Technique for Measuring Regional Two-Dimensional Finite Strains in Canine Left Ventricle

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We developed a technique to measure regional two-dimensional deformations in the myocardium. Three piezoelectric crystals were implanted in a triangular array in the left ventricular anterior midwall in six anesthetized dogs. Each crystal was used in a dual function, to both transmit and receive ultrasonic signals from the other two crystals. In this manner, the three segment lengths of the crystal triangle throughout the cardiac cycle were simultaneously recorded. The orientation of the crystal triangle with reference to the left ventricular long and minor axes was determined. The orientation and three segment lengths of the crystal triangle were used to calculate the circumferential strain $E_{c}$, the longitudinal strain $E_{l}$, the in-plane shear strain $E_{s}$, and the mutually perpendicular principal strains $E_1$ and $E_2$. Also, the orientation of the first principal direction or the in-plane angle was determined, which was defined as the angle between the first principal direction ($E_1$) and the circumferential direction ($0^\circ$). This information fully describes the regional two-dimensional myocardial deformations. This technique was applied to measure regional myocardial deformations at three different left ventricular end-diastolic pressures (LVEDP) of 2 ± 1 (mean ± SD), 8 ± 1, and 17 ± 2 mm Hg. The first principal direction at end-systole was oriented away from the circumferential direction at low LVEDP ($-43 ± 21^\circ$) but became progressively closer in each animal to the circumferential direction as LVEDP increased to mid ($-26 ± 18^\circ$) and high ($-14 ± 13^\circ$) levels. The end-systolic ratio $E_{c}/E_{l}$ was 0.6 ± 0.2 at low LVEDP, but increased toward unity in each animal to 0.9 ± 0.1 at mid and high LVEDP. Thus, at low LVEDP, the greatest systolic deformation occurred in a direction different from the circumferential orientation. Therefore, circumferential strain measurements ($E_c$) significantly underestimated the greatest systolic deformation ($E_1$). However, as LVEDP increased, the first principal direction rotated closer toward the circumferential orientation, and circumferential strain measurements adequately estimated the greatest systolic deformation. Nevertheless, the presence of significant amounts of shortening along either the longitudinal ($E_{l}$) or the second principal direction ($E_2$) in the midwall necessitated the use of the two-dimensional method. The change in end-diastolic configuration as LVEDP increased from 1 ± 1 to 16 ± 1 mm Hg was also examined. Unlike the end-systolic data, the end-diastolic first principal direction did not deviate significantly from the circumferential direction at any LVEDP. Although the end-diastolic circumferential strain $E_c$ was a good approximation of the greatest end-diastolic deformation $E_1$ at all LVEDP, there were significant lengthening deformations in the longitudinal direction that were not accounted for by the uniaxial circumferential measurements. In summary, we have described a technique to measure regional two-dimensional myocardial deformations and have demonstrated that under common physiological conditions, the measurement of regional ventricular function in a single, fixed orientation is not adequate to fully describe complex regional myocardial deformations.

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a triangular array in the plane of the left ventricular anterior midwall. Each crystal was used in dual roles, to both transmit and receive ultrasonic signals from the other two crystals. In this manner, the three segment lengths of the "crystal triangle" were measured continuously and used to compute the three components of a finite strain tensor in the local midwall plane. By fully accounting for the deformation in the plane of measurement with respect to a local cardiac coordinate system (e.g., circumferential, longitudinal, and in-plane shear strains), the two-dimensional principal strains could be determined in a cardiac coordinate system. These principal strains and their associated principal directions were independent of the original reference coordinate system.

As a first application of this technique, two-dimensional finite deformations in the left ventricular anterior midwall were measured at different left ventricular volumes. These results demonstrate the importance of assessing two-dimensional finite deformations and the limitations of measuring a myocardial dimension change in a single, fixed direction.

**Materials and Methods**

**Experimental Preparations**

Six adult mongrel dogs were anesthetized with intravenous pentobarbital (30 mg/kg), intubated, and ventilated with a Harvard respirator. The heart was exposed through a median sternotomy and bilateral fifth intercostal space thoracotomy and suspended in a pericardial cradle. A Konigsberg P-20 micromanometer (Pasadena, California) was inserted into the left ventricle through an incision in the left ventricular apex and secured in place with a purse-string suture. A 100-cm 7F fluid-filled pigtail catheter was introduced into the left ventricle from a femoral artery, and a short, fluid-filled, polyvinyl tube was inserted into the left atrium through an incision in the left atrial appendage. Both fluid-filled catheters were attached to Statham P23DB transducers with zero reference at the level of the right atrium. Left ventricular pressure waveforms from the fluid-filled and micromanometer catheters in the left ventricle were matched. Then, the fluid-filled catheter was withdrawn into the ascending aorta just above the aortic valve to record central aortic pressure. The match between the fluid-filled catheter and micromanometer catheters was checked periodically throughout the experiment. A femoral vein was cannulated for administration of intravenous fluids. Inflatable occlusion cuffs were positioned around the superior and inferior venae cavae. Limb leads for an electrocardiogram were placed.

Regional ventricular deformations were measured with three piezoelectric crystals (2 mm diameter) placed in a triangular array (Figure 1). Each crystal was inserted into the midwall of the myocardium through an epicardial stab wound. The three crystals formed a triangle in the midwall that was roughly equilateral. The segments delineated by each pair of crystals were approximately 1.0–1.5 cm long. The area of the triangle was approximately 1 cm². It was assumed that strains were uniform within the triangle and that the crystals were not so close that they disturbed the deformation field. To ensure that all crystals were implanted to the same midwall depth, a small strip of tape was wrapped around the plastic introducer sleeve with the edge of the tape placed 0.5 cm from the center of the crystal. The tape provided a stop that prevented the crystal from being plunged in deeper than 0.5 cm from the epicardial surface. In larger animals, the tape was placed 0.55–0.60 cm from the center of the crystal.

The orientation of the piezoelectric crystal triangle was determined from an epicardial reference system (Figure 1). A length of suture material was used to visually define the epicardial long axis of the anterior

**Figure 1.** Schematic representation demonstrating the location and position of the crystal triangle in left ventricle. External long axis of anterior left ventricular wall is shown by dashed line from bifurcation of left main coronary artery to left ventricular apex (see text). Bottom inset, midwall location of each crystal and how each crystal was required to perform dual functions, to both transmit (T) and receive (R) ultrasonic signals. Top inset, method of wiring three crystals (crystals 1, 2, and 3) to three channels (channels A, B, and C) of sonomicrometer amplifier. Transmitter outputs of three channels are shown in top row (T), and receiver inputs are in bottom row (R). Piezoelectric crystal 1 was connected to transmitter output of channel A, and crystal 2 was connected to receiving input of channel A. Therefore, channel A of sonomicrometer amplifier measured the segment between crystals 1 and 2, or segment A of crystal triangle. Crystal 3 was connected to transmitter of channel B. Receiving input of channel B was connected to receiver of channel A, which received its input from crystal 2. Therefore, channel B measured the segment between crystals 2 and 3, or segment B of crystal triangle. Transmitter output of channel C was connected to transmitter of channel A, which was connected to crystal 1. Receiving input of channel C was connected to transmitter of channel B, which was connected to crystal 3. Therefore, channel C measured the segment between crystals 1 and 3, or segment C of crystal triangle. There was no interference in signals with this wiring because each channel of the sonomicrometer amplifier transmits and receives signals in a cycled sequence.
left ventricular wall. One end of the suture was attached to the epicardial surface near the bifurcation of the left main coronary artery. The suture then was placed along the anterior left ventricular epicardial surface, and the other end of the suture was attached to the apical dimple. Prior studies in this laboratory have demonstrated that this external (epicardial) anterior wall long axis accurately reflects the internal z axis that Streeter et al used to define myocardial fiber directions. The crystal triangle was implanted so that one side of the triangle was purposefully aligned with the external long axis of the left ventricle anterior wall (Figure 1).

Measurements of the three segment lengths of the crystal triangle required minor modifications of standard sonomicrometry techniques and equipment. Standard 2-mm piezoelectric crystals in which resin had been added very carefully were used to produce lenses with a spherical shape. This allowed the ultrasonic signal to be both transmitted and received over a wide angle. Each crystal had to be simultaneously focused on the other two crystals of the triangle; this was accomplished by implanting each crystal so that it faced roughly halfway between the other two crystals of the triangle. The crystals were wired to a standard sonomicrometer amplifier. However, wiring modifications (as shown in Figure 1) were required so that each crystal could be used for dual functions, that is, to both transmit and receive ultrasonic signals. Signal interference did not occur with the modified wiring scheme because each channel of the sonomicrometer amplifier transmits and receives signals in a cycled sequence (sampling rate per channel, 375 Hz). Thus, each of the three piezoelectric crystals transmitted and received ultrasonic signals at different times in the cycle.

The electrocardiographic lead, aortic pressure, high-and low-gain left ventricular pressures, left atrial pressure, and the three segment length signals from the crystal triangle were recorded on an eight-channel forced-ink polygraph (model 2000, Brush-Clevite, Cleveland, Ohio) at a paper speed of 200 mm/sec and on FM magnetic tape (Figure 2). All measurements were obtained with respiration suspended at end-expiration for approximately 10–15 cardiac cycles.

In three of the six animals, the accuracy of the crystal triangle technique was evaluated by an independent technique. Five small radiopaque beads (2 mm diameter) were sutured to the epicardial surface. One bead was placed near the left main coronary artery, a second bead was placed over the apical dimple, and three beads were placed over the epicardial entry sites of the three piezoelectric crystals. High-speed biplane cineradiograms (120 frames/sec, 16 mm) were obtained over several cardiac cycles with respiration suspended at end-expiration. The data from the cineradiograms were used to assess whether all three crystals were in the same midwall plane, whether one leg of the crystal triangle was properly aligned with the left ventricular long axis at end-diastole, and whether the segment lengths and strains determined by the ultrasonic technique were comparable when determined independently from radiographic measurements.

**Figure 2.** Representative tracing is shown with electrocardiogram (EKG), pressure tracings from the central aorta (Ao Pressure), left ventricle (LV, low and high gain), and left atrium and the segment length signals from three sides of crystal triangle (segments A, B, and C). In this example, segment A was oriented parallel to left ventricular long axis. Vertical lines denote time of end-diastole (ED), aortic valve opening (AVO), aortic valve closure (AVC), and mitral valve opening (MVO). AVO, AVC, and MVO were determined by crossover pressures between central aorta (AVO, AVC) or left atrium (MVO) and left ventricle.
Experimental Protocol

In each animal, the influence of ventricular volume on two-dimensional finite deformations in the left ventricular anterior midwall was assessed. Measurements were obtained at three levels of left ventricular end-diastolic pressure (LVEDP), at approximately 2, 8, and 17 mm Hg. Low levels of LVEDP were obtained by inflation of the inferior and/or superior venae cavae cuffs. Higher levels of LVEDP were produced by intravenous infusion of 6% dextran in 0.9% saline. The animals were allowed to reach steady state after each volume manipulation before measurements were obtained.

Data Analysis

All data were played back from FM tape and converted from analog to digital data at 5-msec intervals using a minicomputer (Digital Equipment Corporation, model PDP 11/03). Data from 10 cardiac cycles were averaged. For timing purposes, two events in the cardiac cycle were defined: 1) left ventricular end-diastole was taken from the left ventricular pressure tracing as the aortic valve closure was defined by transposing the central pressure trough following atrial systole. When this was not readily apparent, the peak of the R wave on the electrocardiogram was used, and 2) the time of aortic valve closure was defined by transposing the central aortic pressure at the dicrotic notch onto the high-fidelity left ventricular pressure tracing. For convenience, the term "end-systole" is used in the present study to denote the time of aortic valve closure.

In the three dogs with high-speed biplane cineradiography, the lateral and anterior-posterior films were projected separately onto a digitizing pad and analyzed. A laboratory-developed computer program was used to reconstruct the three-dimensional position of each of the three crystals and the five reference beads. The three epicardial reference beads placed at the entry sites of the three crystals were used to define a local epicardial tangent plane. The depth of each of the three crystals from this plane was determined to evaluate how well the crystal triangle was placed in the midwall plane. A local cardiac coordinate system (circumferential, longitudinal, and radial coordinate directions) was constructed from the five reference markers. The angle between the longitudinal direction of the left ventricle and the reference leg of the crystal triangle was measured to evaluate how closely the two were aligned. Postmortem examinations were also carried out to measure the depth of each crystal from the local epicardial surface.

Strain Analysis

Finite strains are calculated directly from an equation that defines a symmetric strain tensor, $E_{ij}$. Here, strains are related to small (but finite) distances between crystals before ($\Delta s$) and after ($\Delta s$) deformation in the following way:

Here, $\Delta s$ are the coordinate components of the triangle edges in the end-diastolic configuration with reference to a Cartesian cardiac coordinate system. This equation is applied to three line segments in the original configuration (end-diastole) and in the deformed configuration (at any time after end-diastole). This provided a set of three simultaneous linear algebraic equations that are solved to calculate the three independent components of the symmetric strain tensor. The three strains consist of two normal strains (normal in the parlance of solid mechanics), circumferential ($E_{11}$) and longitudinal ($E_{22}$) strains, and one in-plane shear strain ($E_{12}$). Because of the symmetry of the strain tensor, an algebraic eigenvalue problem can be solved to calculate the principal strains, $E_1$ and $E_2$, and the directions of the principal axes with respect to the starting coordinate system. As is well known in continuum mechanics, the magnitudes of the principal strains are independent of the reference coordinate system, and the principal axes are mutually orthogonal. Therefore, in two dimensions, a single angle (in-plane angle) indicating the orientation of, say, the first principal direction with respect to the circumferential reference coordinate is sufficient to define the principal axes. In a coordinate system defined by the principal axes, all deformations are extensional (i.e., shortening or lengthening, but no shear). The principal strains are ranked arbitrarily as $|E_1| > |E_2|$. Therefore, in systole, $E_1$ is the most negative of the two principal strains and represents the greatest shortening. For the diastolic strains, $E_1$ is the most positive of the principal strains and represents the greatest lengthening.

The end-diastolic configuration (at each ventricular volume) was used as the reference configuration for systolic strain calculations. Strains were computed at 5-msec intervals following end-diastole. End-systolic strain data were calculated from end-diastole to the time of aortic valve closure.

End-diastolic strain calculations were referenced to the end-diastolic configuration at the lowest LVEDP available, analogous to a low pressure $P_L$ used to normalize uniaxial shortening measurements. This reference configuration, obtained at LVEDP of 1.5 ± 1.2 mm Hg, provided an estimate of the end-diastolic configuration at a zero transmural LVEDP. It was not always possible to obtain a configuration at an LVEDP of 0 mm Hg because of severe hypotension and arrhythmias. The "deformed" configurations were the end-diastolic configurations at each of the higher LVEDP.

Statistical Analysis

All data were expressed as mean ± 1 SD. The statistical significance of changes in end-systolic strains and end-diastolic strains as LVEDP increased (from low to mid and high values) were tested by analysis of variance for repeated measures. A $p$ value of less than 0.05 was considered to indicate statistical significance.

Results

The mean LVEDP values used for the end-systolic strain analysis were 2.3 ± 1.2, 8.0 ± 0.8, and 17.0 ± 1.9 mm Hg for the low, mid, and high volumes,
respectively. For convenience, these will be subse-
quently referred to as LVEDP levels of 2, 8, and 17 mm
Hg. The mean heart rate was 102 ± 25 beats/min and
did not differ significantly between the three LVEDP
levels. For the end-diastolic strains, the reference
configuration was determined at a LVEDP of 1.5 ± 1.2
mm Hg. The end-diastolic strains in the "deformed"
configurations (see "Strain Analysis") were determined
at LVEDP of 4.1 ± 0.5, 7.7 ± 0.8, and 16.1 ± 1.2 mm
Hg for the low, mid, and high volumes, respectively.

Systolic Strains

A typical tracing of the pressure and segment length
data from a crystal triangle is shown in Figure 2. Note
that all three segments of the crystal triangle shorten
(although to varying degrees) during systole. In this
example, segment A was aligned parallel to the left
ventricular long axis and shortened less than the
other two segments of the crystal triangle (segments B
and C).

A typical plot of the strain data is shown in Figure
3. Graphs in the left, middle, and right columns display
strain data as a function of time at low, mid, and high
LVEDP, respectively. Finite strains (referred to a
cardiac coordinate system) are shown in the top row of
graphs. These strains were most negative near the time
of aortic valve closure (end-systole). As LVEDP
increased, the end-systolic circumferential strain ($E_{c}$)
became progressively more negative, while the end-
systolic longitudinal strain ($E_{l}$) and the in-plane shear
($E_{s}$) did not change significantly. The middle row of
graphs show the principal strains. As LVEDP in-
creased, the first principal strain ($E_{1}$) became more
negative, while there was less of a change in the second
principal strain ($E_{2}$). In each case, $E_{1}$ was associated
with the greatest shortening. The first principal strain
was significantly more negative than the circumferen-
tial strain at the low LVEDP, but these two strains were
similar in magnitude at the mid and high LVEDP. In
the bottom row of graphs, the in-plane angle between

![Figure 3. Representative graphs from one animal show strain data as function of time in cardiac cycle at low left ventricular
end-diastolic pressure (LVEDP) (left panel), mid LVEDP (middle panel), and high LVEDP (right panel). In this specific example,
the low, mid, and high LVEDP were 3, 8, and 14 mm Hg, respectively. Top row plots finite strains: circumferential ($E_{c}$), longitudinal
($E_{l}$), and in-plane shear ($E_{s}$) strains. Middle row shows first principal strain ($E_{1}$), which is greatest shortening regardless of direction,
and second principal strain ($E_{2}$), which is strain perpendicular to $E_{1}$. Bottom row shows first principal direction, as measured by
in-plane angle (counterclockwise, positive) between direction of greatest shortening and circumferential (0°) direction. Vertical lines
in each graph mark time of end-diastole (ED) and aortic valve closure (AVC) for cardiac cycle (see text for additional discussion).]
FIGURE 4. First principal direction (in-plane angle) at time of aortic valve closure is shown for each of six animals. Each animal is identified by a unique symbol so that change in this angle as a function of left ventricular end-diastolic pressure (LVEDP) can be seen. Note that at low LVEDP, in-plane angle deviates from circumferential (0°) direction. As LVEDP increased, in-plane angle in each animal rotated closer towards circumferential direction (see text for additional discussion).

The first principal direction at the time of aortic valve closure is shown for each of the six animals in Figure 4. At low LVEDP, it was significantly different from the circumferential direction. As LVEDP increased, the first principal direction rotated progressively towards the circumferential or hoop direction in all animals. To evaluate the accuracy of using the circumferential strain to estimate the greatest deformation, the ratio of the end-systolic circumferential strain ($E_{11}$) to the first principal strain ($E_1$) at the time of aortic valve closure was calculated. This ratio is shown for each of the six animals in Figure 5. At low LVEDP, the end-systolic circumferential strain was significantly less negative than the first principal strain, and the mean ratio $E_{11}/E_1$ was 0.6. As LVEDP increased, the magnitudes $E_{11}$ and $E_1$ were more similar, and the ratio $E_{11}/E_1$ approached unity.

Table 1 summarizes the systolic strain data at the time of aortic valve closure. As LVEDP increased from low to mid and high levels, the circumferential strain ($E_{11}$) and the first and second principal strains ($E_1, E_2$) became significantly more negative, while there was a significant increase in the ratio $E_{11}/E_1$. The in-plane shear ($E_{12}$) was significant when compared with the normal strains ($E_1$ and $E_2$) at the low LVEDP, but it decreased significantly (became less positive) as LVEDP increased. The end-systolic circumferential strain $E_{11}$ was significantly more negative than the longitudinal strain $E_{22}$ at the mid and high LVEDP. The longitudinal strain $E_{22}$ did not change significantly with changes in LVEDP. The presence of comparable normal strains and significant in-plane shear at low LVEDP was concomitant with an in-plane angle close to $-45°$. There was a significant shift in the in-plane angle as LVEDP increased from the low to high level.

End-Diastolic Strains

The data for end-diastolic strains (see "Materials and Methods") are summarized in Table 2. The end-diastolic circumferential strain $E_{11}$, longitudinal strain $E_{22}$, first principal strain $E_1$, and second principal strain $E_2$ all became significantly more positive as LVEDP increased from low to mid and high levels. At the high LVEDP, $E_{11}$ was significantly more positive than $E_{22}$. The end-diastolic in-plane shear $E_{12}$ was small at the low LVEDP and did not change significantly with increasing LVEDP. Unlike the systolic strains, there was no significant change in the end-diastolic ratio $E_{11}/E_1$ as LVEDP increased. In addition, there was no significant change in the end-diastolic first principal
were used to successfully orient segment length gauges well with our prior studies, where epicardial landmarks ventricular long axis was 2.4° and 6.9° (determined biplane cineradiography. These segment lengths were determined by both the ultrasonic technique and from ventricular midwall plane that was reasonably parallel within 15° of the internal 2 axis. In the third animal, cineradiography was performed to verify the position the long axis. In this animal, the reference leg of the crystal triangle deviated 28° from the external long axis. Thus, there was an excellent correlation between the segment lengths as determined by the ultrasonic and radiographic techniques (Panel A). This linear relation was parallel to the line of identity but shifted to the left. Thus, the segment lengths were consistently longer (by approximately 2 mm) when determined radiographically as compared with the ultrasonic crystal technique. There were six segment lengths measured in the two animals (three segment lengths from each triangle in each of the two animals). The correlation coefficients for these six segment lengths (i.e., comparing measurements by the ultrasonic versus radiographic technique) ranged from r = 0.77 to r = 0.96, with a mean value of 0.89 ± 0.07. There was also a linear correlation between the finite strains as determined by the two techniques (Panels B and C). However, unlike the segment lengths, this relation centered around the line of identity. The correlation coefficient for the finite strains ($E_{11}$, $E_{22}$, and $E_{12}$) ranged from r = 0.80 to r = 0.96 in the two animals. The correlation coefficients for the principal strains ($E_1$ and $E_2$) and the first principal direction (Panel D) ranged from r = 0.81 to r = 0.97 in the two animals. Thus, there was an excellent correlation between the strain data calculated from the ultrasonic measurements as compared with the same data calculated from the radiographic measurements.

**Discussion**

We have described a technique to measure two-dimensional finite deformations in a local region of myocardium. This technique used three piezoelectric crystals implanted in a triangular array, with each

**Table 1. End-Systolic Strain Data**

<table>
<thead>
<tr>
<th>LVEDP (mm Hg)</th>
<th>Low (2.3 ± 1.2)</th>
<th>Mid (8.0 ± 0.8*)</th>
<th>High (17.0 ± 1.9*)</th>
</tr>
</thead>
<tbody>
<tr>
<td>$E_{11}$</td>
<td>-0.06 ± 0.05</td>
<td>-0.11 ± 0.01*</td>
<td>-0.13 ± 0.01*</td>
</tr>
<tr>
<td>$E_{22}$</td>
<td>-0.04 ± 0.04</td>
<td>-0.07 ± 0.03</td>
<td>-0.06 ± 0.03</td>
</tr>
<tr>
<td>$E_{12}$</td>
<td>0.05 ± 0.02</td>
<td>0.02 ± 0.02*</td>
<td>0.02 ± 0.02*</td>
</tr>
<tr>
<td>$E_1$</td>
<td>-0.11 ± 0.02</td>
<td>-0.12 ± 0.01</td>
<td>-0.14 ± 0.02*</td>
</tr>
<tr>
<td>$E_2$</td>
<td>0.01 ± 0.05</td>
<td>-0.05 ± 0.02*</td>
<td>-0.05 ± 0.02*</td>
</tr>
<tr>
<td>$E_{11}/E_1$</td>
<td>0.61 ± 0.21</td>
<td>0.87 ± 0.09*</td>
<td>0.94 ± 0.06*</td>
</tr>
<tr>
<td>$E_{22}/E_2$</td>
<td>-43 ± 21°</td>
<td>-26 ± 18°</td>
<td>-14 ± 13°</td>
</tr>
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</table>

Systolic strain data at time of aortic valve closure are presented as mean ± SD for six animals. Data are presented at low, mid, and high left ventricular end-diastolic pressures (LVEDP).

$E_{11}$, circumferential strain; $E_{22}$, longitudinal strain; $E_{12}$, in-plane shear strain; $E_1$, first principal strain; $E_2$, second principal strain; $E_{11}/E_1$, ratio of $E_{11}$ to $E_1$; Angle, first principal direction or in-plane angle between direction of greatest shortening and circumferential (0°) orientation.

* $p<0.05$ when compared with same measurement at low LVEDP.

**Table 2. End-Diastolic Strain Data**

<table>
<thead>
<tr>
<th>LVEDP (mm Hg)</th>
<th>Low (4.1 ± 0.5)</th>
<th>Mid (7.7 ± 0.8*)</th>
<th>High (16.1 ± 1.2*)</th>
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<tr>
<td>$E_{11}$</td>
<td>0.04 ± 0.03</td>
<td>0.17 ± 0.08</td>
<td>0.30 ± 0.08*</td>
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<tr>
<td>$E_{22}$</td>
<td>0.04 ± 0.05</td>
<td>0.10 ± 0.09*</td>
<td>0.17 ± 0.09*</td>
</tr>
<tr>
<td>$E_{12}$</td>
<td>0.03 ± 0.04</td>
<td>0.03 ± 0.06</td>
<td>0.02 ± 0.07</td>
</tr>
<tr>
<td>$E_1$</td>
<td>0.07 ± 0.07</td>
<td>0.20 ± 0.11*</td>
<td>0.34 ± 0.10*</td>
</tr>
<tr>
<td>$E_2$</td>
<td>0.01 ± 0.01</td>
<td>0.07 ± 0.05*</td>
<td>0.14 ± 0.07*</td>
</tr>
<tr>
<td>$E_{11}/E_1$</td>
<td>0.71 ± 0.29</td>
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<td>0.89 ± 0.06</td>
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<tr>
<td>$E_{22}/E_2$</td>
<td>1 ± 13°</td>
<td>-7.5 ± 32°</td>
<td>-8.5 ± 28°</td>
</tr>
</tbody>
</table>

End-diastolic strain data are presented as mean ± SD for six animals. End-diastolic reference configuration was determined at a left ventricular end-diastolic pressure (LVEDP) of 1.5 ± 1.2 mm Hg. End-diastolic strain data are presented at low, mid, and high LVEDP. $E_{11}$, circumferential strain; $E_{22}$, longitudinal strain; $E_{12}$, in-plane shear strain; $E_1$, first principal strain; $E_2$, second principal strain; $E_{11}/E_1$, ratio of $E_{11}$ to $E_1$; Angle, first principal direction or in-plane angle between direction of greatest shortening and circumferential (0°) orientation.

* $p<0.05$ when compared with same measurement at low LVEDP.
crystal simultaneously focused on the other two crystals of the triangle. Each crystal was used for dual functions, to both transmit and receive ultrasonic signals from the other two crystals. This permitted the continuous measurement of all three segment lengths of the crystal triangle. If the orientation of one leg of the crystal triangle with respect to a cardiac reference coordinate system (e.g., the left ventricular long axis) is known, then the three segment lengths of the crystal triangle can be used to determine the circumferential \( (E_\theta) \), longitudinal \( (E_z) \), and in-plane shear \( (E_{12}) \) strains. The two principal strains also could be determined. These principal strains are independent of the original reference coordinate system. The first principal strain \( (E_\theta) \) is the greatest deformation regardless of direction. The second principal strain \( (E_z) \) is the strain perpendicular to \( E_\theta \). The first principal direction (associated with the greatest strain) is quantified by measuring the in-plane angle between \( E_\theta \) and the circumferential coordinate direction (circumferential coordinate direction of \( 0^\circ \)).

There are several advantages to this technique. Only three crystals are required to provide a comprehensive description of the two-dimensional deformation of a local region of myocardium. If spherical crystals are carefully selected, it is relatively easy to implant and simultaneously focus the three crystals so that each crystal both transmits and receives ultrasonic signals from the other two crystals. Only simple alterations in the wiring of the crystals to a standard sonomicrometer amplifier are required to allow crystals to perform this dual function. This ultrasonic crystal technique provides a high-fidelity method to quantify regional myocardial function, with excellent temporal and spatial resolution. Because of the relative ease and flexibility of this technique, it is not difficult to perform several interventions and analyze the regional alterations in two-dimensional finite deformations. This technique provides several practical advantages over previously described methods for measuring two-dimensional myocardial deformations.1417

The accuracy of this ultrasonic crystal triangle technique was verified by an independent measure of regional segment lengths and finite strains. The segment lengths determined by the ultrasonic crystal technique were linearly related to the segment lengths determined by biplane cineradiography. Although the slope of this relation was parallel to the line of identity, the data were shifted leftwards by an average of 2 mm. Thus, the ultrasonic crystals consistently measured segment lengths that were approximately 2 mm shorter than those determined by the radiographic techniques. This discrepancy is related to the crystal lens. The ultrasonic signal generated by the piezoelectric crystals is transmitted much more rapidly through the crystal lens than through myocardial tissue. Therefore, the ultrasonic technique measured the distance between the two edges of the crystal lens, rather than between the piezoelectric crystals themselves. In contrast, the distance between the midpoints of the piezoelectric crystals measured radiographically was measured with the radiographic technique. This discrepancy accounts for the longer segment lengths determined by the ultrasonic crystal technique. This discrepancy was measured radiographically than ultrasonically. Although there was a disparity in segment length measurements, there was an excellent correlation between the finite strains determined by the two techniques.
correlation was possible because the changes in segment length were small in comparison to the absolute segment lengths.

In the present study, regional myocardial deformations were measured in a local plane determined by the three ultrasonic crystals. We chose to examine deformations in the midwall plane, in a plane parallel to the local epicardial tangent plane. Using epicardial landmarks and mechanical "stops" in our crystal introducer sleeves, we were able to implant ultrasonic crystals consistently in a desired orientation (i.e., with one leg of the triangle aligned parallel to the left ventricular long axis) and to a desired depth (i.e., in the midwall). The current technique would be equally applicable for examining two-dimensional finite deformations in any other plane, for example, in the subepicardium or in the subendocardium.

An important assumption of this method is that the finite strains of the midwall plane adequately describe the regional deformation. This technique does not account for radial strains (wall thickening) or shear strains in the two coordinate planes mutually orthogonal to the epicardial tangent plane. Furthermore, it is presumed that the three-dimensional orientation of the first and second principal directions is within the midwall plane. These assumptions appear reasonable as demonstrated in our prior studies.\(^1^\)\(^2\)\(^3\) In those studies, three-dimensional finite deformations were measured with a generalization of the current method.

Three columns of radiopaque beads were implanted in the anterior wall of the left ventricle to determine the transmural finite strains. In the anterior midwall, the orientation of the first and second principal directions were located reasonably well within the midwall plane.\(^11\)\(^13\) Nevertheless, a rigorous comparison of two- and three-dimensional methods awaits further study.

This technique does not account for rigid body motions or rotation, such as those described by Meier et al.\(^1^8\) However, rigid body rotation in itself is not part of the strain tensor and does not contribute to the strain energy of the observed volume of muscle or to the stresses that may develop in the muscle. Although this technique does not describe the rigid body motion, torsional or twisting forces that deform the myocardium locally are measured as a shearing deformation (\(E_{\text{sh}}\)).

A potential limitation of this technique occurs if the curvature of the ventricle is significant in the region of the crystal triangle. If this occurs, the plane measured by the crystal triangle would not represent a homogeneous plane of fibers with similar orientations but would rather represent an admixture of fiber layers with different orientations. The magnitude of this curvature artifact can be estimated. If the separation between two crystals is 1.0 cm at end-diastole and 0.9 cm at end-systole, then the measured strain will be \(-0.0950\). The actual strain will be less due to the curvature artifact. Based on the study by Rankin et al,\(^3\) if we estimate the midwall minor axis diameter to be 5.0 cm and this diameter shortens 20% during systole, then the actual strain of the "fibers" between the two crystals is \(-0.0935\). Thus, the error in strain (i.e., observed versus actual strain) due to a curvature artifact is \(<0.002\), or \(<2\%\). This corresponds to an error in the principal direction of \(0.5\). The magnitude of these errors in strains and principal directions are very small, smaller than the spatial resolution of current radiographic and ultrasonic techniques. These strain errors are also smaller than the significant changes in strain measured. Thus, errors due to the curvature of the ventricle appear to be only a minimal limitation of this technique.

To test the utility of this technique, the end-systolic and end-diastolic regional in-plane myocardial deformations accompanying changes in ventricular volume were examined. At low LVEDP, the greatest shortening deformation in the left ventricular anterior midwall occurred in a direction that was significantly different from the local circumferential fiber orientation. At end-systole, the first principal direction was oriented approximately \(-45^\circ\) (rotated clockwise) from the circumferential direction. As a result, the end-systolic circumferential strain (\(E_{\text{c}}\)) significantly underestimated the greatest systolic deformation (\(E_{\text{s}}\)), and the mean value of the ratio \(E_{\text{s}}/E_{\text{c}}\) was 0.6. As LVEDP increased, the first principal direction rotated closer to the circumferential direction. At higher LVEDP, \(E_{\text{c}}\) closely approximated the value of \(E_{\text{s}}\), and the ratio \(E_{\text{s}}/E_{\text{c}}\) approached unity.

These systolic results demonstrate that a significant amount of myocardial deformation occurs in a direction different from the local fiber orientation, particularly at low LVEDP. These findings are in agreement with the previous observations by Meier et al\(^1^8\) in the right ventricle and in left ventricular studies from this laboratory.\(^1^1\)\(^1^3\) In addition, this study demonstrates that the first principal direction changes significantly as the volume increases. Therefore, if regional ventricular function is assessed by measuring the change of a myocardial dimension in only a single, fixed direction (e.g., with a circumferential segment or a minor axis chord), the greatest myocardial deformation may be accurately estimated at mid and high LVEDP but will be significantly underestimated at low LVEDP. There are several important implications of these results. First, the disparity between measurements of circumferential and greatest deformation may lead to an error in the interpretation of the magnitude of the Frank-Starling effect. For example, an increase in ventricular volume may produce an increase in minor axis chord or circumferential segment shortening. However, this greater shortening may be due to a rotation in the first principal direction more in line with the circumferential orientation (i.e., in the direction of measurement) and not due to a greater regional myocardial shortening per se. Second, there were significant systolic deformations in a direction perpendicular to the first principal direction, as reflected by the second principal strain \(E_{\text{sh}}\). These deformations occurred in the longitudinal direction at the mid and high LVEDP and are not accounted for by uniaxial measurements in the circumferential direction. Third, regional myocardial...
work has been estimated using wall tension-regional area loops. Although the regional area can be estimated from two perpendicular segment lengths, the accuracy of these regional area determinations depends on how closely the orientation of the dimension measurements coincide with the orientation of the principal strains. Because the relation between the principal direction and a fixed coordinate system can vary, the magnitude of this error may vary depending on the volume.

The change in end-diastolic configuration with loading (i.e., end-diastolic deformation) was also examined. Although the reference configuration for these strains was chosen at very low transmural pressures, that configuration was not necessarily unstrained due to the possibility of residual strains. Unlike systolic strains, the end-diastolic first principal direction was not significantly different from the circumferential orientation at any LVEDP. Furthermore, there was no significant change in the first principal direction or the end-diastolic ratio $E_{1}/E_{2}$ as LVEDP increased. Accordingly, the end-diastolic strain $E_{1}$ accurately reflected $E_{1}$ over a wide range of physiologic LVEDP. Nevertheless, substantial magnitudes of the end-diastolic second principal strain $E_{2}$ occurred at both mid and high LVEDP. Therefore, despite the fact that circumferential measurements accurately reflected greatest end-diastolic deformations at mid and high LVEDP, such measurements did not fully describe the two-dimensional deformations that occurred.

The mechanisms for these findings are uncertain. It is possible that the interaction between fiber layers of differing orientations influences the principal direction of deformation. If this were the case, then the more longitudinally oriented fibers of the subendocardium and subepicardium may exert a greater influence on deformation of the midwall fibers at low but not mid or high volumes. This may occur either with a selective recruitment of proportionally greater midwall than subendocardial or subepicardial fiber shortening (as volume increases) or due to a volume-induced reduction in “tethering” effects. Partial support for this concept is derived from the systolic strain data, which demonstrate that volume loading causes the midwall to deform more in the circumferential than the longitudinal direction. This is consistent with prior studies that demonstrate that the elliptical shape of the ventricle at low volumes becomes more spherical with higher volumes. This preferential increase in diastolic strain in the circumferential direction may result in a greater influence of circumferential fibers on the principal direction of shortening.

In summary, we have described a technique for the determination of regional in-plane two-dimensional finite strains and the principal directions of deformation. This technique is relatively simple to implement and offers a much more detailed description of regional myocardial function than current unidirectional dimension measurements. We have applied this technique to examine the alterations in regional myocardial deformation that occur with different ventricular volumes. This preferential increase in diastolic strain and offers a much more detailed description of regional myocardial function than current unidirectional determination of regional in-plane two-dimensional finite strains and the principal directions of deformation during systole as ventricular volume was varied over a physiological range. These studies demonstrate the importance of considering two-dimensional finite strains as well as the principal directions of deformation for the analysis of regional wall motion.

Appendix

Segment lengths from the three sides of the crystal triangle were used to calculate a set of two normal strains and one in-plane shear (Figure 7). Because one leg of the crystal triangle is aligned parallel with the left ventricular long axis at end-diastole, we know its length, $(\Delta s_0)$, and its coordinate components, $(\Delta a_0)$, in a reference coordinate system. Coordinate components for the other two sides of the triangle are found from their lengths and an application of the Pythagoras rule as follows:

- Segment 1 ($a$): $\Delta a_1 = 0$, $\Delta a_2 = a$
- Segment 2 ($b$): $\Delta a_1 = h$, $\Delta a_2 = -\sqrt{b^2-h^2}$
- Segment 3 ($c$): $\Delta a_1 = h$, $\Delta a_2 = a-\sqrt{b^2-h^2}$

These lengths and coordinate components at end-diastole $(\Delta s_0$ and $\Delta a_0)$, together with lengths at time $t$ $(\Delta s_t)$, are substituted into the well-known quadratic form that defines a symmetric strain tensor:

$$\begin{align*}
(\Delta s^2 - \Delta s_0^2) &= (2E_{11}\Delta a_1^2 + 4E_{12}\Delta a_1\Delta a_2 + 2E_{22}\Delta a_2^2, i = 1,3 \text{ Principal strains, } E_{11}, E_{12}, \text{ and the directions of the principal axes are calculated from an algebraic eigenvalue problem and are independent of the reference system to which the coordinate components of the three sides are referred.}
\end{align*}$$

![Figure 7. Triangular arrangement of piezoelectric crystals showing three line segments used for two-dimensional finite deformation analysis. Note that one side of triangle is aligned with longitudinal coordinate direction of a local cardiac coordinate system (circumferential and longitudinal coordinates). Besides the three lengths, coordinate components of each of the three sides in cardiac coordinate system are required (see "Appendix").](http://circres.ahajournals.org/doi/abs/10.1161/01.RES.62.4.720?intcid=figurelink)
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References


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