cut and stored in the icebox for 24 to 36 hours prior to an experiment. All specimens were cut parallel to the long axis of the vessel and were approximately 1 cm. wide by 5 cm. long. Immediately before mounting in the chamber, a specimen was given about 10 cycles of stretch and retraction, by hand, at the rate of 1 per second. The experimental stretch and retraction cycle involving both static and dynamic measurements occupied about 30 minutes.

In figure 1, at the top, is the static force-length data and on the bottom the dynamic data for a
specimen of dog thoracic aorta. The dynamic data are displayed as a plot of \( \omega^2L_o/g \) against \( \alpha \) where \( \omega \) represents \( 2\pi \) times the natural frequency of resonance for the system, \( L_o \) is the unstretched length of the specimen and \( g \) is the gravitational acceleration. Here \( \alpha \) is the relative length \( L/L_o \), where \( L \) is the length of the stretched or loaded specimen. Stretch data are shown as crosses and retraction data as circles.

The quantity \( \omega^2L_o/g \) for a nondissipative system is a nondimensional function of the relative length \( \alpha \) and an elastomeric parameter \( \beta \) (Appendix). Values of this parameter are determined uniquely from the relative length \( \alpha \) associated with the minimum frequency. With increasing values of \( \beta \) the frequency minimum moves upward and to the left and is associated with smaller values of \( \alpha \). The parameter may be founded graphically from a plot of \( \alpha_{\text{min}} \) vs. \( \beta \). The data for such a curve are tabulated in table 1. The parameter \( \beta \) is defined as the ratio of the unstretched length of the macromolecular chain \( L_o \) to the maximum extension \((L_{\text{max}} - L_o)\) which it can undergo. In figure 1, both static and dynamic curves have the same value of \( \beta \), namely, 0.95. Thus, a single number characterizes the elastic behavior over a wide range of elongations and under both static and dynamic conditions.

Because of internal viscous damping, however, the theoretical values of \( \omega^2L_o/g \) are smaller than the observed values. If a fraction of the macromolecular chain \((1 - \gamma)\) is subject to viscous effects, then the remaining nondamped fraction enters into the vibratory motion of the specimen. This consideration has the effect of raising the value of \( \omega^2L_o/g \) by a factor \( 1/\gamma \) without any shift along the \( \sigma \)-axis. For the specimen of figure 1, \( \gamma \) equals 0.75.

In figure 1, the first stretch-retraction cycle following the hand stretching procedure shows evidence of hysteresis. Additional cycles may bring about nearly complete reversibility. The retraction phase differs from the stretch phase in figure 1 by having somewhat lower resonant frequencies and by a larger final value for \( L_o \). If \( \omega^2L_o/g \) and \( \alpha \) for the stretch data are computed from the initial value of \( L_o \) and, for the retraction data, from the final value of \( L_o \), the dynamic response is still clearly nonreversible, but the static curves are made to almost coincide, as can be seen in figure 1. This method of analysis is adopted for the following calculations. Actually, the hysteresis is reduced only a small amount, inasmuch as the change in \( L_o \) usually is less than 5 per cent. The preliminary stretching by hand also increases \( L_o \) and so reduces the hysteresis effect.

Some of the results of this method of calculation and qualitative information on the precision of the measurements are given in table 2. An aortic strip from each of 2 dogs was carried through a series of stress-strain cycles. For the first cycle, the strip was 6 cm. long. For each succeeding cycle a 1 cm. piece was cut off between runs. Values of \( \alpha_{\text{min}} \) and \( (\omega^2L_o/g)_{\text{min}} \) are tabulated for both the stretch and retraction phases. The 2 sets of results for retraction are obtained by introducing the 2 different values of \( L_o \) as measured before and after a run. Comparing these results, we see that \( \alpha_{\text{min}} \) is slightly larger and \( (\omega^2L_o/g)_{\text{min}} \) is slightly smaller for the results derived from the initial value of \( L_o \). Variations in both these quantities with progressive decreases in strip length probably arise from the inherent reduction in accuracy of data obtained from shorter strips and also from drying and deterioration of the specimen over the long course of the experiment.
VISCOELASTIC CONSTANTS FOR AORTA

Statistical Analysis

The variation in parameter \( \beta \) and fraction \( \gamma \) of the vessel walls were studied as a function of position along the central arterial axis. Measurements were made during loading and unloading of 47 aortic strips from 34 dogs. Mean values of \( \beta \) and analysis of variance for them are entered in table 3. Values of \( \beta \) were obtained from values of \( \alpha \) corresponding to minimum resonant frequencies during loading or elongation of the strips and during unloading or retraction. The temperature range for all the data is 25 to 30 C. The mean \( \beta \)-values for the arterial strips are arranged according to the location from which they were obtained and according to the experimental conditions (stretch or retraction). The analysis of variance for unequal samples was done by means of a method outlined by Snedecor.\textsuperscript{13} The F-value for variations in \( \beta \) with location and with condition, and their interaction, are considered significant at the 5 per cent level. Only variations because of location appear to be significant. Four degrees of freedom can be obtained by adding the location-condition sum of squares to the error term, and then the F-value for location becomes significant at the 1 per cent level. The error term in this analysis is the variance within subclasses. It is approximately 0.05, so that the standard deviation is about 0.23 \( \beta \)-units.

In table 4, a similar analysis is given for values of \( \gamma \). Variations with location along the arterial axis and with experimental conditions (stretch or retraction) are significant at the 1 per cent level. The interaction between location and condition is not significant. The standard deviation equals 0.11.

Among the aortic strips for the foregoing analysis, no 2 are from the same dog. In order to study location effects within an individual dog, therefore, specimen strips were taken from the aortic arch and from the thoracic aorta in each of 5 dogs and from the thoracic and abdominal aortas in each of 7 dogs. In all cases, dog variability was significant at the 1 per cent level. The F-values for variation in \( \beta \) with location and condition were insignificant, although a significant location-dog interaction was present for the aortic arch-thoracic aorta pairs. Variation in \( \gamma \) with location was insignificant, but with condition the F-value was significant at the 1 per cent level. A significant location-dog interaction for \( \gamma \) was present in both groups of dogs. These data were also analyzed for location, using small sample nonparametric methods\textsuperscript{14, 15} with similar statistical significance. These results indicate that over the short paths from the arch to the descending thoracic aorta
Variations in $\alpha$ and $(\omega^2 L_0/\rho)$ Retraction Minima for Two Estimates of $L_0$

<table>
<thead>
<tr>
<th>Location</th>
<th>$\alpha_{\text{min}}$ (cm$^{-1}$)</th>
<th>$(\omega^2 L_0/\rho)_{\text{min}}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thoracic aorta</td>
<td>$S_1$</td>
<td>$R_1$</td>
</tr>
<tr>
<td>Dog no. 1</td>
<td>6.0</td>
<td>1.50</td>
</tr>
<tr>
<td></td>
<td>5.0</td>
<td>1.43</td>
</tr>
<tr>
<td></td>
<td>4.0</td>
<td>1.46</td>
</tr>
<tr>
<td></td>
<td>3.0</td>
<td>1.42</td>
</tr>
<tr>
<td>Dog no. 2</td>
<td>6.0</td>
<td>1.60</td>
</tr>
<tr>
<td></td>
<td>5.0</td>
<td>1.63</td>
</tr>
<tr>
<td></td>
<td>4.0</td>
<td>1.54</td>
</tr>
<tr>
<td></td>
<td>3.0</td>
<td>1.53</td>
</tr>
</tbody>
</table>

Table 3

Analysis of Variance of Values of $\beta$

<table>
<thead>
<tr>
<th>Location</th>
<th>df</th>
<th>Sums of squares</th>
<th>Variance</th>
<th>$F$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stretch</td>
<td>4</td>
<td>0.7107</td>
<td>0.1777</td>
<td>3.46*</td>
</tr>
<tr>
<td>Retraction</td>
<td>84</td>
<td>4.3836</td>
<td>0.0522</td>
<td>0.09</td>
</tr>
<tr>
<td>Location</td>
<td>4</td>
<td>0.7107</td>
<td>0.1777</td>
<td>3.55t</td>
</tr>
<tr>
<td>Condition</td>
<td>1</td>
<td>0.0740</td>
<td>0.0740</td>
<td>1.48</td>
</tr>
<tr>
<td>Error</td>
<td>88</td>
<td>4.4010</td>
<td>0.05</td>
<td></td>
</tr>
</tbody>
</table>

$*P < 0.05.$

$*P < 0.01.$

and from the descending thoracic aorta to the abdominal aorta, variations in elasticity and damping are not significant for small samples.

In the larger samples of tables 3 and 4, location has been shown to be significant at the 1 per cent level for both $\beta$ and $\gamma$. We may now inquire as to which of the location means are significantly different from one another. This is done by the $t$-test. For the stretch mean values of $\beta$ and $\gamma$, significance is found only in the extremes of the data. During stretch, carotid and abdominal $\beta$-means differ at the 5 per cent level, and arch and iliac $\gamma$-means differ at the 1 per cent level. In figure 2, the $\beta$ and $\gamma$ retraction mean values are arranged in their orders of magnitude. Means differing at the 1 per cent level are connected by a solid line, and those differing at the 5 per cent level are connected by a dashed line. Notice that, with the exception of the carotid, the $\gamma$-means are arranged in their anatomic order along the arterial axis.

This analysis indicates that the elasticity and damping within segments of the upper aorta (arch and descending thoracic), and within segments of the lower aorta (abdominal and iliac), cannot be distinguished. There are, however, significant differences in $\beta$ and $\gamma$ between segments from the upper and lower aorta. The carotid artery appears to occupy an anomalous position, having elastic properties associated with the larger upper aorta and damping associated with the more muscular vessels of the smaller lower aorta.

Geometric Considerations

In the course of these experiments, measurements of the unstressed wall thickness and the unstressed internal circumference were made on the excised specimens. The means for vessel thickness, $e_0$, and internal radius, $r_0$, are shown in table 5, arranged according to specimen location. Thickness measurements were made in a uniform manner, using a stage micrometer. These measurements involve slight compression of the specimen, so that values are perhaps somewhat less than the true thickness. A progressive decrease in thickness and radius from the arch of the aorta to the iliac artery is demonstrated. The product $e_0 r_0$, which equals the cross-sectional area of the tube wall divided by $2\pi$, progressively decreases with distance from the heart along the arterial axis. The values of $e_0 r_0$ in table 5 may be compared with those computed by Bazett, Cotton, LaPlace, and Scott, from Kani's data. They obtained values from 0.21 to 0.42 for 20-year-old human aortas, carotids, and iliacs post mortem. Whereas they found $e_0 r_0$ increased in value with distance from the heart in humans, we observe no significant similar correlation in dogs. The mean values...
of \( e_0/r_0 \) for the thoracic aorta and the carotid artery in table 5 differ at the 10 per cent level by t-test.

Effective values for the average static Young's modulus (\( E_s \)) and the average dynamic modulus (\( E_d \)) have been computed for the various locations and are given in table 5. Equations for these calculations are given elsewhere. Dynamic moduli are approximately double the static moduli. The 2 largest values for the dynamic modulus correspond to the smallest and largest values of \( \beta \). Calculation of \( E_s \) and \( E_d \) involve errors in the measurement of cross-sectional area which may account in part for the large variations. In computing the static moduli, a value of 0.5 is assumed for Poisson's ratio. Values of static moduli, given by Krafa\(^1\) for the dog, compare favorably when similarly corrected.

Correlations between measures of mechanical behavior (\( \beta \) and \( \gamma \)) and measures of blood vessel geometry (\( e_0 \) and \( r_0 \)) were studied. A highly significant correlation between radius and thickness was found (0.83). There was no significant correlation between \( \beta \) and \( \gamma \), and the \( \beta \)-values were not correlated with either \( e_0 \) or \( r_0 \). Damping, however, is correlated with vessel radius at the 5 per cent level.

In general, the \( \alpha \)-value of an elastomeric strip at a frequency minimum can be related to the relative radius at the inflection point of the pressure-volume curve for the tube from which the strip is taken. By means of equation 3, Appendix, therefore, mean arterial pressures at such points have been computed for the aortas reported here. For this purpose, the experimental mean values of \( \beta \), \( e_0/r_0 \), and initial cross-sectional area \( S_0 \), were introduced in equation 2, Appendix, to compute \( \Lambda \). The computed pressure values at the several locations along the aorta are also given in table 5. The over-all pressure average is 106 ± 14 mm. Hg. This value is somewhat higher than the normal mean arterial pressure for the dog.

**Discussion**

The application of a simple relation, such as Hooke's law, to study the complex nonlinear elastic behavior of aortic strips, cannot be justified except as a matter of convenience. The assumption of linearity over a limited region leads to the calculations of an elastic constant, not generally useful over a wide range of elongations and which is different for static and dynamic conditions. The end result is a series of numbers, rather than a single constant, characterizing the specimen. A persuasive reason for adopting an elastomeric model for the description of the mechanical behavior of aortic strips is that a single constant serves to characterize the elastic behavior over a wide range of elongations and under both static and dynamic conditions.

The elastomeric model presented here is far too simple for a full description of the mechanical behavior of isolated aortic strips, much less the intact aorta. In a study of the thermodynamics of aortic strips,\(^1\) it was shown that, over a rather wide range of elongations, a negative internal energy term is present. The model we have employed does not account for this. At high elongations, the internal energy becomes positive and again the model does not include this effect. For mechanochemical systems, such as the reaction of myosin filaments with ATP, simple elastomeric theory has been extended to include electrostatic and polymer-solution mixing effects.\(^1\) Although elastomeric theory meets the problem of nonlinearity, other models are
Table 4

Analysis of Variance of Values of $\gamma$

<table>
<thead>
<tr>
<th>Location</th>
<th>Arch</th>
<th>Thoracic</th>
<th>Abdominal</th>
<th>Iliac</th>
<th>Carotid</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stretch</td>
<td>0.62</td>
<td>0.59</td>
<td>0.55</td>
<td>0.48</td>
<td>0.54</td>
</tr>
<tr>
<td>Retraction</td>
<td>0.70</td>
<td>0.68</td>
<td>0.61</td>
<td>0.57</td>
<td>0.60</td>
</tr>
<tr>
<td>Location</td>
<td>4</td>
<td>0.2298</td>
<td>0.0575</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Condition</td>
<td>1</td>
<td>0.1992</td>
<td>0.1192</td>
<td></td>
<td></td>
</tr>
<tr>
<td>L x C</td>
<td>4</td>
<td>0.0139</td>
<td>0.0035</td>
<td></td>
<td></td>
</tr>
<tr>
<td>W/in subclass</td>
<td>84</td>
<td>1.1039</td>
<td>0.0131</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Total</td>
<td>93</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

*P < 0.01.

available which also have this feature. Perhaps the most widely studied is the "long-short" conversion model. This latter model is almost completely nonspecific concerning molecular mechanisms.

In the application of elastomeric theory, no assumptions are made concerning the role of the individual fibers in the aortic wall. In fact, mechanical homogeneity is assumed here, as is done when applying Hooke's law. Actually, there is evidence that the mechanical behavior of circumferential and axial aortic strips differ, the former being slightly more extensible than the latter. How this effect varies along the arterial axis is not known. The fact that the data of this paper was obtained exclusively from axial strips may account in part for the higher than normal mean pressure computed in table 5. The elastic tissue in the dog's aorta is distributed in uniform laminae, which suggests that these differences in extensibility are perhaps due to the distribution of collagen and smooth muscle fibers.

A fundamental assumption of polymer theory is that changes in microscopic or molecular dimensions are proportional to changes in macroscopic dimensions. Thus, it has been suggested that if the molecular chains composing arterial walls all have the same average length, the $\beta$-value for any major vessel might be computed from the $\beta$-value for 1 vessel and the ratio of their undistended internal radii. This suggestion is not supported by the foregoing data. Similarly, one might suppose that longer molecular chains (smaller $\beta$-values) would involve more chain entanglement, and, thus, more internal dissipation in the specimen. The correlation coefficient between $\gamma$ and $\beta$ of the specimens in this study was, however, only 0.03. Thus, the data appear to support a concept of changing composition of the aortic wall with distance from the heart. The change in composition is associated with a change in average molecular dimensions as well as a change in macroscopic size. The damping is independent of the $\beta$-value perhaps in a way analogous to the effects of some fillers and plasticizers in rubber.

The damping observed in these experiments should be distinguished from that associated with hysteresis loops in large amplitude stress-strain cycles, where differential length changes may be as much as 50 to 100 per cent of initial length. In the experiments reported here, little or no over-all static stress-strain hysteresis is observed (fig. 1) and the damping fraction $\gamma$ appears to be characteristic of the specimen only under dynamic conditions. The damping is minimal at the resonance frequency and values of $\gamma$ for a specimen do not markedly change with load.

The differences in dynamic properties observed during stretch and retraction is of interest from the point of view of tissue mechanics rather than from that of normal physiologic function (fig. 1). These changes can be ascribed to changes in molecular entanglement or internal friction as reflected in $\gamma$-values, rather than changes such as molecular rupture or chemical phase change which might alter $\beta$-values. This behavior is perhaps related to molecular changes associated with excision of the specimen and the time lapse before re-extension. When specimens are excised, the molecular chains may contract and become preferentially oriented. Extension of the specimen restores a more random arrangement among the molecules. During the initial stretching, higher values for internal damping are thus found as compared with those for the retraction phase. It seems unlikely that such changes can contribute significantly to the over-all damping of vessels in...
the normal dynamic steady state during the cardiac cycle. In situ behavior probably is more closely represented by the retraction phase than by the elongation phase.

The effects of variation among dogs deserve emphasis. To demonstrate the changes in $\beta$ and $\gamma$ with location requires a rather large number of determinations. The analysis presented here suffers because of the unequal groups representing the different locations and from the lack of replication in each dog, so that in these studies it is principally the extremes of the central arterial axis that can be distinguished mechanically. In situ, the aorta is held in longitudinal tension and restraint, which is conceived to arise from the growth process. In addition, the external loading of the aortic wall by the viscera varies substantially between the thorax and abdomen. Both of these factors play a role in the mechanical behavior of the aorta in situ; therefore, data on isolated aorta specimens must be interpreted with caution.

Summary

The application of elastomeric theory to the study of the viscoelastic properties of the dog's aorta shows that the same value of the elastomeric parameter $\beta$ may be used to describe the elastic behavior of aorta strips under both static and dynamic conditions. The variations in the parameter $\beta$ and the damping fraction $\gamma$ with the strip's location may be distinguished only for the extremes of the central arterial axis because of the wide variability of mechanical properties among dogs. The viscoelastic properties apparently do not correlate with the measured geometric properties of the unstressed aortic strip.

Appendix

A statistical mechanical analysis of the deformation of a macromolecular chain acted upon by a force has been presented by Wall,21 James and Guth,22 and King.23 Consider a macromolecular chain of $N$ links, each of length $l$, so that the total length of the chain is $NL$. Each link on the average makes a contribution in the $x$-direction of an amount $l_0$, so that under the conditions of no load the end-to-end length of the chain is $NL_0$. Suppose that a force $F$ is applied to 1 end of the chain to produce an increase in length. Each link then must make an additional contribution to the specimen length. Formerly12, 22 it was assumed that the link $l$ could contribute any amount between $-l$ and $+l$ by free rotation. Although highly improbable, it was not explicit that such contributions could result in $l_0 = 0$. In other words, such a chain could conceivably turn upon itself or collapse under no load. In a more recent analysis by King,10 which will be followed here, it was assumed that at all loads, each link contributes at a minimum $l_0$. Each link may make an additional contribution of an amount between $-(1 - l_0)$ and $(1 - l_0)$ under a force $F$. Evaluation of this additional term by statistical methods yields the relation between the force $F$ and the relative elongation $\alpha$ for a single elastomeric chain as follows:

$$F = C \xi^{-1} \beta(\alpha-1). \quad (1)$$

In this equation $\beta$ represents $L_0/N \ (l - l_0)$ and may have values greater than unity. $\xi^{-1}(\beta(\alpha-1))$ is the symbol for the inverse Langevin function of the argument $\beta(\alpha-1)$. Values for this function may be found in tabular form elsewhere.12 The coefficient $C = \beta NkT/l_0$, where $k$ is the Boltzmann constant and $T$ is the absolute temperature.

King,10 in an analysis of the force-length relation of a 3-dimensional network of chains, obtained the following relation:

$$F = AS_o \xi^{-1} \{\beta(\alpha-1)^2\} + \alpha^{3/2} \xi^{-1} \beta(1-\alpha^{1/2}) \} \quad (2)$$

where $S_o$ = the initial cross-sectional area and the coefficient $A = M/BNKT/L_0$. Here $M$ represents the average number of chain ends per unit area.
on the surface of the network block of length $L_0$.

Equation 2 is applied in the study of the stress-strain relation in aortic strips (fig. 1).

For a cylindrical tube, the relation between the excess pressure ($p - p_0$) and the relative change in radius $\alpha$ is:

$$p - p_0 = A\left(\frac{\epsilon_0}{r_0}\right)\alpha^{1/2} \left[\beta(\alpha^{-1}) + \alpha^{1/2} \beta(1 - \alpha^{-1/2})\right].$$

(3)

Here $\epsilon_0$ represents the thickness of the wall and $r_0$ the internal radius of the tube when the pressure inside equals that outside $(p = p_0)$. The thickness and the length of the tube are assumed to change by the factor $\alpha^{1/2}$. The pressures in table 5 were calculated from equation 3.

If there is no dissipation and if the amplitude is small, a mass suspended from a strip of rubber-like substance may oscillate with simple harmonic motion at a frequency given by the equation

$$\frac{w^2 L_o}{g} = \frac{B\beta^3 + 1/2 B\beta + 3/2 \nu \alpha^{1/2}}{(\nu \alpha^{1/2} \beta/2)}$$

(4)

where

$$\frac{I}{B} = \frac{1}{u^2} - \frac{1}{\sinh^2 u}$$  and $$\frac{D}{D} = \frac{1}{v^2} - \frac{1}{\sinh^2 v}$$

and $u$ and $v$ are the inverse Langevin function of $\beta(\alpha^{-1/2})$ and $\beta$ $(\alpha^{-1})$. $g$ is the gravitational acceleration, $L_o$ is the unstretched length of the specimen and $\alpha$ is the relative length $L/L_0$. If internal dissipation of energy occurs, measured values of $w^2 L_o/g$ are greater than those predicted by equation 4 by a factor $1/y$. The factor $y$ then may be considered a measure of the viscous damping in the specimen.

Detailed calculations were carried out on equation 4 on an IBM card calculator. Values of $w^2 L_o/g$ were computed for sufficient values of $\alpha$, so that the value of $\alpha_{\text{min}}$ could be determined to within 0.005. Values of $\beta$ between 0.20 and 2.00 were taken in steps of 0.05. A plot of $\beta$ vs. $\alpha_{\text{min}}$ was used to determine the values of $\beta$ to the nearest 0.01.

$\beta$-values determined in this way made use of only 1 experimental point. Entire $w^2 L_o/g$ versus $\alpha$ curves were plotted in families on log-log coordinates. Experimental data was plotted on a cellulose acetate sheet which overlaid the plots. The cellulose acetate sheet was then positioned by trial and error over the curve which fitted best. This method is laborious but has the advantage of using all the experimental points and allowing an evaluation of $\alpha_{\text{min}}$ in terms of all values of $\alpha$ used in the particular curve. In addition, $L_o$, can be treated as a variable. Thus, when the best fit is found, the deviation of the cellulose acetate origin of coordinates can be compared to that of the family of curves in the plot.

Acknowledgment

The author wishes to acknowledge the technical assistance of Mr. Herman Sharma and the advice and counsel of Professor Allen King.

Summario in Interlingua

Le application del teoria elastomico al studio del proprietates viscoclastic del aorta del can demonstra le mesme valor pro le parametro elastomico $\beta$ pote esser usate pro describer le comportamento elastic de pecias aortic sub conditiones tanto static como etiam dynamico. Le variatione in le parametro $\beta$ e le fraction amortiente $\gamma$ che occurre secundo le loco de origine del pecia de aorta pote esser distinguite solmente pro le extreme axe arterial central in consequentia del estense variabilitate in le proprietates mechanique quo existe in differente canes. Il pare che le proprietatas viscoclastic non es correlationate con le mesurate proprietates geometric in non-stressate pecias de aorta.

References

2. —: Changes in the elasticity of the aorta with age. Arch. Path. 29: 303, 1940.
VISCOELASTIC CONSTANTS FOR AORTA


Cardiopulmonary Effects of Pulmonary Venous Hypertension with Special Reference to Pulmonary Lymphatic Flow—pp. 324-335

Summario in Interlingua

Acute elevaciones del tension sinistro-atrial poteva esser regulata precisamente a non importa qual desirare niveau usque a, al media, 60 mm de Hg in cases a thorace intacte. Le sequente observationes eseva facite in un serie do 15 canes. Le fluxo del ducto dextero-lymphatic non esseva studiate a non alte tensiones proque non succesceva a mantenere tensiones sinistro-atrial medie a niveaux do plus que 25 mm de Hg. Le quantitate total de lympha remaineva augmentate durante periodos de usque a un hora pote que le tension sinistro-atrial habeva essitc restaurate a niveaux normal. Le elevation del tension sinistro-atrial non afficeva le fluxo lymphatic in le ducto thoracique. Edema pulmonar non occasoccava prestoce a tensiones sinistro-atrial medie solmente pawn eleventa supra le tension oncotic de plasma. Le production de edema pulmonar esseva observate solmente post elevations considerable del tension sinistro-atrial supra le tension oncotic de plasma durante periodos de 30 minutas o plus in alteremente normal cases. Le hematoerite montava significativemente post un periodo de deche minutas de alte tension sinistro-atrial. Le resistencia pulmonovascular non esseva reducida acutely quando le tension sinistro-atrial medie esseva eleventa ab 0 a 15 mm de Hg. Le resistencia montava gradualmente quando le tension sinistro-atrial esseva augmentate inter 15 e 30 mm de Hg. Le tension pulmono-arterial montava uniformemente e le rendimento cardiaco declinava uniformemente con lo augmento del tension sinistro-atrial.

Elevation chronic del tension sinistro-atrial esseva effectuate in 15 cases. Le tension sinistro-atrial medie variava inter 10 e 23 mm de Hg. Late cases esseva tenite sub observation durante periodos de usque a 10 menses, e le sequente observationes esseva facite. Le fluxo del ducto dextero-lymphatic non esseva trovate morte in los cagias 1 a 2 dies post le application del ligature. Non multes del usual alteraciones functional e structural que es frequemente incontrate in le pulmones de patientes con stenosis mitral (con le exception de hemosiderosis) esseva observate in nostre animals experimental, ben que le grado del stenosis supravalvular producite esseva trovate morte in los cagias 1 a 2 dies post le application del ligature. Non multes del usual alteraciones functional e structural que es frequemente incontrate in le pulmones de patientes con stenosis mitral (con le exception de hemosiderosis) esseva observate in nostre animals experimental, ben que le grado del stenosis supravalvular producite esseva trovate morte in los cagias 1 a 2 dies post le application del ligature.

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