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Characterization of Three-Dimensional Myocardial Deformation in the Mouse Heart: An MR Tagging Study

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Abstract

Purpose—To develop a 3D MR tagging method that combines harmonic phase (HARP) and homogeneous strain analysis methods for quantification of regional myocardial wall motion in mice.

Materials and Methods—3D tagged images were acquired from seven C57BL/6 mice. Intersecting tag points were reconstructed and 3D strains were quantified at apical, midventricular, and basal levels. Circumferential and radial strains quantified with 2D MR tagging were compared with those calculated from 3D tagged images.

Results—Our data showed significant heterogeneity in radial, circumferential, and shear strains. Longitudinal strain was more homogeneous. The circumferential-longitudinal shear strain, a unitless measure of ventricular torsion, was positive throughout the left ventricle. There were strong correlations between 2D and 3D studies at the basal and midventricular levels.

Conclusion—This work demonstrates the feasibility of 3D characterization of cardiac function in mouse via the combination of HARP and homogeneous strain analysis.

Keywords

3D ventricular function; mouse; myocardial wall mechanics; MR tagging; strain

THE LEFT VENTRICLE (LV) of mammalian hearts has a unique complex 3D fiber and sheet structure that plays an important role in the contractile performance of the heart as a pump (1–5). Genetically manipulated mouse models are becoming increasingly popular laboratory animals in the investigation of cardiovascular diseases. The development of noninvasive methods for characterizing regional myocardial wall motion in these mice can provide new opportunities to correlate the functional alterations with structural and cellular changes. MRI tissue tracking techniques such as myocardial tagging (6–8), velocity-encoded imaging (9, 10), harmonic phase (HARP) MRI (11), and displacement-encoded imaging (12,13) have been developed for the quantification of regional myocardial deformation. The application of these techniques in transgenic or knockout mouse models may allow us to gain insights into the molecular mechanisms underlying cardiac dysfunction in various forms of heart diseases.

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Although MRI tagging has made remarkable contributions in assessing ventricular functions in normal and diseased mouse hearts (14–16), previous 2D MRI studies still face the problem of through-plane motion, ie, the movement of myocardium through short-axis (SA) planes. Further, such 2D methods cannot provide a comprehensive evaluation of the 3D motion of the myocardial wall. In the present study we aimed to develop a noninvasive method for assessing 3D myocardial wall motion patterns in the mouse heart using MR tagging. HARP method was employed for semiautomated tracking of tag lines in three orthogonal sets of tagged images that covered the whole cardiac cycle (17,18). Homogeneous strain analysis was used to determine the regional Lagrangian strains in the LV. These data provide a full 3D functional characterization of the normal mouse heart.

MATERIALS AND METHODS

Animal Preparation and Monitoring

Two-month-old C57BL/6 mice ($n = 7$) were examined for global and regional cardiac functions. Animals were anesthetized with 1% isoflurane by nose cone and placed into the coil in prone position. Electrodes were attached to the front paws and right leg. Preamplified electrocardiographic (ECG) signals were transferred through an optical fiber to a trigger unit for ECG gating and monitoring of the vital signs. A thresholding method was used to generate the trigger signal. A Lab-View-based system was developed for the monitoring of ECG and trigger signals. The amplitude of the threshold and the inhibition period were manually adjusted to provide consistent ECG gating. The animals were kept warm by blowing hot air into the magnet using a blow dryer. The heat flow and the anesthesia level were manually adjusted to maintain the heart rate. The animal protocol was approved by the Animal Studies Committee of Washington University Medical Center.

MR Imaging

MRI experiments were performed on a 4.7T Varian INOVA system (Varian Associates, Palo Alto, CA) equipped with a gradient insert (60 G/cm, 10 cm inner diameter). A 2.5-cm surface coil was built for image acquisition. A series of scout images were acquired to obtain the long-axis (LA) and SA planes. After acquiring a horizontal LA image (four-chamber view), five contiguous, tagged SA images were prescribed from apex to base by placing the SA slices parallel to the tricuspid and mitral valve plane (Fig. 1a,c). Tagged images were acquired with the SPAMM1331 tagging sequence applied immediately after the detection of the R-wave, followed by the acquisition of cine images that covered the whole cardiac cycle. The duration of the SPAMM1331 sequence was 3.5 msec. Imaging parameters were: flip angle 30°; TE, 3 msec; field of view, 4 cm × 4 cm; matrix size, 256 × 128; tag resolution, 0.6 mm; slice thickness, 1 mm; number of averages, 2. TR was adjusted such that 15 cine images were acquired to cover the whole cardiac cycle. Two sets of tagged SA images were acquired with tags in two perpendicular directions for each SA plane, yielding a grid tagging pattern when the two datasets were combined.

To allow the quantification of 3D myocardial wall motion, a third set of tagged LA images were acquired from four radially distributed LA views spaced every 45°, with the tags perpendicular to the LV long-axis (Fig. 1b,d). Imaging parameters were the same as those used for SA images.

Nontagged cine images, which provided better contrast between myocardium and blood, were also acquired for calculation of LV volumes and the ejection fraction (EF). These images were acquired at the same timepoints as the tagged cine images.

Image Analysis

Image analysis was performed offline with an in-house developed MatLab-based (MathWorks, Natick, MA) software package. LV contours were traced manually using B-spline with eight control points. LV volumes at end-diastole (ED) and end-systole (ES), the ejection fraction (EF), and the cardiac output (CO) were calculated accordingly. The LV was further divided into four segments, ie, septum, posterior, lateral, and anterior, for each SA slice. The intersections of the interventricular septum with the right ventricle were manually identified. The remaining LV wall was evenly divided into the other three segments.

Tag lines in SA and LA images were traced using a HARP-based semiautomatic method (17, 18). The two sets of tagged SA images were combined by multiplication to yield a grid tagging pattern. The tagging mesh in these SA images was then tracked with coupled cubic spline snakes using a HARP-based method as previously described (18). Tag lines in tagged LA images were also traced by cubic spline snakes with six control points. Representative tagged images with traced tag lines are shown in Fig. 2.

To reconstruct the 3D motion of the heart, SA and LA images were registered in a Cartesian coordinate system illustrated in Fig. 1a, with the origin located at the center of LV cavity of the midventricular short-axis slice, and the z-axis aligned with LV long-axis. Displacement of those material points that were initially located at the intersections of the three undeformed orthogonal tagging planes was calculated using an iterative point-tracking technique similar to the approach proposed by Moore et al (19).

Specifically, the initial coordinates of these displaced material points were taken directly from the corresponding tag points in SA images, with the z-coordinates the same as the SA image plane. Intersecting lines of the two tagging planes orthogonal to the SA imaging planes were generated by cubic spline snakes of these tag points (Fig. 3a,b). Calculation of the intersecting points of the corresponding tag lines in the LA images with these spline snakes yielded the new z-coordinates for these displaced material points. The x-y coordinates of these points were then updated to those of the corresponding points on the spline curve with new z-coordinates. New spline snakes were generated from the updated x-y-z coordinates, followed by next iteration of the same process until the new coordinates were less than 0.01 mm from the previous iteration. Figure 3c,d illustrates the reconstructed tagging markers in 3D myocardial coordinates at ED and ES.

Strain Calculation

Homogenous strain analysis was employed to calculate the 3D strain tensor once the 3D displacement of tagging markers was reconstructed. Specifically, myocardium was divided into tetrahedrons using four non-coplanar adjacent tag points as the vertices. The six components of the symmetric 3D strain tensor were calculated from the deformation of these tetrahedrons from end-diastole. The 3D Lagrangian strain tensor was 1 calculated from the

relationship $E = \frac{1}{2}(F^T F - I)$, where F is the 3D gradient tensor, and I is the identity matrix.

The 3D tensor was further diagonalized to yield principal strains E_1 , E_2 , and E_3 , with E_1 and E_3 being the maximal and minimal strains, respectively. Orientation of the E_1 strain was determined by calculating the prime angle θ between the corresponding eigenvector and the radial direction. For each tetrahedron the strain tensor was also transformed to a local myocardial coordinate system defined by the radial (r), circumferential (c), and longitudinal (l) directions (Fig. 1a), yielding three normal strains (E_{rr} , E_{cc} , E_{ll}) represented by the diagonal elements in the transformed strain tensor, and the three shear strains (E_{rc} , E_{cl} , E_{rl}) from the off-diagonal elements.

In addition to Lagrangian strains, ventricular twist and torsion were also quantified as described previously (20). Average twist angles in the corresponding segments were calculated. Torsion was calculated as the difference in twist angles between apical and basal slices, normalized by the distance between the two slices.

Statistics

All measurements are presented as mean $\pm$ SD. Mean values of myocardial strains in septal, anterior, lateral, and posterior segments, or at basal, midventricular, and apical levels, were compared separately by one-way analysis of variance (ANOVA). If there were statistical differences, multiple pairwise comparisons were performed using Tukey's test with a confidence interval of 95%. *P* values less than 0.05 were considered statistically significant.

RESULTS

Animal Characteristics

The seven animals had a mean age of 7.3 ± 2.3 weeks. The body weight (BW) and heart weight (HW) were 26.4 ± 1.6 g and 140 ± 20 mg, respectively, yielding a HW to BW ratio of 5.17 ± 0.45 mg/g. Global functional parameters were determined from the cine SA images. The mean heart rate during the MRI study was 439 ± 57 bpm; LV volumes were 48 ± 10 μ L at ED, and 18 ± 10 μ L at ES. Ejection fraction and cardiac output were $62.7 \pm 8\%$, and 13.12 ± 2.97 mL/min, respectively.

Principal Strains

Typical postprocessing time for a complete dataset was ≈ 3 hours. Peak systolic principal strains (E_1 , E_2 , and E_3) at apical, midventricular, and basal levels of the heart are shown in Fig. 4a and Table 1. E_1 , an index of radial wall thickening, was positive and of the largest magnitude, while both E_2 and E_3 were of negative values. E_1 was larger at the midventricular level as compared to that at basal and apical levels ($P < 0.05$). E_2 and E_3 were less heterogeneous from base to apex ($P = \text{NS}$).

Comparison of strains in segmented regions showed that E_1 was greater at the midventricular level in the anterior-lateral wall ($P < 0.05$; Fig. 5a). Comparing with other segments, anterior E_2 was also the largest at basal and midventricular levels (Fig. 5b). E_3 demonstrated less heterogeneity with only slightly greater values in the anterior-lateral wall of the apex (Fig. 5c).

The prime angle θ between the primary eigenvector and the radial direction was also calculated (Fig. 5d). Small θ suggests the alignment of the directions of maximal lengthening and radial wall thickening. θ was less than 15° at midventricle. It increased slightly at base and apex but still remained within 20° , which is consistent with a previous 2D study (15).

Normal Strains

The normal strains in radial, circumferential, and longitudinal directions were calculated after transformation of strain tensor to the local myocardial coordinate system for each tetrahedron. As shown in Fig. 4b, radial strains were positive and of largest values among all three strains and were the largest at the midventricular level ($P < 0.05$), which indicated largest thickening of myocardium wall. Such heterogeneity was also observed in segmented regions (Fig. 6a). Circumferential and longitudinal shortening was quantified by the circumferential and longitudinal strains. Anterior wall displayed a trend of greater circumferential strain at basal and apical levels. (Fig. 6b). Longitudinal strains were relatively homogenous from base to apex and in different segments (Fig. 6c).

Twist, Torsion, and Shear Strains

Twist angles shifted from negative values at base to positive values at the apex, indicating transition from counterclockwise twist at base to clockwise twist at apex (Fig. 7a). Such a twist pattern led to positive torsion in all the segments (Fig. 7b). As a result, the circumferential-longitudinal strain (E_{cl}), a unitless measure of LV torsion, was also positive in all the segments (Fig. 7c). The other two shear strains, E_{rc} and E_{rl} , displayed greater variability among objects, with values spanning the axis of zero. These observations are similar to previous findings in humans (21).

Comparison of 2D and 3D Strain Quantification

To investigate the accuracy of strain calculation from 2D MR tagging, radial and circumferential strains in segmented regions were calculated from 2D tagged SA images only and were compared to those calculated from full 3D datasets. Figure 8 shows the quantitative comparisons of the two methods using linear regression at base, midventricle, and apex. There was strong agreement between the two methods at base and midventricle, with the correlation coefficient (R^2) above 0.80 for both radial and circumferential strains. However, R^2 was low at apex, 0.57 for radial strain, and 0.24 for circumferential strain.

DISCUSSION

Numerous studies have employed 2D MRI tagging to evaluate myocardial wall motion in the radial and circumferential directions. Such studies typically entail the acquisition of tagged SA images with tag planes perpendicular to the SA plane (22–26). In order to track the 3D motion of the heart, a third dataset of tagged LA images must be acquired, with tag planes parallel to the SA plane. Two image acquisition schemes are commonly used in acquiring the tagged LA images. Huang et al (27) used the cubic scheme with parallel LA images. Such a method has the advantage of straightforward image acquisition scheme and data reconstruction, as the three orthogonal image sets form a natural 3D Cartesian coordinate system. However, only one LA image has the full coverage of LV from base to apex, rendering the delineation of 3D wall motion incomplete at the apical region of some segments. Alternatively, LA images can be acquired that are radially distributed around LV LA (28). This radial scheme guarantees that each LA image has the full coverage of the heart, at the cost of more complicated imaging planning and data reconstruction. Given the small size of a mouse heart and the relatively large slice thickness, we used the radial scheme to reduce the errors associated with the cubic scheme in the current study.

The tagging protocol comprised of the acquisition of five SA and four LA slices with 0.6 mm tagging resolution (Fig. 1). The five consecutive 1-mm SA slices provided coverage of the majority of LV except the apical tip. The four LA slices intersected eight segments of the LV. Longitudinal displacement of the LV was interpolated from these eight segments. Such interpolation should allow accurate quantification of longitudinal displacement given the relatively homogeneous distribution of longitudinal shortening. However, with more heterogeneous myocardial wall motion under certain pathological states, eg, postinfarction, more tagged LA slices will be desirable.

Several 2D tagging studies have quantified myocardial strains in both normal and infarct mice (14–16). More recently, Young et al (29) applied 3D tagging analysis to the quantification of myocardial strains in postinfarct mouse hearts. In addition, 3D DENSE MRI has also been developed for the assessment of myocardial deformation in mice (30). Comparing with these previous studies, twist angles and radial strains (E_l and E_{rr}) obtained from the current study are in agreement with those reported previously, while the circumferential (E_2 and E_{cc}) and longitudinal (E_{ll}) strains are slightly lower. It remains to be further investigated whether these

differences were due to the differences in the physiological conditions of the mice or to the difference in the definition of the myocardial coordinate system, as Young et al used a finite element model in their data reconstruction.

Previously, myocardial strain patterns in mice and humans were compared using 2D MR tagging (20). It was found that the circumferential strain and normalized radial shortening were smaller in mouse hearts as compared to those in humans. In addition, decreased longitudinal shortening in mouse heart was also observed from nontagged cine MR images. However, a 2D study has only limited scope since it cannot provide quantification of myocardial strain in the longitudinal direction. In the current study we present a full assessment of the 3D myocardial wall motion patterns in mouse heart. Compared to a previous comprehensive study on the quantification of 3D systolic strain patterns in normal human subjects by Moore et al (21), all three principal strains (E_1 , E_2 , and E_3) and the three normal strains (E_{rr} , E_{cc} , and E_{ll}) in mice are smaller in magnitude than those in humans. Such findings are consistent with the previous 2D studies (20). The current study also confirms that myocardial strain in the longitudinal direction is smaller in mouse heart.

With 3D tagging it is also possible to quantify the CL shear strain, which is considered the scale-invariant measure of ventricular torsion. Ventricular torsion is an important index of myocardial function because changes in torsion may result in altered stress distribution and thus may play a role in the disease progression of the left ventricle. Twist angle per unit length has been used as a measure of torsion (16). In our previous study we found that torsion was significantly larger in mice than in humans (20). However, since twist per unit length is not a scale-invariant measure, such comparison may be biased due to the large differences in ventricular sizes between mice and humans. Since the underlying structural basis for ventricular torsion is the shearing motion of the myocardium, using the unitless CL shear strain as a measure of ventricular torsion may eliminate variations due to differences in ventricular size. Although direct statistical comparison is not possible, the CL shear we observed in mouse hearts were in general larger than that in humans reported by Moore et al (21). Therefore, it is possible that a mouse heart relies more on the shear strain for the ejection of blood.

One limitation of the current study is that the body temperature was not recorded during the experiment. However, heat flow and anesthesia level were constantly adjusted to maintain the heart rate at 439 ± 57 bpm, which was similar to what has been reported in the literature of the same strain at normal body temperature (14). Therefore, our measurements of local myocardial strains should be comparable to those at normal body temperature.

Homogeneous strain analysis method was employed to calculate myocardial strains in the current study. This method has been validated in a deformable silicone gel phantom by Young et al (31). It was shown that the error of noise-free strain estimation was within 10% of the model-calculated values. Therefore, the dominant source of error is associated with the inaccuracy in the selection of the intersecting tag points due to image resolution and noise, tag line smearing, and heart rate variation. Further, our previous study comparing homogeneous strain analysis and HARP also showed good agreement between the two methods. However, homogeneous strain analysis is based on the assumption that myocardial strains in each finite element are homogeneous. Therefore, it may not be adequate to characterize the heterogeneity of myocardial wall motion with high spatial resolution. Other methods such as model fitting or a pixel-based method will allow more accurate quantification of strain heterogeneity.

One major limitation of the current method is the prolonged image acquisition and analysis. While the HARP method enables semiautomatic tag tracking, image acquisition still comprises the collection of 14 tagged cine image series, as a typical 3D tagging dataset has five parallel SA in two orthogonal tagging directions and four radially distributed LA slices. As a result,

typical imaging time for a single mouse is more than 2 hours. Therefore, careful monitoring and maintenance of the body temperature and hemodynamics are critical. For some mouse models with cardiovascular diseases such prolonged data acquisition times may not be practical, as the mice may not survive the anesthesia. Nevertheless, the current method is still feasible in the detection of early cardiac dysfunction when global functional abnormalities have not been manifested. Such comprehensive evaluation of the regional myocardial function can provide important insights into disease development and progression. Furthermore, with the current advance in gradient technology, very short echo-time can be achieved, which allows the development of multiframe, multislice methods for cardiac cine imaging on mice. Such techniques will allow us to acquire multiple SA tagged images simultaneously, which will significantly reduce the data acquisition time.

Alternatively, most of the investigations of regional myocardial wall motion in mice have used 2D methods with tagged SA images only. Compared with 3D tagging, the 2D tagging method has the advantage of simpler image planning and shorter data acquisition times. However, it suffers from the through-plane motion artifact that may render 2D strain quantification method less accurate. To evaluate the accuracy of 2D tagging method, we compared radial and circumferential strains calculated with the 2D method to those with 3D analysis. Data acquired at basal and midventricular levels showed strong agreement between the 2D and 3D studies. At the apical level the correlation is less satisfactory, possibly due to fewer tagging voxels and larger variations in wall motion. The strong correlations between the two methods suggest that, despite the through-plane motion, strain quantification from 2D MR tagging can provide reliable measurements of ventricular deformation in the radial and circumferential directions, especially at base and midventricular regions. Recently, Tecelao et al (32) also reported agreement in the quantification of circumferential strain between 2D and 3D studies in human subjects. Nevertheless, 2D tagging studies cannot provide the comprehensive evaluation of the 3D motion of the myocardial wall.

The application of MRI tagging has also been limited by its prolonged postprocessing time. As a result, the utility of other MR methods that do not require laborious postprocessing have been explored for 3D characterization of myocardial wall motion. Velocity-encoded MR imaging is a motion-mapping technique that has been used to quantify myocardial wall motion with high spatial resolution. However, strain calculation involves the integration from phase-velocity data, which can lead to the propagation of measurement errors (33). Recently, multislice DENSE MRI method has been developed for the quantification of 3D displacement in mouse heart (30). While limitations in the signal-to-noise ratio leave the multiphase DENSE in mouse heart yet to be demonstrated, this method has the potential to allow the quantification of myocardial strain with high spatial resolution but rapid postprocessing.

In conclusion, a 3D MR tagging and strain analysis method was developed for the quantification of myocardial wall motion mechanics in mouse heart. The utility of such a method was demonstrated by characterizing the regional cardiac function in C57BL/6 mice. By using a SPAMM1331 tagging sequence, a tagging resolution of 0.6 mm was achieved with a temporal resolution of 8–10 msec. The reported data provide quantitative evaluation of the 3D contractile behavior of the normal mouse heart. This method should allow more comprehensive evaluation of alterations in regional 3D myocardial wall motion in transgenic mouse models of cardiovascular diseases.

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Figure 1.

3D MRI planning and coordinate systems used for 3D motion reconstruction and strain calculation. Five SA planes (a) are prescribed from LA scout image (c). Four radially distributed LA planes (b) are prescribed from SA image (d). Tagged short-axis (e) and long-axis (f) images are registered in a 3D Cartesian coordinate system (x, y, z) shown in (a). The registration of (e) and (f) is shown in (g). The calculated strain is converted to a local myocardial coordinate system (R, C, L).

Figure 2.
Representative SA and LA tagged images at end-diastole (**a, c**) and end-systole (**b, d**) superimposed with traced tag lines.

Figure 3.

a, b: Schematic diagrams of the 3D reconstruction of tagging markers at end-diastole (a) and in a deformed state (b). **c, d:** Reconstructed tagging markers at end-diastole and end-systole. At end-diastole, P_1 to P_5 are the material points located at the intersection of three orthogonal tag planes. Tag planes 1 and 2 are the two tag planes orthogonal to SA imaging planes. In a deformed state the displaced material points (P'_1 to P'_5) are determined by finding the intersecting points of the long-axis tag planes with the intersecting line of deformed tag planes 1 and 2.

Figure 4.

Peak systolic strains at apex (open bars), midventricle (gray bars), and base (solid bars). **a:** Principal strains. **b:** Radial (E_{rr}), circumferential (E_{cc}), and longitudinal (E_{ll}) strains. * $P < 0.05$ compared with the other two levels.

Figure 5.

Principal strains in segmented regions and the direction of maximal lengthening. **a–c:** E_1 , E_2 , and E_3 strains at apex (open bars), midventricle (gray bars), and base (solid bars), respectively. **d:** the prime angle, ie, the angle between the primary eigenvector of the strain tensor and the radial direction. S, Septum; P, Posterior; L, Lateral; A, Anterior. * $P < 0.05$ compared with the other two levels, # $P < 0.05$ compared with other segments of the same level.

Figure 6. Radial (a), circumferential (b), and longitudinal (c) strains in segmented regions at apex (open bars), midventricle (gray bars), and base (solid bars). S, Septum; P, Posterior; L, Lateral; A, Anterior. * $P < 0.05$ compared with the other two levels.

Figure 7.

Twist angle (a), ventricular torsion (b), and circumferential-longitudinal shear strain (c) at apex (open bars), midventricle (gray bars), and base (solid bars). S, Septum; P, Posterior; L, Lateral; A, Anterior; AV, average.

Figure 8.

Comparison of peak systolic radial and circumferential strains in segmented regions quantified by 2D and 3D tagging methods. **a–c:** Radial strain at apex, midventricle, and base, respectively; **d–f:** Circumferential strain at apex, midventricle, and base, respectively.

Table 1

Segmented Systolic Strain at Base, Midventricle, and Apex

	Septum	Posterior	Lateral	Anterior	Average
E_l Apex	0.20 ± 0.08	0.22 ± 0.09	0.21 ± 0.06	0.25 ± 0.08	0.20 ± 0.05
Midventricle	0.30 ± 0.13	0.31 ± 0.09	0.34 ± 0.15*	0.40 ± 0.15*	0.32 ± 0.08*
Base	0.22 ± 0.05	0.20 ± 0.07	0.22 ± 0.08	0.19 ± 0.05	0.20 ± 0.03
E_2 Apex	-0.08 ± 0.02	-0.08 ± 0.02	-0.07 ± 0.02	-0.07 ± 0.04	-0.07 ± 0.01
Midventricle	-0.08 ± 0.02	-0.07 ± 0.02	-0.07 ± 0.02	-0.10 ± 0.02	-0.07 ± 0.02
Base	-0.07 ± 0.02	-0.06 ± 0.01	-0.07 ± 0.02	-0.10 ± 0.02**	-0.07 ± 0.02
E_3 Apex	-0.19 ± 0.03	-0.21 ± 0.03	-0.23 ± 0.02	-0.22 ± 0.03	-0.21 ± 0.02
Midventricle	-0.19 ± 0.03	-0.19 ± 0.03	-0.19 ± 0.02	-0.19 ± 0.02	-0.19 ± 0.02
Base	-0.21 ± 0.03	-0.19 ± 0.02	-0.19 ± 0.05	-0.20 ± 0.04	-0.19 ± 0.03
E_{rr} Apex	0.12 ± 0.05	0.13 ± 0.14	0.12 ± 0.07	0.16 ± 0.09	0.12 ± 0.06
Midventricle	0.26 ± 0.10*	0.28 ± 0.08*	0.30 ± 0.15*	0.37 ± 0.15*	0.30 ± 0.07*
Base	0.14 ± 0.05	0.11 ± 0.08	0.17 ± 0.06	0.14 ± 0.07	0.12 ± 0.02
E_{cc} Apex	-0.10 ± 0.02	-0.07 ± 0.04	-0.08 ± 0.02	-0.13 ± 0.02	-0.09 ± 0.02
Midventricle	-0.12 ± 0.03	-0.12 ± 0.03*	-0.10 ± 0.03	-0.13 ± 0.03	-0.12 ± 0.02
Base	-0.10 ± 0.02	-0.06 ± 0.02	-0.08 ± 0.03	-0.12 ± 0.02	-0.08 ± 0.01
E_{ll} Apex	-0.11 ± 0.06	-0.10 ± 0.04	-0.11 ± 0.03	-0.11 ± 0.03	-0.10 ± 0.02
Midventricle	-0.11 ± 0.03	-0.12 ± 0.03	-0.12 ± 0.03	-0.13 ± 0.02	-0.11 ± 0.02
Base	-0.11 ± 0.03	-0.11 ± 0.03	-0.11 ± 0.05	-0.12 ± 0.04	-0.11 ± 0.03
E_{cd} Apex	0.03 ± 0.03	0.06 ± 0.03	0.06 ± 0.05	0.05 ± 0.02	0.06 ± 0.04
Midventricle	0.04 ± 0.02	0.05 ± 0.01	0.05 ± 0.01	0.03 ± 0.02	0.04 ± 0.01
Base	0.06 ± 0.04	0.04 ± 0.03	0.06 ± 0.05	0.05 ± 0.04	0.05 ± 0.04
E_{td} Apex	-0.11 ± 0.06*	-0.10 ± 0.05*	-0.09 ± 0.04*	-0.12 ± 0.03*	-0.09 ± 0.03*
Midventricle	0.01 ± 0.10	0.02 ± 0.06	-0.01 ± 0.06	-0.03 ± 0.05	-0.01 ± 0.04
Base	0.08 ± 0.06	0.07 ± 0.03	0.06 ± 0.03	0.05 ± 0.01	0.05 ± 0.01
E_{rc} Apex	0.04 ± 0.05*	0.05 ± 0.06*	0.08 ± 0.05	0.08 ± 0.05	0.06 ± 0.04*
Midventricle	-0.02 ± 0.05	-0.04 ± 0.07	0.03 ± 0.06	0.02 ± 0.06	0.01 ± 0.03
Base	-0.04 ± 0.05	-0.06 ± 0.03	-0.04 ± 0.05	-0.04 ± 0.03	-0.04 ± 0.03

Data are mean $\pm$ SD. "Average" represents strain values averaged over the whole short-axis slice.

* $P < 0.05$ compared with other two levels in the same segments;

** $P < 0.05$ compared with other segments at the same level.