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In Vivo Validation of Longitudinal–Circumferential Area Change Ratio to Estimate Myofiber Shortening in the Heart

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Abstract

The aim of this paper was to validate area change ratio (%AC) against myofiber shortening ($\% \lambda_f$) in the heart *in vivo*. %AC is emerging as a mechanical index that may approximate $\% \lambda_f$ by incorporating both circumferential and longitudinal shortening. However, the physiological significance of %AC remains unclear. We studied the time course of %AC in the anterior midleft ventricular wall of normal canine heart *in vivo* ($n = 14$) during atrial pacing over the entire cardiac cycle using transmurally implanted markers and biplane cineradiography (8 ms/frame). %AC was calculated as the myocardial area change relative to the elemental material area on the circumferential–longitudinal plane at the reference configuration (=end diastole). %AC was compared with $\% \lambda_f$ that was determined from the transmural fiber orientation directly measured in the heart tissue. The time course of both %AC and $\% \lambda_f$ was determined in the subepicardial, midwall, and subendocardial layers. The time course of %AC and $\% \lambda_f$ was significantly different, and the difference was more pronounced towards the endocardium. %AC consistently overestimated $\% \lambda_f$. The timing of the peak %AC was significantly delayed compared to that of the peak $\% \lambda_f$. We conclude that %AC is significantly different from $\% \lambda_f$ both in magnitude and timing *in vivo*. %AC overestimates $\% \lambda_f$, and the overestimation is worse toward the endocardial layers. This may be a potentially important limitation when applying %AC to optimization and responder identification for cardiac resynchronization therapy.

Index Terms

Biomechanics; biomedical imaging; strain

I. Introduction

Area change ratio (%AC) reflects the change in the area of a regional unit square on the longitudinal–circumferential surface of the myocardium (see Fig. 1). It is a relatively new

index of regional myocardial deformation mainly used in 3-D speckle-tracking imaging (STI) [1], and is emerging as a useful index to quantify ventricular dyssynchrony [2] and predict response to cardiac resynchronization therapy (CRT) [3]. Compared with traditional 1-D indices such as circumferential and longitudinal strains, %AC is a 2-D index that incorporates information from both the circumferential and longitudinal directions. Therefore, it is expected to minimize the error and thus improve the SNR ratio of the measurements of myocardial deformation.

Since myofibers are approximately aligned within the longitudinal–circumferential plane [4], %AC may correlate well with regional myofiber shortening ($\% \lambda_f$). However, %AC has not been validated against myofiber shortening in the heart *in vivo*, and its physiological significance remains unclear.

In this study, we directly compared the time course between %AC and $\% \lambda_f$ during the cardiac cycle to validate %AC against $\% \lambda_f$. To precisely measure the 3-D displacement field and the myofiber orientation within the myocardium, where the 3-D displacement field is measured, we used a set of three transmural columns of 0.8-mm-diameter markers in the left ventricle (LV) anterior wall and high-speed biplane cineradiography in the normal canine heart *in vivo*. %AC and $\% \lambda_f$ during the cardiac cycle were derived from the 3-D displacement field of the myocardium within the transmurally implanted markers.

II. Methods

All protocols were approved by the Animal Subjects Committee of the University of California, San Diego, which is accredited by the American Association for Accreditation of Laboratory Animal Care.

A. Experimental Protocol

Fourteen adult mongrel dogs (20–30 kg) were anesthetized with intravenous thiopental sodium (8–10 mg/kg), intubated, and mechanically ventilated with isoflurane (0.5–2.5%), nitrous oxide (3 l/min), and medical oxygen (3 l/min) to maintain a surgical plane of anesthesia. To measure 3-D myocardial deformation, three transmural columns of four to six 0.8-mm-diameter gold beads were placed within the LV anterior wall between the first and second diagonal branches of the left anterior descending coronary artery via median sternotomy [5]–[8] [see Fig. 2(a)].

After the bead insertion was complete, the platform was removed and a 1.7-mm-diameter surface gold bead was sewn onto the epicardium above each column. Gold beads (2-mm diameter) were sutured to the apical dimple (apex bead) and on the epicardium at the bifurcation of the left anterior descending and left circumflex coronary arteries (base bead) to provide end points for the long axis of the LV [8]–[11]. Bipolar pacing wires were sutured to the left atrium (LA) for atrial pacing. All pacing protocols were conducted via a square-wave, constant-voltage electronic stimulator at a frequency 10% above baseline heart rate to suppress native sinus rhythm, and stimulation parameters (voltage 10% above threshold, pulse width 8ms, and pacing cycle length) were kept constant in each animal.

Each animal was positioned in a biplane radiography system, and synchronous biplane cineradiographic images (temporal resolution = 8ms) of the bead markers were digitally acquired with mechanical ventilation suspended at end expiration. Left ventricular pressure (LVP), central aortic pressure (AoP), LA pressure (LAP), and surface electrocardiograms (ECG) were recorded simultaneously with the cineradiographic images. At the end of the study, the animal was euthanized with pentobarbital sodium and the heart was perfusion fixed with 2.5% buffered glutaraldehyde at the end-diastolic pressure measured in the study [8], [12]. Because the heart was fixed at end-diastolic pressure, fiber orientations in the fixed hearts were assumed to represent the fiber structure in the end-diastolic reference configuration *in vivo* [8].

B. Histology

To avoid the distortional effects of dehydration and shrinkage associated with embedding, histologic measurements were obtained using freshly fixed heart tissue. In the transmural block of tissue within the implanted bead set, the mean myofiber angle α was measured with reference to the positive circumferential direction and was determined from epicardium to endocardium at every 1-mm-thick section sliced parallel to the epicardial tangent plane [8].

C. Data Analysis

The digital images from biplane X-ray were corrected for magnification and spherical distortion [8] to reconstruct the 3-D coordinates [13] of the bead markers. End diastole was defined as the time of the peak of the ECG R-wave.

Myocardial deformation was calculated from continuous, nonhomogeneous transmural distributions of 3-D finite deformation within the bead set. The deformation for each frame was calculated from the reference state at end diastole [14].

With the continuous polynomial mapping from reference position to current position, differentiation with respect to reference position gives the 3-D deformation gradient tensor $\mathbf{F}$ (= a 3×3 matrix) in the cardiac coordinate system (X_1, X_2, X_3) (1 = circumferential, 2 = longitudinal, and 3 = radial) [15], which depends on position. The Lagrangian Green's strain tensor $\mathbf{E}$ was then calculated at various values of reference wall depth as follows:

$$\mathbf{E} = \frac{1}{2}(\mathbf{F}^T \mathbf{F} - \mathbf{I}) = \begin{pmatrix} E_{11} & E_{12} & E_{13} \\ E_{21} & E_{22} & E_{23} \\ E_{31} & E_{32} & E_{33} \end{pmatrix}$$

where $\mathbf{F}^T$ is the transpose of $\mathbf{F}$ and $\mathbf{I}$ is the identity matrix. The three normal strain components E_{11} , E_{22} , and E_{33} reflect myocardial stretch or shortening along the circumferential, longitudinal, and radial axes, respectively. The three shear strains E_{12} , E_{13} , and E_{23} represent angle changes between pairs of the initially orthogonal coordinate axes.

To relate the finite strains to the local 3-D structure of the LV wall, a local myofiber coordinate system (X_f, X_{cf}, X_3) was constructed in each heart, which defines the myofiber axis (X_f), the cross-fiber axis (X_{cf}), and the radial axis (X_3). With values of myofiber angle α

at a given depth obtained from the fitted transmural distributions, an orthogonal transformation was performed to convert the strain tensor $\mathbf{E}$ from the local cardiac coordinate system to the local myofiber coordinate system.

$$\mathbf{E}^{\text{fiber}} = \mathbf{M} \mathbf{E}^{\text{cardiac}} \mathbf{M}^T$$

where $\mathbf{M}^T$ is the transpose of the 3×3 coordinate transformation matrix given by the following [14]:

$$\mathbf{M} = \begin{pmatrix} \cos\alpha & \sin\alpha & 0 \\ 0 & 0 & 1 \\ \sin\alpha & -\cos\alpha & 0 \end{pmatrix}.$$

The resulting “fiber” strains ($\mathbf{E}^{\text{fiber}}$) include stretch or shortening along the myofiber (E_{ff}).

$$E_{ff} = E_{11} \cos^2 \alpha + 2E_{12} \sin \alpha \cos \alpha + E_{22} \sin^2 \alpha.$$

Myofiber shortening in each direction is represented as percentage

$$\% \lambda_1 = (\lambda_1 - 1) \times 100 \text{ (\%, circumferential shortening)}$$

$$\% \lambda_2 = (\lambda_2 - 1) \times 100 \text{ (\%, longitudinal shortening)}$$

$$\% \lambda_3 = (\lambda_3 - 1) \times 100 \text{ (\%, radial thickening)}$$

$$\% \lambda_f = (\lambda_f - 1) \times 100 \text{ (\%, myofiber shortening)}$$

where λ_1 , λ_2 , λ_3 , and λ_f are stretch ratios that represent the relative length normalized to its length at end diastole

$$\begin{aligned} \lambda_1 &= \sqrt{2E_{11} + 1} \\ \lambda_2 &= \sqrt{2E_{22} + 1} \\ \lambda_3 &= \sqrt{2E_{33} + 1} \\ \lambda_f &= \sqrt{2E_{ff} + 1}. \end{aligned}$$

In an infinitesimal volume element in the reference configuration [see Fig. 2(b)] with edges parallel to the local cardiac coordinate axes given by

$$\begin{aligned} d\mathbf{X}_1 &= dX_1 \mathbf{e}_1 \\ d\mathbf{X}_2 &= dX_2 \mathbf{e}_2 \\ d\mathbf{X}_3 &= dX_3 \mathbf{e}_3 \end{aligned}$$

where e_1 , e_2 , and e_3 are the orthogonal unit vectors along the circumferential (=1), longitudinal (=2), and radial (=3) axes, respectively, the elemental material area dA on the circumferential–longitudinal plane at time = 0 is given as follows:

$$dA = dX_1 dX_2.$$

The spatial vectors in the deformed configuration are given by

$$\begin{aligned} dx_1 &= \mathbf{F} d\mathbf{X}_1 = dX_1 \mathbf{F} e_1 \\ dx_2 &= \mathbf{F} d\mathbf{X}_2 = dX_2 \mathbf{F} e_2 \\ dx_3 &= \mathbf{F} d\mathbf{X}_3 = dX_3 \mathbf{F} e_3. \end{aligned}$$

The norm of the vector product of these elemental vectors on the circumferential–longitudinal plane gives the deformed area da as follows:

$$da = |dx_1 \times dx_2| = |\mathbf{F} e_1 \times \mathbf{F} e_2| dX_1 dX_2.$$

Therefore, the myocardial area change AC relative to the unit myocardial area (=1.00) at the reference configuration (=end diastole) is given by

$$AC = \frac{da}{dA} = |\mathbf{F} e_1 \times \mathbf{F} e_2|.$$

Area change ratio was defined as percent change of AC relative to its area at end diastole

$$\%AC = (AC - 1) \times 100(\%).$$

The time course of each variables was determined at three wall depths: 20% (subepicardium), 50% (midwall), and 80% (subendocardium) wall depth from the epicardial surface.

D. Statistical Analysis

Values are mean $\pm$ SD. Student's paired t -tests were performed to compare the peak shortening values and the time of peak shortening. Statistical analysis was performed using SigmaStat 3.0 (SPSS, Inc., Chicago, IL).

III. Results

The site of myocardial deformation measurement was located at $68 \pm 11\%$ of the distance from base to apex along the LV long axis in a region of the anterior LV free wall 1 to 2 cm septal of the anterolateral papillary muscle. The hemodynamic parameters are summarized in Table 1.

A. Myofiber Orientation

The transmural distribution of the measured myofiber angles (α) was consistent among all the animals studied (see Fig. 3). The myofiber angles (α) ranged approximately from -50° to $+80^\circ$, from epicardium to endocardium, resulting in a transmural gradient of $\sim 130^\circ$. The myofiber angle at the midwall (50% wall depth) was $9^\circ \pm 15^\circ$ off from the pure circumferential direction ($= 0^\circ$).

B. Correlation Versus Myofiber Shortening

$\% \lambda_1$, $\% \lambda_2$, $\% \lambda_3$, and $\% AC$ showed significant correlation with $\% \lambda_f$ at three wall depths (subepicardial, midwall, and subendocardial layers) ($P < 0.01$) (see Fig. 4). Overall, correlation between $\% \lambda_1$ and $\% \lambda_f$ at midwall showed the highest correlation coefficient ($R^2 = 0.95$), which likely reflects the fact that the myofiber orientation is close to the circumferential direction at midwall. Correlation between $\% AC$ and $\% \lambda_f$ was excellent at subepicardial and midwall layers ($R^2 > 0.84$), but it was worse at the subendocardial layer ($R^2 = 0.68$). The slope of the least square line was progressively larger toward the endocardium, suggesting that $\% AC$ overestimates $\% \lambda_f$. However, at the subendocardial layer, $\% AC$ showed the best correlation with $\% \lambda_f$ ($R^2 = 0.68$) compared to $\% \lambda_1$ ($R^2 = 0.51$), $\% \lambda_2$ ($R^2 = 0.60$), and $\% \lambda_3$ ($R^2 = 0.64$).

C. Time Course of Area Change Ratio

Although the time course of $\% AC$ roughly followed that of $\% \lambda_f$, there were two distinct differences (see Fig. 5) between these two indices. First, the peak change of $\% AC$ was significantly greater than that of $\% \lambda_f$ (see Fig. 6). The difference was relatively small in the subepicardial layer ($\% \lambda_f - 10.7 \pm 4.5$ versus $\% AC - 14.2 \pm 4.5$, $P < 0.0001$), but was progressively more pronounced toward the endocardium ($\% \lambda_f - 14.6 \pm 4.2$ versus $\% AC - 22.0 \pm 5.4$ at midwall, $P < 0.0001$, and $\% \lambda_f - 14.2 \pm 5.7$ versus $\% AC - 28.9 \pm 6.1$ at subendocardial layer, $P < 0.0001$). There was a significant transmural gradient in the peak $\% AC$ ($P < 0.0001$) but not in the peak $\% \lambda_f$ ($P = 0.053$). Second, the time to reach the peak $\% AC$ was significantly different from that of $\% \lambda_f$ in the subepicardial and subendocardial layers (see Fig. 5). In the subepicardial layer, $\% AC$ reached the peak significantly later than the peak $\% \lambda_f$ ($+10.3 \pm 13.8$ ms, $P = 0.016$) (see Fig. 7). In contrast, in the subendocardial layer, $\% AC$ reached the peak significantly earlier than the peak $\% \lambda_f$ (-32.0 ± 54.8 ms, $P = 0.0478$). There was a significant transmural gradient in the time to reach peak $\% \lambda_f$ ($P = 0.016$) but not in the time to reach the peak $\% AC$ ($P = 1.00$).

IV. Discussion

The aim of this paper was to validate area change ratio ($\% AC$) against myofiber shortening ($\% \lambda_f$) *in vivo*. In summary, our results indicate that the following:

1. $\% AC$ correlates well with $\% \lambda_f$ in the subepicardial and midwall layers.
2. $\% AC$ correlates best with $\% \lambda_f$ in the subendocardial layer compared with other 1-D indices ($\% \lambda_1$, $\% \lambda_2$, and $\% \lambda_3$).

3. %AC significantly overestimates the magnitude of $\% \lambda_f$ and the overestimation is worse toward the endocardial layer.
4. In the epicardial layer, the time to the peak %AC follows that of the peak $\% \lambda_f$.
5. In the midwall, the time to the peak %AC is roughly equal to that of the peak $\% \lambda_f$.
6. In the endocardial layer, the time to the peak %AC precedes that of the peak $\% \lambda_f$.

The last conclusion is particularly important when clinical application is considered because the measurement of %AC using the 3-D STI technique is most accurately performed in the endocardial surface by relatively simple image segmentation (“endocardial surface tracking”). Although there are regional and species variations in the transmural myofiber orientation, we believe that the previous conclusions remain the same in any left ventricular region in any animals where myofiber orientation has been measured (i.e., dogs, pigs, sheep, and human).

A. Area Change Ratio and Myofiber Shortening

Clinicians would like to estimate regional myofiber mechanics from variables that are clinically measurable, such as 3-D tissue deformation relative to the local cardiac coordinates (longitudinal, circumferential, and radial axes). Since the imbrication angle between the epicardial tangent plane and the myofiber direction is small [4], myofibers can be considered to align within the longitudinal–circumferential plane.

Therefore, with knowledge of local myofiber orientation, myofiber function can be accurately calculated from information in the longitudinal–circumferential plane. Because the area change ratio (%AC) incorporates this information, it may approximate regional myofiber shortening ($\% \lambda_f$).

However, our results indicate that the area change ratio is a relatively poor estimate of myofiber shortening in both the magnitude and the timing. The key to understanding our results is that these two indices are similar but fundamentally different. Essentially, the area change ratio is equivalent to a product of longitudinal and circumferential shortening, whereas myofiber shortening is a weighed sum of longitudinal and circumferential shortening. It is not surprising that the area change ratio overestimates the magnitude of myofiber shortening because the area change ratio is a 2-D index, whereas myofiber shortening is a 1-D index. With the known myofiber angle (α), myofiber strain (E_{ff}) is calculated as follows (see Section II):

$$E_{ff} = E_{11} \cos^2 \alpha + 2E_{12} \sin \alpha \cos \alpha + E_{22} \sin^2 \alpha.$$

It is evident from the equation above that the myofiber strain (E_{ff}) depends only on the circumferential strain (E_{11}), longitudinal strain (E_{22}), and circumferential–longitudinal shear strain (E_{12}). The contribution, or the scale factor, of each strain component (E_{11} , E_{22} , and E_{12}) to E_{ff} is unequal, and is determined by the myofiber angle (α), a function of the wall depth (see Fig. 8). For example, in both the subepicardial ($\alpha \sim -60^\circ$) and the subendocardial ($\alpha \sim +60^\circ$) layers, the longitudinal strain contributes more (scale factor = 0.75) to the

myofiber strain than the circumferential strain (scale factor = 0.25). In the midwall layer ($\alpha \sim 0^\circ$), the myofiber strain is determined only by the circumferential strain (scale factor = 1), and the contribution of the longitudinal strain and the circumferential–longitudinal shear strain is zero (scale factor = 0). In contrast, the area change ratio equally incorporates the circumferential and the longitudinal strains. This fundamental difference accounts for the significant difference in the timing of peak change between the area change ratio and myofiber shortening. In addition, in Fig. 5, subepicardial stretching of $\% \lambda_f$ during the isovolumic contraction phase is clearly observed, while $\%AC$ shows no such stretching. This result is also consistent with our conclusion that $\%AC$ and $\% \lambda_f$ are two fundamentally different indices.

B. Physiological Significance of Area Change Ratio

We have shown that there is a significant transmural gradient in timing of peak myofiber shortening or the onset of myofiber relaxation [10]. Our results showing that the peak myofiber shortening was earlier in the subepicardial layer and later in the subendocardial layer are consistent with our previous report.

The difference in the magnitude and the timing of peak change between the area change ratio and myofiber shortening results from the fact that the area change ratio inherently includes both myofiber and crossfiber deformation. In fact, in the subendocardial layer, the crossfiber direction is the principal direction of deformation with the greatest shortening [16]. Since the cross-fiber shortening reflects the mechanics of the myolaminar sheet structure [17], the area change ratio in the subendocardial surface is a measure of sheet deformation, primarily a combination of sheet extension and sheet shear [14].

C. Clinical Implications

Although the area change ratio and the myofiber shortening are two fundamentally different indices, one may still be able to use the area change ratio as a measure of the regional myocardial deformation. There are two distinct advantages of the area change ratio over other traditional indices of myocardial deformation. First, as described previously, its SNR ratio is relatively high and it is highly reproducible. This is particularly important in 3-D STI where tracking error could lead to significant error in estimating myocardial deformation. Second, since it overestimates the magnitude of myofiber shortening, it provides a more sensitive measure of regional myocardial deformation. This may be potentially helpful for the assessment of extent of myocardial dysfunction in myocardial infarction and quantification of ventricular dyssynchrony to identify patients who would truly benefit from CRT [2], [3]. The disadvantage of the area change ratio is its inability to precisely estimate the timing of the local myofiber shortening. This may be a potentially important limitation to consider when applying the area change ratio to optimization and responder identification for CRT.

$\%AC$ derived from the 3-D echocardiography hardware clinically available at this point has much lower spatial and temporal resolution than the technique used in this study. Lower temporal resolution per se does not affect the comparison between $\%AC$ and $\% \lambda_f$ at each time point. However, lower spatial resolution and resultant spatial averaging over the

transmural layers could lead to inaccurate measurement because there is significant transmural variation in myofiber mechanics [10]. The best way to avoid such inaccuracy is to limit the %AC measurements to a specific transmural layer such as the endocardial surface.

D. Estimating Myofiber Shortening Using 3-D STI

A clinically meaningful application of 3-D STI is the estimation of myofiber shortening in human *in vivo*, which directly reflects the physiological and pathophysiological function of the heart. Three-dimensional echocardiogram allows definition of 3-D ventricular geometry. Myofiber orientation in the entire ventricles can be estimated based on a rule-based method [18], [19], where the transmural myofiber direction is determined as a function of the distance from the endocardium [20]. Studies have shown that the transmural myofiber orientation is preserved even in failing hearts [21]. The combination of the 3-D ventricular geometry, the estimated myofiber orientation, and the 3-D displacement field by 3-D STI allows accurate estimation of myofiber shortening in any ventricular regions. Real-time estimation of myofiber shortening will be extremely helpful for periprocedural quantification of ventricular dyssynchrony to identify the ideal pacing lead location for CRT.

E. Limitations

Our measurements were restricted to the midanterior LV. Therefore, our findings may not be precisely applicable to other LV regions where the transmural myofiber orientation is different. The 3-D finite deformation that we measured in open-chest, anesthetized dogs may not accurately reflect the transmural deformation in closed-chest, conscious animals. In addition, our data were restricted to myocardial deformation in normal synchronous contraction with the intact atrioventricular conduction system. We speculate that the difference between %AC and λ_f would be more pronounced in measurements in dyssynchronous hearts.

V. Conclusion

Area change ratio is significantly different from myofiber shortening both in magnitude and timing *in vivo*. Area change ratio overestimates the magnitude of myofiber shortening, and the overestimation is worse toward the endocardial layers, likely resulting from crossfiber deformation inherently included in area change ratio.

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Biographies

Hiroshi Ashikaga received the M.D. Ph.D. degrees from the University of Tokyo, Tokyo, Japan, in 1996 and 2007, respectively.

He completed his clinical training in Internal Medicine at Beth Israel Medical Center, New York, in 2001, research training in Biomedical Engineering at the University of California, San Diego, research training in Cardiovascular MRI at the National Institutes of Health, Bethesda, MD, in 2006, and clinical training in Cardiology and Cardiac Electrophysiology at the Johns Hopkins Hospital, Baltimore, MD, in 2012. He is currently a Clinical Fellow in Cardiac Electrophysiology at the Johns Hopkins Hospital. His research interests include image-guided diagnosis, intervention and device therapy for heart failure, ventricular arrhythmia, and sudden cardiac death.

Dr. Ashikaga's awards include a finalist for the Young Investigator Award at the Heart Rhythm Society in 2006, an Elite Reviewer for the *Journal of the American College of Cardiology* in 2006, the winner of the Howard S. Silverman Award at the Johns Hopkins University in 2010, the winner of the Young Investigator Award at the Asia-Pacific Heart Rhythm Society in 2011, and a finalist for the Melvin Judkins Young Investigator Award at the American Heart Association in 2011. He is a member of the American Heart Association, the American College of Cardiology, and the Heart Rhythm Society.

Jeffrey H. Omens received the B.S. degree in engineering sciences, and the M.S. and Ph.D. degrees in bioengineering from the University of California San Diego (UCSD), La Jolla, in 1983, 1985, and 1988, respectively. His M.S. and Ph.D. thesis focused on cardiac mechanics.

He is currently an Adjunct Professor of Medicine and Bioengineering and a Principal Investigator with the Cardiac Mechanics Research Group, UCSD, where he was a Postdoctoral Fellow in Cardiac Pathophysiology at the UCSD School of Medicine from 1988 to 1990. He is the author of more than 80 scientific publications looking at mechanisms of cardiac disease. His current research interests include cardiac mechanotransduction, cardiac remodeling, functional imaging, and instrumentation.

Dr. Omens is a member of American Physiological Society and American Society of Mechanical Engineers and has received several teaching honors at UCSD including the Distinguished Teaching Award in 2005 and the UCSD teacher of the year award in Engineering five times for his teaching efforts in Bioengineering.

Fig. 1.
Calculation of area change ratio (%AC) by endocardial surface tracking used in 3-D STI.

Fig. 2.

(a) Experimental setup. The transmurular bead set (~10 mm) was implanted between the first and the second diagonal branches of the left anterior descending coronary artery to measure 3-D displacement of the myocardial tissue across the wall. (b) Calculation of myocardial deformation within the bead set. Finite deformation can be calculated in an infinitesimal volume element at an arbitrary wall depth in the deformed configuration (dx_1 , dx_2 , dx_3 , time = t) relative to the reference configuration (dX_1 , dX_2 , dX_3 , time = 0).

Fig. 3. Measured myofiber (α) angle versus % wall depth. Values are mean $\pm$ SD ($n = 14$).

Fig. 4. Correlation versus myofiber shortening at three layers. The points are pooled from all time points during the cardiac cycle in 14 animals.

Fig. 5.

Myofiber shortening (black) versus area change ratio (light gray) in three transmurial layers. The data represent the average time course over the cardiac cycle of all animals ($n = 14$). The arrow in respective color indicates the peak %change from the end diastole. LV Pressure: left ventricular pressure.

Fig. 6.

Peak %change of myofiber shortening ($\% \lambda_f$) versus area change ratio ($\%AC$). The values represent the mean $\pm$ SD ($n = 14$).

Fig. 7.

Time to peak area change ratio (%AC) relative to the time to peak myofiber shortening ($\% \lambda_f$). The values represent the mean $\pm$ SD ($n = 14$).

Fig. 8. Contribution of each strain component to myofiber strain depending on myofiber angle (α). E_{11} , circumferential strain; E_{12} , circumferential-longitudinal shear strain; E_{22} , longitudinal strain.

TABLE I**Hemodynamic Parameters**

Parameter	Value
Heart rate (bpm)	114 ± 22
Cycle length (msec)	542 ± 86
LVEDP (mmHg)	10 ± 5
LVP _{max} (mmHg)	109 ± 16
dP/dt _{max} (mmHg/sec)	2322 ± 1422
dP/dt _{min} (mmHg/sec)	-1722 ± 584

Values are mean $\pm$ SD. LVEDP, left ventricular end-diastolic pressure; LVP_{max}, peak left ventricular pressure; dP/dt_{max}, peak positive slope of the left ventricular pressure rise; dP/dt_{min}, peak negative slope of the left ventricular pressure decay.