Viscoelastic Properties of the Diastolic Left Ventricle in the Conscious Dog

J. Scott Rankin, Carl E. Arentzen, Philip A. McHale, David Ling, and Robert W. Anderson

SUMMARY The mechanical properties of the normal left ventricular wall during diastole were studied in 15 chronically instrumented, conscious dogs. Left ventricular minor and major axis diameters and equatorial wall thickness were measured with implanted pulse-transit ultrasonic dimension transducers. Left ventricular and pleural pressures were measured with high-fidelity micromanometers. Circumferential mural stress was calculated by using an ellipsoidal shell theory; circumferential strain was calculated by using a natural strain definition. The static elastic properties of the myocardium were estimated by fitting the stress-strain values at the points of diastasis during a venacaval occlusion to an exponential function. A modified creep test was used to evaluate the series viscous properties of the myocardium. Acute increases in systolic and diastolic loading were produced by inflating implanted aortic occluders for 15 minutes in five dogs. In these dogs, the static stress-strain curves were not altered significantly after this period of pressure loading, indicating that short-term series viscous properties are negligible. Parallel viscous properties were evaluated in 10 dogs by means of the variable rate stretch test of dynamic diastolic filling. A viscoelastic model incorporating a parallel viscous element fit the dynamic stress-strain data better and predicted the static elastic properties more accurately than a simple exponential model. Thus, the mechanical characteristics of the diastolic left ventricle can be represented most precisely by a viscoelastic model that includes a parallel viscous element.

AN EXPONENTIAL relationship between ventricular pressure and volume during diastole was first demonstrated by Frank.1 Influenced by Blix,2 Frank concluded that this observation represented a fundamental relationship between diastolic force and length within the ventricular wall. During the past two decades there has been a renewed interest in describing the mechanical properties of diastolic myocardium, and numerous investigators have confirmed Frank's observations in a variety of experimental preparations.3-6 On the basis of these studies it generally has been accepted that, in both the isolated and the intact heart, an exponential relationship exists between left ventricular pressure and volume during diastole.6 However, several authors recently have suggested that fitting pressure-volume or pressure-dimension data from a single diastolic filling period with a simple exponential function may be an oversimplification.7-12 Indeed, previous experiments that demonstrated exponential pressure-dimension curves in the intact heart utilized static measurements during induced volume changes.13 So far as we are aware, there are no directly measured data in the literature which indicate that ventricular pressure and dimensions are exponentially related during the dynamic filling of a single diastole.

The diastolic mechanical properties of isolated myocardium cannot be described by a simple exponentially elastic model.14,15 Significant viscous dissipation and stress relaxation usually have been observed, requiring more complicated viscoelastic models to approximate the measured data. In fact, nonlinear viscoelasticity is the mechanical hallmark of most biological materials.14 The present study was designed to evaluate the relationship between normalized mural force (stress) and normalized length (strain) in the left ventricles of conscious dogs. Particular attention was given to developing a model, based on classical viscoelastic theory,17 that would represent the observed mechanical properties of the intact diastolic myocardium.

METHODS

EXPERIMENTAL PREPARATION

The preparation used in these studies has been described in detail elsewhere.18 Briefly, 15 adult mongrel dogs were surgically instrumented with pulse-transit ultrasonic dimension transducers to directly measure the dynamic geometry of the left ventricle. External minor axis diameter was obtained by transducers sutured to the anterior and posterior left ventricular walls. Major axis diameter was measured by transducers positioned at the base of the left ventricle and on the apical dimple. A third pair of ultrasonic crystals measured equatorial wall thickness; lateral wall thickness was obtained in four of the dogs, and anterior wall thickness was measured in the remainder. Silicone rubber balloon occluders were implanted around both venae cavae, and a silicone rubber catheter was placed near the epicardial surface of the heart to permit passage of a micromanometer into the pleural cavity. The
pericardium was routinely left open. In five of the dogs a balloon occluder was positioned around the ascending aorta. The dimension transducer connectors, the occluder tubing, and the end of the pleural catheter were implanted in a subcutaneous pouch at the dorsal aspect of the incision, and the thoracotomy was repaired. All dogs received intramuscular injections of procaine penicillin G (6 × 10⁶ U) and dihydrostreptomycin (0.75 g) for 5 days postoperatively.

**INSTRUMENTATION AND DATA ACQUISITION**

Pulse-transit sonomicrometry measures the time delay from transmission of a burst of ultrasound from one piezoelectric transducer to the reception of the sonic wave by an identical transducer. Since the velocity of sound in body tissues and blood is approximately constant, the measured temporal delay is directly proportional to the distance between the transducers. Calibrations are obtained by substituting an electronically generated time delay into the circuitry. The frequency of response of the system is flat to beyond 100 Hz, the minimum resolution is approximately 0.05 mm, and the maximum electronic drift rate is 0.05 mm/hour.

Pressures were measured with Millar PC-350 piezoresistive micromanometers. The manometers were driven with Hewlett-Packard 8805C carrier preamplifiers and were balanced and zeroed at 37°C to atmospheric pressure. The zero drift of each pressure channel was less than 0.25 mm Hg in every study. We found that it was important to warm up the manometers for 12 hours prior to use to obtain the best drift characteristics. The micromanometers were periodically dynamically tested by using a sinusoidally oscillating pressure wave of varying frequencies, and the frequency response of each transducer was consistently flat to beyond 200 Hz. The resonant frequency of the Millar manometer is 25–35 kHz.20

After recovery from implantation (7–10 days postoperatively), each dog was studied in the conscious state. One intravenous injection of morphine sulfate (0.5 mg/kg) was used when necessary for analgesia during catheterization. The right inguinal region and the subcutaneous pouch were anesthetized locally with 1% lidocaine. The pouch intravenous injection of morphine sulfate (0.5 mg/kg) was used when necessary for analgesia during catheterization. The pouch was opened, and the dimension transducer connectors were directly coupled to the sonomicrometer. A micromanometer was introduced percutaneously into the right femoral artery, advanced into the left ventricle under fluoroscopic control, and positioned exactly between the minor axis dimension transducers. Another micromanometer was passed through the implanted pleural catheter to the epicardial surface of the heart and positioned at the same vertical level as the intracavitary manometer. An airtight connector on the external end of the pleural tube prevented pneumothorax.

Analog measurements of minor and major axis diameters, wall thickness, left ventricular intracavitary pressure, and intrapleural pressure were recorded on magnetic tape with a Hewlett-Packard model 3520B FM recorder. Data were recorded in each study during a resting, control period and then during three vena caval occlusions. Each vena caval occlusion was held until a stable minimum

minor axis dimension was observed, and then the occluders were released. The maximum duration of a vena caval occlusion was 30 seconds. After return to control conditions, an aortic occluder was inflated in five dogs to produce a stable peak systolic left ventricular pressure of approximately 200 mm Hg. Fifteen minutes after inflation, repeat vena caval occlusions were performed. It was helpful to monitor the diastolic pressure-dimension relationship during the course of each experiment by plotting high gain transmural pressure (obtained by analog subtraction of intrapleural pressure from intracavitory pressure) vs. minor axis diameter on a Tektronix R513N storage oscilloscope.

**DATA ANALYSIS**

Recorded analog data were digitized at 5-msec intervals with an IBM 1130/System 7 digital computer. Two 10-second segments of control data and one vena caval occlusion from each study were analyzed. The left ventricle was represented as a three-dimensional ellipsoidal shell. The measured major and minor axis diameters were used as the external dimensions of the shell. The measured wall thickness was used as the dynamic equatorial shell thickness; the shell thickness at the base and apex was assumed to be 55% of the equatorial value. The validation of this geometric model has been presented previously.18

Left ventricular intracavitary volume (V) was calculated from the formula for a prolate spheroid:

\[ V = \frac{2}{3} (b - 2h)^2 (a - 1.1h), \]

(1)

where b is the external minor axis diameter, h is the wall thickness, and a is the external major axis diameter. The instantaneous midwall minor axis circumference (l) was calculated from the equation:

\[ l = \pi (b - h). \]

(2)

The midwall circumferential length was normalized using a natural strain definition:

\[ \epsilon = \ln \frac{l}{l_0}, \]

(3)

where \( \epsilon \) is the strain, and \( l_0 \) is the minimum diastolic midwall circumference measured during a maximum vena caval occlusion. A natural strain normalization was selected because it does not assume a uniform deformation or stress distribution along the ventricular wall.21 Circumferential strain rate (\( \dot{\epsilon} \)) or the normalized velocity of diastolic circumferential lengthening was computed as the first time derivative of the strain, using an orthogonal polynomial approximation.

The transmural pressure of the left ventricle was computed from the digitized data as the difference between the intracavitary pressure and the intrapleural pressure. The mean tensile stress in the minor axis circumference was calculated by using a modification of the thin-walled, ellipsoidal shell theory formulated by Sandler and Dodge:22

\[ \sigma = \frac{P}{h} \left[ 1 - \frac{r_s^3}{r_p^3 (2r_p + h)} \right] \]

(4)
where $\sigma$ is the mean circumferential tensile stress in dynes/cm$^2$, $P$ is the transmural pressure of the left ventricle in dynes/cm$^2$, $r_b$ is the minor axis midwall radius, $r_a$ is the major axis midwall radius, and $h$ is the equatorial wall thickness. This equation has been empirically tested and has been shown to estimate satisfactorily the directly measured wall stress during a variety of interventions.\(^{23}\)

In each study, the static elastic characteristics of the myocardium were estimated from vena caval occlusion data. Using a nonlinear least squares regression analysis,\(^{24}\) the stress-strain values at the points of diastasis ($e = 0 \pm 0.02$ sec$^{-1}$) from consecutive diastoles during the course of a vena caval occlusion were fitted to the equation:

$$\sigma = \alpha (e^{\beta e} - 1),$$

where $\alpha$ and $\beta$ are the nonlinear elastic constants. The vena caval occlusion method of determining the static elastic properties of the diastolic myocardium was assumed to be the standard of reference. The dog-to-dog variability of the diastolic elastic properties was determined by covariance matrix analysis of the logarithmically transformed nonlinear elastic constants.

Three models of the mechanical properties of the diastolic myocardium were evaluated:\(^{17}\) model I, a simple nonlinear elastic model; model II, a viscoelastic model incorporating a series viscous element (Maxwell model); and model III, a viscoelastic model incorporating a parallel viscous element (Kelvin or Voigt model). To test model II, $I_0$ and the static elastic constants were determined from vena caval occlusion data before and 15 minutes after inflation of the implanted aortic occluders in five dogs. The abrupt elevation in systolic and diastolic stresses produced by inflating the aortic occluder was assumed to be approximately analogous to a creep test. Using a $t$-test of the logarithmically transformed nonlinear stress-strain data, the control static elastic curve in each dog was compared statistically to that obtained after 15 minutes of inflation. Similarly, the values for $I_0$ and the logarithmically transformed nonlinear static elastic constants obtained during control conditions in the five dogs were compared to those determined after the period of pressure loading using a paired $t$-test.

Model III was evaluated in 10 dogs, using the variable rate stretch test of dynamic diastolic filling. The significance of parallel viscous properties was determined by comparing the relative ability of model I and model III to predict the measured dynamic stress-strain data. Three control diastoles, entirely within expiratory pauses, were randomly selected from each study for analysis. The period of diastole was defined as beginning at the minimum transmural pressure after ventricular systole and ending at the minimum transmural pressure after atrial systole. Using a nonlinear least squares regression analysis, the dynamic stress-strain data from the selected diastoles were fitted to the constitutive equation of model I (Eq. 5). The sum of squares deviation of the data points predicted by this model from the measured data points was computed. The fit of the measured data was evaluated statistically by using an $F$ test of the variance of the predicted data around the measured data.

The measured dynamic diastolic stress-strain data also were fitted to the constitutive equation of model III:

$$\sigma = \alpha (e^{\beta e} - 1) + \eta \dot{e},$$

where $\alpha$ and $\beta$ are the nonlinear elastic constants and $\eta$ is the linear viscous constant. The sum of squares deviation of the data points predicted by this model was similarly calculated, and the fit was evaluated with an $F$ test. The elastic constants predicted from the dynamic data with model III and model I in each of the 10 dogs were logarithmically transformed and compared to the static elastic constants determined from the vena caval occlusion data, using a paired $t$-test.

**Results**

Representative analog data obtained during a control period are illustrated in Figure 1. The dynamic geometric pattern observed in the conscious dog and the validation of this model have been presented elsewhere.\(^{18}\) The wall thickness measurements obtained from the lateral free wall in the present study were similar to those previously described from the anterior wall,\(^{18}\) except that at normal end-diastolic volumes the lateral wall more frequently thinned slightly during diastolic contraction (Fig. 1). Total systolic wall thickening measured at the equator in the control state ranged from 26% to 41% of the end-diastolic value.

The transmural pressure of the left ventricle obtained with the technique described in this paper was relatively stable throughout the respiratory cycle. In Figure 1 it can be seen that although left ventricular intracavitary pressure and pleural pressure decreased over 10 mm Hg during inspiration, the transmural pressure increased slightly consistent with the dimension change. In most dogs the diastolic transmural pressure-minor axis diameter relationship was constant throughout the respiratory cycle. Occasionally, however, measuring pleural pressure in this manner did not totally cancel respiratory variations in the diastolic pressure-dimension curve, necessitating analysis of expiratory diastoles. As shown in Figure 1, slight systolic perturbations of the pleural pressure were observed frequently when measurements were obtained on the cardiac surface. The average end-diastolic transmural pressure was 11.5 mm Hg $\pm$ 0.8 mm Hg, the average end-diastolic circumferential stress was $33.2 \times 10^3$ dynes/cm$^2$ $\pm$ $1.9 \times 10^3$ dynes/cm$^2$, and the average end-diastolic volume was $70.7 \pm 4.2$ ml (Table 1).

The changes that typically were observed in left ventricular dimensions and pressures during a vena caval occlusion are shown in Figure 2. Both ventricular diameters progressively decreased, and the wall thickened until a stable minimum volume was reached. Left ventricular pressures progressively fell until, at maximum vena caval occlusion, the diastolic transmural pressure was approximately 0 mm Hg. The $I_0$ measured at maximum vena caval occlusion ranged from 10.2 cm to 14.5 cm (mean, 12.3 $\pm$ 0.4 cm). The average diastolic transmural pressure at the point where $I_0$ was measured ($P_a$) was 0.7 mm Hg $\pm$ 0.3 mm Hg (range, $-0.39$ to $+2.39$ mm Hg), and the average intracavitary volume at $I_0$ ($V_a$) was $23.1 \pm 1.8$ ml (Table 1).
FIGURE 1  Representative analog recordings of the measured physiological data in the conscious dog. The transmural pressure of the left ventricle was obtained by analog subtraction of the intrapleural pressure from the intracavitary pressure.

Calculated stress-strain data from several diastoles during a vena caval occlusion are illustrated in Figure 3. The static elastic curve determined by fitting the diastatic points to Equation 5 is represented by the smooth exponential. The individual and mean static elastic constants determined by this method are given in Table 1.

During the creep tests, inflation of the aortic occluder significantly increased peak systolic pressure, end-diastolic pressure, and end-diastolic stress (P < 0.003). Figure 4 is a comparison of the diastatic stress-strain data obtained during a vena caval occlusion in the control state and after 15 minutes of pressure loading. The static diastolic stress-strain curve was not significantly changed (P > 0.3) in any dog by the 15-minute period of elevated diastolic stress (Table 2). The $\alpha$, $\beta$, and $l_0$ values determined after the period of pressure loading were not significantly different from control (P > 0.2).

A representative dynamic stress-strain curve from a single control diastole is illustrated in Figure 5A. Figure 5B is a graph of diastolic strain rate vs. strain on the same scale. The interval between each pair of data points represents 5 msec. The latter half of the rapid-filling phase, two damped oscillations, the slow-filling phase, and the atrial systolic-filling phase can be identified. Half of the rapid filling occurred before the transmural pressure reached the first diastolic minimum. The dynamic stress-strain relationship did not appear to be a simple exponential function in any study. During the phases of diastolic filling when the velocity of lengthening of the muscle was high, the stress appeared to be greater than would be predicted by a simple exponential model. Indeed, the fit of the measured data with model I was statistically significant only in two of the dogs (Table 1).

A typical fit of the measured dynamic diastolic stress-strain data with model III is illustrated in Figure 6. The data points predicted by model III (open circles) approximated the measured data (filled circles) with a confidence level of greater than 95% in 9 of the 10 dogs (Table 1). The sum of squares deviation from the measured dynamic data points was better in every study with model III than with model I. The beat-to-beat variation in the calculated constants was also less with model III in most dogs. Figure 7 is a comparison of the static elastic curve calculated from vena caval occlusion data and the elastic curves calculated from a single dynamic diastole by model I and model III in the same dog. The elastic constants calculated from dynamic data with model III were significantly different from those determined from the vena caval occlusion data (P < 0.001). There was no significant difference between the elastic constants calculated from dynamic data with model III and those obtained from static vena caval occlusion data (P > 0.6). In every study, the static elastic properties determined during the vena caval occlusion were predicted more accurately from dynamic data by model III. Figure 7 also illustrates that, in the recumbent conscious dog, normal diastolic filling occurs on the rela-
TABLE 1  Hemodynamic and Diastolic Mechanics Data Obtained in Conscious Dogs under Control Conditions

<table>
<thead>
<tr>
<th>Dog no.</th>
<th>Wt (kg)</th>
<th>HR (min⁻¹)</th>
<th>EDP (mm Hg)</th>
<th>EDV (ml)</th>
<th>$\epsilon_a$ (cm)</th>
<th>$V_b$ (ml)</th>
<th>$a^*$</th>
<th>$\beta$</th>
<th>$\Sigma Sq/n$</th>
<th>$a^*$</th>
<th>$\beta$</th>
<th>$\eta$</th>
<th>$\Sigma Sq/n$</th>
<th>$\Sigma Sq$ model I</th>
<th>$\Sigma Sq$ model III</th>
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Wt = body weight; HR = heart rate; EDP = end-diastolic pressure; and EDV = end-diastolic volume.

The static elastic constants derived from vena caval occlusion data are considered the standard of reference. For comparison, elastic constants calculated from dynamic diastoles with model I and model III are presented for each dog. The dynamic constants are represented as the mean ± average deviation from the mean of three control diastoles to illustrate the beat-to-beat variability. $\Sigma Sq/n = \sum$ the sum of the squares deviation of the predicted data points from the measured data points divided by the number of data points. $\Sigma Sq/n$ of less than 2.0 represents a statistically significant fit with a confidence level of greater than 95%. The ratio $\Sigma Sq/n$ model I/$\Sigma Sq/n$ model III quantifies the relative fits of the two models.

* Expressed as dynes x $10^9$/cm².
† Expressed as dyne sec x $10^9$/cm².

**FIGURE 2**  Effects of a vena caval occlusion on left ventricular dimensions and pressures in the conscious dog.
tively stiff part of the elastic curve. It is also evident that the left ventricle of the resting dog fills and shortens over a mean circumferential deformation of 15-35% extension from the unstressed dimension (a)

A composite graph of the static elastic curves determined for each of the 10 dogs is illustrated in Figure 8A. The data in Figure 8B illustrate the variability of the viscoelastic properties of the normal left ventricular wall as calculated with this method.

Discussion

During the initial phases of the present study, several factors were found to be of importance in obtaining meaningful diastolic mechanics data in conscious dogs. Because negative intrapleural pressure accounts for a significant portion of the diastolic transmural loading in closed-chest preparations, it was necessary to measure this variable. After attempting several techniques previously used for obtaining intrapleural pressure, 25-26 we developed the method described in this paper; this technique proved to be very satisfactory when properly performed. It was important to position the intrapleural manometer near the epicardial surface of the heart at the same vertical level as the intracavitary manometer, because significant gravitational pressure gradients exist throughout the pleural cavity. 26, 27 Unfortunately, the reliability of the intrapleural pressure measurement was limited to 10-14 days after implantation because of the formation of pleural adhesions after this period. In addition, there was occasionally minor respiratory variation in the diastolic pressure-dimension curve, requiring the use of expiratory diastoles for quantitative analysis. For the most part, however, this technique worked very well.

One potentially important variable that was not measured in this study is right ventricular pressure. Several investigators have shown that right ventricular pressure may represent the external loading of the septum and, thus, may influence left ventricular pressure measurements to some extent. 28, 29 If this hypothesis is valid for the intact heart, diastolic right ventricular pressure would have to be measured and taken into consideration in some cases.

Table 2 Hemodynamic and Diastolic Mechanics Data in Conscious Dogs before and after Pressure Loading

<table>
<thead>
<tr>
<th>Dog no.</th>
<th>PSP (mm Hg)</th>
<th>EDP (mm Hg)</th>
<th>EDσ* (cm)</th>
<th>α*</th>
<th>β</th>
<th>PSP (mm Hg)</th>
<th>EDP (mm Hg)</th>
<th>EDσ* (cm)</th>
<th>α*</th>
<th>β</th>
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<td>219.4</td>
<td>17.9</td>
<td>63.3</td>
<td>12.2</td>
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<td>13.68 &gt;0.7</td>
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<td>Mean</td>
<td>134.5</td>
<td>9.6</td>
<td>28.8</td>
<td>13.1</td>
<td>13.1</td>
<td>1.330</td>
<td>11.82</td>
<td>198.3</td>
<td>14.9</td>
<td>48.6</td>
<td>13.1</td>
<td>1.191</td>
<td>12.77</td>
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<tr>
<td>SEM</td>
<td>4.4</td>
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<td>2.8</td>
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<td>0.4</td>
<td>6.7</td>
<td>0.9</td>
<td>4.1</td>
<td>0.4</td>
<td>1.330</td>
<td>11.82</td>
<td>&gt;0.5</td>
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</table>

Abbreviations are the same as in Table 1. PSP = peak-systolic pressure and EDσ = end-diastolic stress. The elastic constants were obtained from vena caval occlusion data. P<sub>α</sub> represents the P value of a t-test which evaluated the difference between the control and 15-minute static stress-strain curves in each dog. P<sub>β</sub> represents the P value of a paired t-test which evaluated the difference between control and 15-minute data from all of the dogs.

* Expressed as dynes × 10<sup>5</sup>/cm<sup>2</sup>.
complex way. However, if this phenomenon had been significant in the present study, the measured left ventricular pressure should have decreased inappropriately immediately after occlusion of the venae cavae, because right ventricular pressure should have fallen several beats before left ventricular volume. An initial downward shift of the measured diastolic transmural pressure at the onset of a vena caval occlusion was never observed. This finding suggests that right ventricular pressure was not an important factor. However, the definitive answer to this problem awaits further systematic investigation in intact preparations.

The accuracy of left ventricular dimension measurements is essential in the calculation of diastolic mechanics. The geometric measurements obtained with the ultrasonic method, and the ellipsoidal model used in the present study, are particularly critical. The data were fitted to the constitutive equation of model III.

The measured dynamic diastolic stress-strain data (filled circles) were fitted to the constitutive equation of model III. The data predicted by this model are illustrated by the open circles. The lower exponential is the static elastic curve predicted from dynamic data by model III.

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The measured dynamic diastolic stress-strain data (filled circles) were fitted to the constitutive equation of model III. The data predicted by this model are illustrated by the open circles. The lower exponential is the static elastic curve predicted from dynamic data by model III.
study have been extensively validated in a previous communication. Since the major axis diameter is 30-40% greater than the minor axis diameter, an ellipsoidal model is required to represent the geometry of the left ventricle. Moreover, spherical shell theories have been shown to significantly underestimate the circumferential tensile stress further emphasizing the importance of ellipsoidal modeling when precise mechanical data are desired.

The use of shell theory to calculate wall stress is based on the assumption that wall inertia is negligible and that a balance of forces continually exists across the ventricular wall. A modification of Newton's second law, based upon the minor axis radial acceleration and the mass of the circumferential hoop, predicts diastolic inertial forces that are very small in comparison to the viscoelastic forces. This is true of most viscoelastic materials. Inertial considerations, therefore, were not included in the present analysis.

A number of shell theories have been derived for the calculation of tensile stress in the intact heart. Equations taking into account anisotropy, inhomogeneity, fiber orientation, and large deformation theory have been presented. Although each of these concepts has theoretical propriety, it is at present impossible to test the various hypotheses on which each of these theories is based. We choose, therefore, to confine our calculations to the mean tensile stress in the equatorial direction, using a stress theory that has been empirically validated. Of course, the force resultants within a shell form a three-dimensional vector system, and each of the three dimensions within the ventricular wall probably has characteristic mean stress-strain properties. Furthermore, it is likely that myocardium is an inhomogeneous material and that the three-dimensional mechanical characteristics at various points within the wall are different. However, it must be emphasized that regional wall force and regional deformation, when integrated across the wall, would reduce to the empirically validated mean tensile stress and strain used in the present study. The end-diastolic circumferential stress calculated during control conditions in dogs averaged 33.2 x 10^3 dynes/cm^2 ± 1.9 x 10^3 dynes/cm^2 at an average end-diastolic pressure of 11.5 mm Hg ± 0.8 mm Hg. These data correlate well with the average end-diastolic wall stress of 38.8 x 10^3 dynes/cm^2 ± 10.7 x 10^3 dynes/cm^2 directly measured by Burns et al. at an average end-diastolic pressure of 12.3 mm Hg.

The pioneering experiments of Noble et al. suggested that parallel viscous properties were important determinants of left ventricular diastolic mechanics and that series viscous properties were negligible. Several investigators subsequently demonstrated that, during rapid diastolic filling, the pressure within the left ventricle was higher than would have been predicted by a simple exponential model, implying the significance of parallel viscosity. The data obtained in the present study support these hypotheses. In the creep experiments, the pressure loading did not alter the static elastic characteristics over the time period tested, demonstrating the insignificance of short-term series viscous properties. Certainly, at some diastolic pressure, the yield stress of the myocardium would be exceeded, and plastic deformation of the wall would occur. It is also likely that longer periods of pressure loading would induce a creep phenomenon.

On the other hand, parallel viscous properties were significant during dynamic diastolic filling. The dynamic stress-strain data were approximately better by model III, which included a parallel viscous element; the static elastic properties of the myocardium were predicted more satisfactorily from dynamic data when this model was used. The increment in mural stress above the static elastic curve during filling could be directly related to the rate of circumferential lengthening. Thus, the rate of mural deformation seemed to contribute importantly to diastolic mural force. On the basis of these data it is concluded that the Kelvin model most accurately represents the mechanical properties of the left ventricle during diastole and should be used in the analysis of dynamic data.

One alternative explanation for a higher measured transmural pressure during rapid filling is the end-pressure artifact where the kinetic energy of blood impacting a manometer is converted into pressure. On the basis of linear filling velocities calculated from data obtained in the present study, end-pressure artifact would account for less than 10% of the pressure increment observed at peak rapid filling. This phenomenon, therefore, was not taken into consideration.

The diastolic elastic constants determined for the intact left ventricle in the present study are quite similar to those published previously for isolated myocardium. However, the data obtained in the intact heart are in some ways different from previous observations in isolated preparations. For example, Pinto and Fung found that viscous dissipation (a parallel viscous property which they defined as the area within a hysteresis loop) was nearly independent of the rate of deformation. In the present study, we found that the parallel viscous forces were linearly related to the rate of deformation. There are several possible explanations for this difference. First, the strain rates that occur in the intact heart (Fig. 5) are many times greater than those used by Fung in his studies. Perhaps at higher rates of deformation, the internal frictional forces are related more directly to the strain rate. Furthermore, as Fung suggested, the "preconditioning" procedure, devascularization of the isolated tissue, or some other factor related to the in vitro state might alter the normal homeostatic mechanical characteristics of the myocardium and account for the more complex mechanical properties that are observed in isolated tissue.

It is evident in Figure 5 that one or two damped oscillations of the left ventricular wall occurred at the end of the rapid-filling phase of diastole. This finding was consistently noted in these experiments. Such damping would not occur in the absence of parallel viscous properties and offers further evidence of the viscoelastic nature of myocardium. The frequency of the oscillations was uniformity 10-15 Hz in the normal ventricle. More than likely, the third and fourth heart sounds encountered in certain pathological states are produced by this type of oscillation.
change of the vibratory frequency to the audible range in such conditions may reflect alterations in the viscoelastic properties of the myocardium, and oscillatory motion analysis potentially could be useful in determining diastolic mechanical properties.

It has been suggested by several authors that diastolic suction plays a significant role in the early phase of filling of the left ventricle. In the present study, such a diastolic recoil phenomenon could not be demonstrated. Even during maximal vena caval occlusion, the minimum diastolic transmural pressures that were measured were approximately zero (Fig. 2), and significantly negative pressures were never observed. Thus, the left ventricle of the conscious dog does not appear to shorten beyond $l_0$, and elastic restorative forces do not contribute to diastolic filling under normal conditions. Previous data which suggested that such a phenomenon could exist were probably a function of the artificial experimental preparations employed. The values for $V_o$ determined in the present study compare favorably to those directly measured by Brecher et al.

It is likely that the diastolic mechanical properties of the left ventricle are changed during global cardiac dilation such as occurs in valvular heart disease. Alterations in the diastolic properties may play a significant role in the deterioration of myocardial function that accompanies cardiac dilation in these diseases. Thus, investigation of the viscoelastic properties of diastolic myocardium in man may provide insight into the pathophysiology of left ventricular dysfunction.

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References

2. Blix M: Die lange and die spannung des muskels. Skand Arch Physiol 4: 399-409, 1893
34. Gould P, Ghista D, Bromberg L, Minsky I: In vivo stresses in the human left ventricular wall; analysis according for irregular three-dimensional geometry and comparison with idealised geometry analyses. J Biomech 5: 521-539, 1972
Viscoelastic properties of the diastolic left ventricle in the conscious dog.
J S Rankin, C E Arentzen, P A McHale, D Ling and R W Anderson

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