

Chapter 8

Global electromechanics in infarct heart

8.1 Abstract

To test the hypothesis that alterations in electrical activation sequence contribute to depressed systolic function in the infarct border zone, the anatomical correlation of abnormal electromechanics and the infarct geometry was examined in the canine post-myocardial infarction (MI) heart, using a high-resolution MR-based cardiac electromechanical mapping technique. Three to eight weeks after creating an MI in six dogs, a 247-electrode epicardial sock was placed over the ventricular epicardium under thoracotomy. MI location and geometry were evaluated with delayed hyperenhancement MRI. Three-dimensional systolic strains in epicardial and endocardial layers were measured in five short axis slices using motion-tracking MRI (DENSE). Epicardial electrical activation was determined from sock recordings immediately prior to and following the MR scans. The electrodes and MR images were spatially registered to create a total of 160 nodes per heart that contain mechanical, transmural infarct extent, and electrical data. The average depth of the infarct was $55 \pm 11\%$ and the infarct covered $28 \pm 6\%$ of the LV mass. Significantly delayed activation ($>\text{mean} + 2\text{SD}$) was observed within the infarct zone. The strain map showed abnormal mechanics, including abnormal stretch and loss of the transmural gradient of radial, circumferential and longitudinal strains, in the region extending far beyond the infarct zone. In conclusion, the border zone is characterized by abnormal mechanics directly coupled with normal electrical depolarization. This indicates that impaired function in the border zone is not contributed by electrical factors, but results from mechanical interaction between the ischemic and normal myocardium.

8.2 Introduction

Depressed systolic function in the ischemic border zone with normal perfusion has been recognized for several decades (135). The presence of hypocontractile, border zone myocardium has consistently been substantiated by a variety of modalities, including echocardiography (124, 156), radiopaque bead arrays (163, 238), and most recently, MRI (84, 98, 103, 143). The abnormal mechanics of the border zone is clinically important because it may negatively affect ventricular remodeling and hypertrophy (61, 125).

The mechanism underlying the depressed function in the border zone has been explained primarily by mechanical factors. Most investigators concluded that it results from mechanical interactions, or tethering, between normal and ischemic myocardium (93, 94, 156, 163, 210, 238). However, the abnormal mechanics in the border zone may also involve a contribution of abnormal electrical activation. For example, the timing of electrical activation in the border zone may be delayed by abnormal electrical sequence in the infarct zone and depressed shortening, or relative stretch, in the border zone. A relative delay in electrical activation may augment the depressed function in the border zone, because the myofiber in the border zone has to contract against a higher afterload generated by earlier shortening of remote nonischemic zones.

It was hypothesized that alterations in electrical activation sequence contribute to depressed systolic function in the infarct border zone. To test this hypothesis, the anatomical correlation of abnormal electromechanics with reference to the infarct geometry was qualitatively and quantitatively examined in a canine model of chronic myocardial infarction. A high-resolution cardiac electromechanical mapping technique allowed acquisition of anatomical, electrical, and mechanical data with high precision in a near-simultaneous fashion (88).

8.3 Materials and Methods

All animal protocols were reviewed and approved by the Animal Care and Use Committee of the National Heart, Lung and Blood Institute.

8.3.1 Surgical procedures

Six adult mongrel dogs (18–28kg) were anesthetized with intravenous sodium thiopental (10–20mg/kg), intubated, and mechanically ventilated with a mixture of oxygen, medical air and isoflurane (1–3%). An 8Fr arterial introducer was placed in the left carotid artery through a skin cut down (1–2”) under sterile conditions. The surface ECG and the arterial pressure were recorded throughout the procedure. A bolus iv injection of heparin 1,000 IU and lidocaine 40 mg was administered, followed by a continuous iv infusion of lidocaine 110 mg/hr for the entire procedure. A 6Fr guiding catheter was advanced through the introducer to the left coronary ostium under fluoroscopic guidance, and a balloon angioplasty catheter was advanced to LAD through the guiding catheter over a guide wire (0.014”). The LAD was then occluded at its origin by inflating the balloon. Myocardial ischemia was confirmed with ST-T changes on ECG and transient decrease in arterial blood pressure upon balloon inflation. The balloon occlusion was maintained for a total of 120 minutes. Another bolus iv injection of lidocaine 40 mg was given 15 minutes prior to reperfusion to avoid serious ventricular arrhythmia. All the animals showed frequent ventricular ectopic beats after reperfusion. The catheters and the introducer were removed and the surgical wound closed. The surface ECG was recorded 3–4 times per day for the first 48–72 hours to monitor the baseline rhythm.

Three to eight weeks following MI, the animal underwent a median sternotomy under general anesthesia. The heart was placed in a pericardial cradle, and a multi-electrode epicardial sock consisting of a nylon mesh fitted with 247 silver electrodes attached in an ordered fashion was placed over the ventricular epicardium, as described previously (88). The sock was placed in a consistent and predetermined orientation for all experiments and secured with several sutures. Ten to fifteen 4-mm diameter glass beads (18 μ L) filled with an aqueous solution of gadopentetate dimeglumine (Gd-DTPA) (5mM) were attached to the sock as markers for registering mechanical and electrical data. An MR-compatible pressure micromanometer catheter (Millar, SPC-350, 5Fr) was advanced under fluoroscopic guidance to the left ventricular (LV) cavity through a 6Fr arterial introducer placed in the right carotid artery. A ground reference electrode was sewn onto the fat pad at the aortic root. All sock wires were run directly out of the chest and the animal was transported to the MR scanner.

8.3.2 Data acquisition

MR scanning was performed in a 1.5-T scanner (Siemens Sonata). LV geometry (LV end-diastolic and end-systolic volumes) and function (LV stroke volume and ejection fraction) were evaluated with a cine true FISP sequence (BW \pm 1,395 Hz/pixel, TE/TR 1.9/3.7 ms, readout flip angle 50°, FOV 225 x 300 mm, image matrix 192 x 256, spatial resolution 1.2 x 1.2 x 6.0 mm, 8–10 slices without a gap). MI geometry was evaluated with a phase-sensitive delayed hyperenhancement (DHE) inversion recovery sequence (134) 10 to 30 minutes following an intravenous injection of a contrast agent (Gd-DTPA, Berlex Magnevist) at 0.25 mmol/kg (BW \pm 140 Hz/pixel, TE/TR 3.9/8 ms, turbo flash with readout flip angle 30°, FOV 119 x 200 mm, image matrix 122 x 256, spatial resolution 1.0 x 0.8 x 3.0 mm, readout at late diastole, 13 views per segment, 20–25 slices without a gap). For these two sequences, LV short-axis image slices from the LV apex to base were acquired in a consistent manner based on predetermined anatomical landmarks without gaps between slices. Each image acquisition was ECG-gated, and the image was acquired during a single breath hold (30–40 sec) by manually holding mechanical ventilation at end expiration. Three-dimensional (3D) displacement fields were calculated measured in five short axis slices using a motion tracking sequence (DENSE, BW \pm 1000 Hz/pixel, TE/TR 1.55/3.1 ms, 15° readout flip angle FOV 175 x 350 mm, image matrix 128 x

256, spatial resolution 1.36 x 1.36 x 8.0 mm) (5). To spatially register the displacement fields with the MI geometry, encoding and readout of the DENSE sequence was set at end systole and end diastole, respectively. The timing of end systole and end diastole was determined from the cine images.

The epicardial sock electrical recording system in the MR scanner room was described previously (88). Briefly, all 247 electrodes were radiofrequency-filtered at the MR scanner interface, and the data from the epicardial electrodes and physiological monitoring, including LV pressure and ECG, were simultaneously acquired at a minimum sampling rate of 1,000Hz for a duration of 10 sec immediately prior to and following the MR scans (29). Animals were euthanized and their hearts were scanned with a 3D spin-echo sequence to locate the glass bead markers (BW \pm 130 Hz/pixel, TE/TR 12/148 ms, FOV 256 x 256 x 88 mm, image matrix 256 x 256 x 88, spatial resolution 1.0 x 1.0 x 1.0 mm). After excision, the heart was filled with vinyl polysiloxane, maintaining end diastolic shape, and the locations of electrodes and beads were digitized (Microscribe 3DLX, Immersion Corporation, San Jose, CA) (88, 148).

8.3.3 Data analysis

The LV myocardium was manually segmented in the DENSE images. The 3D end-diastolic configuration and 3D displacement of the tissue in each pixel were measured (4). The end-systolic configuration was calculated from the displacement vector field and the end-diastolic configuration. The Lagrangian Green's strain tensor \mathbf{E} was calculated in every tetrahedron of adjacent 4 pixels in the LV myocardium as $0.5(\mathbf{F}^T\mathbf{F}-\mathbf{I})$, where \mathbf{F} is the deformation gradient tensor, \mathbf{F}^T is the transpose of \mathbf{F} , and \mathbf{I} is the identity matrix. The reference and deformed states were defined as end-diastolic and end-systolic configurations, respectively. Three independent finite strains (E_{rr} , E_{cc} , and E_{ll}) were computed in the local cardiac coordinate system with reference to the center of mass of each short-axis LV myocardium image (172). E_{rr} , E_{cc} , and E_{ll} reflect myocardial stretch or shortening along the radial (E_{rr}), circumferential (E_{cc}) and longitudinal (E_{ll}) cardiac axes, respectively. The LV circumference was divided into 32 sectors and two layers (endocardium and epicardium), and average strain values were calculated in each sector.

In the DHE images, MI geometry and endocardial and epicardial borders were manually segmented (164). As in the strain map, the LV circumference was divided into 32 sectors, and the average transmural depth of MI was calculated in each sector. The values were linearly interpolated to determine the MI depth at the DENSE image slice positions.

Electrical signals from the 247-lead sock electrodes were averaged over approximately 20 consecutive heartbeats. The local depolarization time at each electrode was defined as the peak negative dV/dt within the QRS complex (140). The time reference for the local depolarization times was the earliest ventricular depolarization time in each heart. Delayed electrical activation was defined as electrical activation time exceeding the mean +2SD of activation times of all the nodes. Electrical maps were spatially referenced to the strain and MI geometry maps using the least squares fit of glass marker beads (88, 148). The electrical activation time in each of the 32 sectors was calculated as the weighted average of the electrical activation time at three adjacent electrodes (Figure 8-1). In essence, the electrical activation at 160 (= 32 sectors x 5 slices) nodes was calculated from those at the 247 original sock electrodes. A total of 160 nodes that contain mechanical, infarct extent, and electrical data were measured.

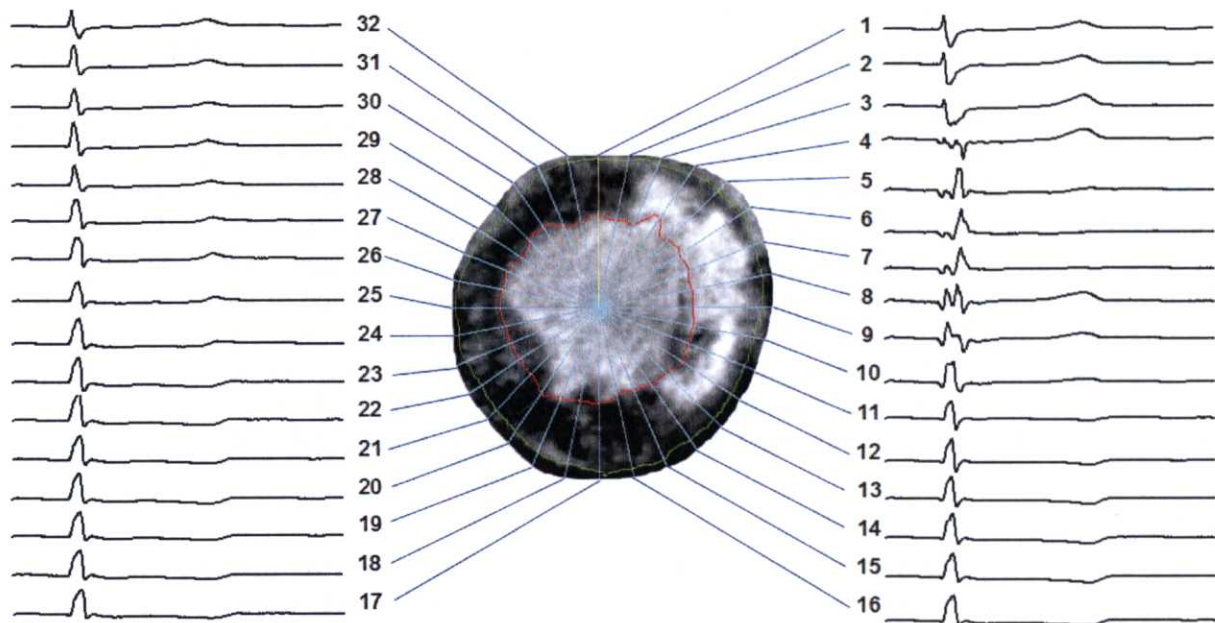


Figure 8-1: Epicardial electrical signals during a cardiac cycle (cycle length = 544 msec, HR = 110 bpm) registered to a short-axis delayed hyperenhancement (DHE) MR image of the LV split into 32 sectors. Regions of bright intensity correspond to MI. In this image, abnormal or delayed activation is seen from sectors 4 through 9, whereas the infarct ranges from sectors 3 through 14.

At the mid-LV level, each sector corresponded to an approximate 3-D dimension of 6 mm (circumference), 6 mm (longitudinal), and 10 mm (radial). The infarct zone was defined as the sectors including MI. The border zone was defined as the sectors immediately adjacent to the infarct zone, and the remote zone was defined as the sectors opposite to the infarct zone in the short-axis images. Division of the LV circumference into 32 sectors was chosen to create nodes with sufficiently high spatial resolution that would robustly detect the depressed function at the border zone, the width of which is reported to be <10 mm (93, 94, 156, 198, 210, 238).

8.3.4 Statistical analysis

Values are means \pm SD unless otherwise specified. A paired t-test was used to compare electrical and mechanical parameters. Statistics were performed using SigmaStat 3.0 (SPSS, Inc. Chicago, IL).

8.4 Results

All the animals went into intermittent ventricular tachycardia 24 hours after reperfusion, which lasted for 24-48 hours, and normal sinus rhythm resumed thereafter. The animals underwent the electromechanical data acquisition 38 ± 12 days post-MI. Hemodynamic parameters at the time of the electromechanical data acquisition are summarized in Table 8-1.

8.4.1 Infarct map

The infarct was located in the antero-septal region, and usually involved the anterolateral papillary muscle on the endocardial border (Figure 8-2). The epicardial border exhibited spatially intricate structures with multiple interdigitations of viable myocardium within the infarct zone.

The average depth of the infarct was $55 \pm 11\%$ and the infarct covered $28 \pm 6\%$ of the LV mass (Figure 8-3A).

Heart rate (bpm)	121 ± 19
LVP _{max} (mmHg)	85 ± 16
dP/dt _{max} (mmHg/s)	1713 ± 661
dP/dt _{min} (mmHg/s)	-1515 ± 643
LVEDP (mmHg)	7 ± 3
LVEDV (mL)	40 ± 6
LVESV (mL)	23 ± 5
LVSV (mL)	17 ± 2
LVEF (%)	43 ± 3
CO (mL/min)	2058 ± 517

Table 8-1: Hemodynamic data. Values are mean \pm SE. LVP_{max}: peak LV pressure, dP/dt_{max}: peak positive dP/dt, dP/dt_{min}: peak negative dP/dt, LVEDP: LV end-diastolic pressure, LVESV: LV end-systolic pressure, LVSV: LV stroke volume, LVEF: LV ejection fraction, CO: cardiac output.

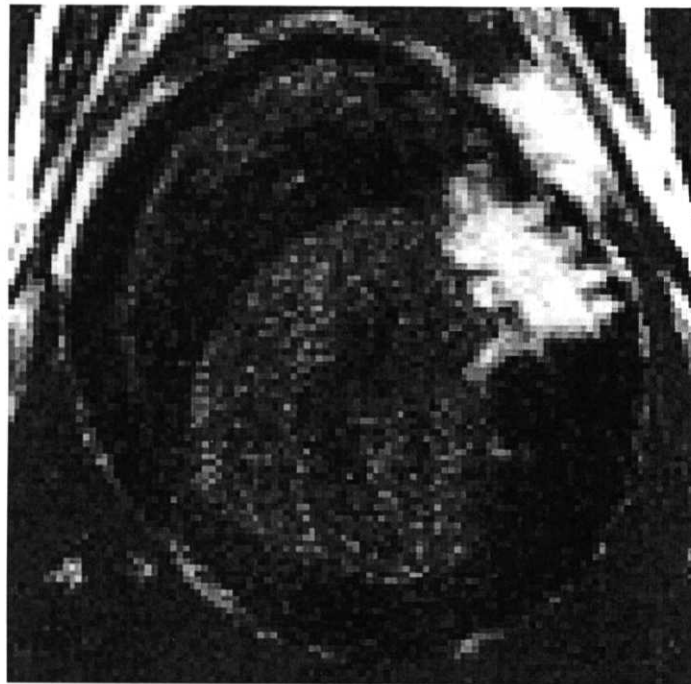


Figure 8-2: Short-axis, delayed hyperenhancement (DHE) image of MI from a single animal. Regions of bright intensity correspond to MI. Note multiple interdigitations of viable myocardium within the infarct region.

8.4.2 Electrical activation map

Electrical breakthrough, or the point of the earliest activation, was located in the anterior RV region. Activation time exceeding the mean + 2SD was observed in a total of 19 nodes, and these nodes were within the infarct zone (Figure 8-3B). The electrical activation time was significantly delayed in the infarct zone compared with that of the border zone (31 ± 9 vs. 19 ± 2 msec, $P < 0.05$, Figure 8-4), whereas the electrical activation time was not significantly different between the border zone and the remote zone (19 ± 2 vs. 21 ± 3 msec, $P = \text{n.s.}$).

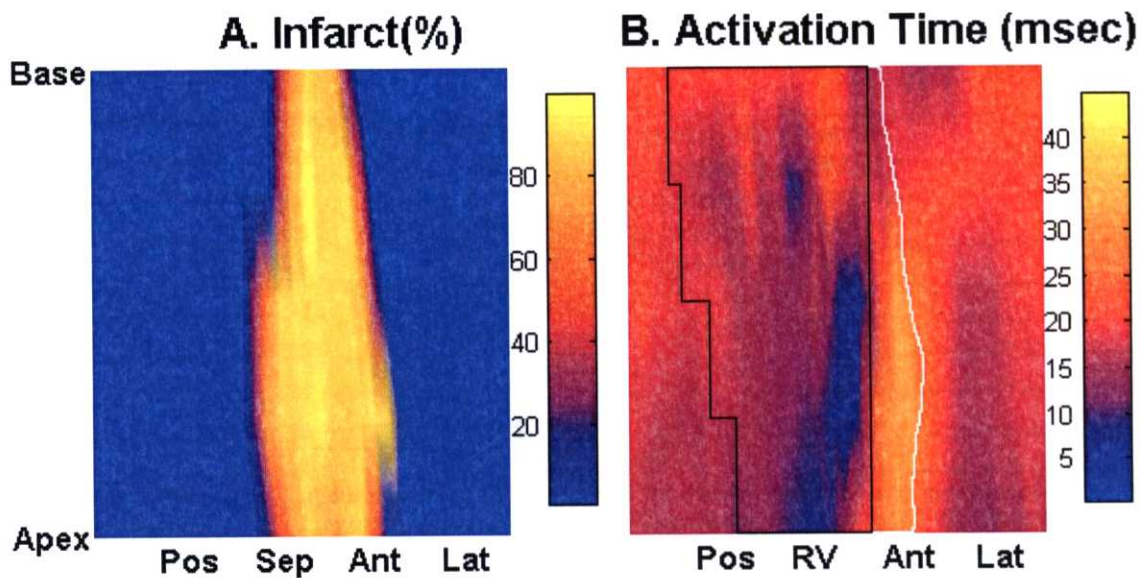


Figure 8-3: Infarct depth and electrical activation from a single animal. A. Infarct map (in percent depth). B. Isochrone map of electrical activation time (in msec). The area circumscribed by a solid black line represents the right ventricle (RV). The area circumscribed by a solid white line represents the infarct zone in the anterior wall (the infarct zone in the septum is covered by the RV). Pos: posterior wall, Sep: septal wall, Ant: anterior wall, Lat: lateral wall.

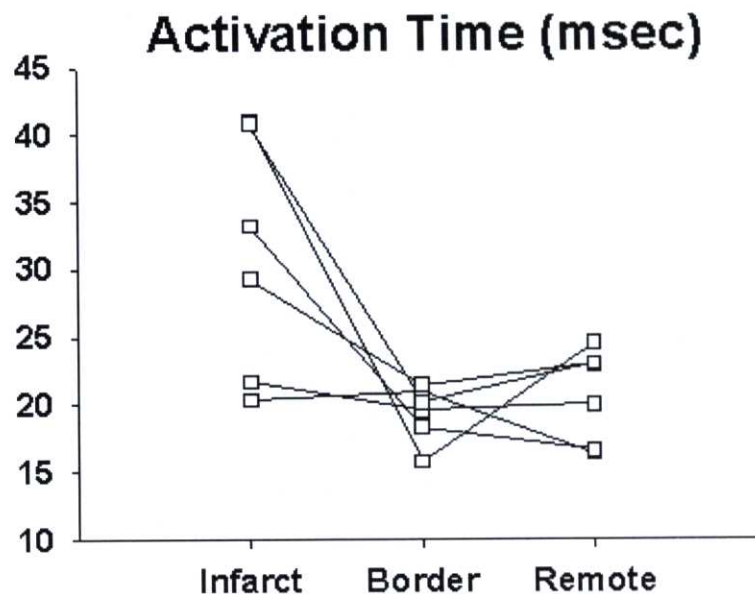


Figure 8-4: Quantitative analysis of electrical activation time in each zone (n=6). Electrical activation time (in msec) was significantly delayed in the infarct zone compared with that of the border zone (31 ± 9 vs. 19 ± 2 msec, $P < 0.05$), whereas it was not significantly different between the border zone and the remote zone (19 ± 2 vs. 21 ± 3 msec, $P = \text{n.s.}$).

8.4.3 Strain map

A 3D displacement map from a single animal is shown in Figure 8-5. Each arrow represents a displacement vector that points from the end-diastolic to end-systolic configuration. The magnitude of displacement is color-coded. It is clear that the displacement magnitude in the infarct zone in the anteroseptal wall (left side of the figure) is small (purple ~ blue) compared to that of the remote zone in the posterolateral wall (right side of the figure, red ~ yellow).

The number of strain calculation points was $3,625 \pm 766$ per heart, and each sector contained 11 ± 3 strain points from which average strains were calculated. E_{rr} , E_{cc} and E_{ll} over the infarct zones were smaller in magnitude than those in the remote zones (Figure 8-6). The regions of abnormal strains, particularly the longitudinal strain, extended far beyond the infarct zone, and the strains were larger in the epicardium than in the endocardium.

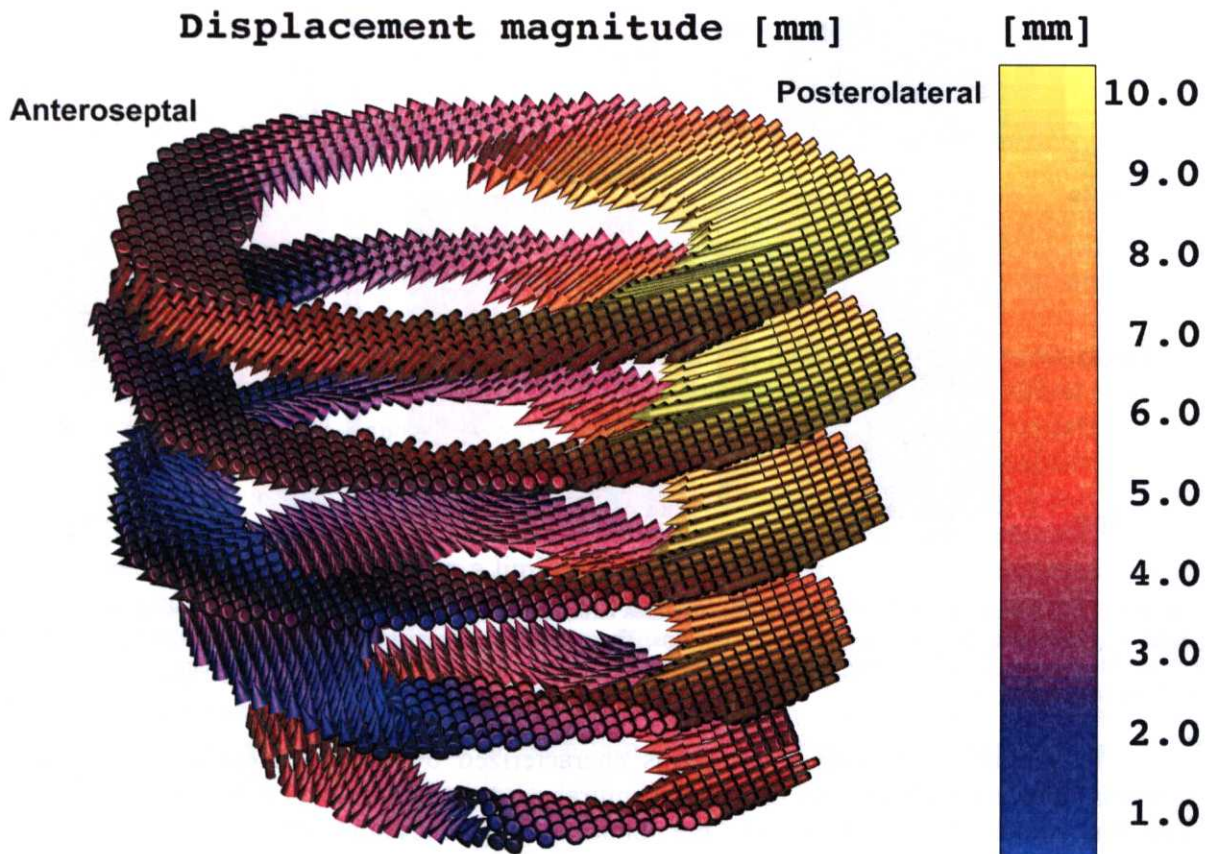


Figure 8-5: 3-D Displacement map from a single animal. Each arrow represents a displacement vector that points from the end-diastolic to end-systolic configuration. The magnitude of displacement is color-coded. It is clear that the displacement magnitude in the infarct zone in the anteroseptal wall (left side of the figure) is small (purple ~ blue) compared to that of the remote zone in the posterolateral wall (right side of the figure, red ~ yellow).

E_{rr} , E_{cc} and E_{ll} in both the infarct zone and the border zone were significantly smaller than those in the remote zone ($P < 0.05$), and there was no significant difference between the infarct zone and the border zone ($P = \text{n.s.}$, Figure 8-7). However, E_{rr} , E_{cc} and E_{ll} in the border zone were more heterogeneous than those in the infarct and remote zones. For example, two animals showed positive circumferential strains in the border zone, which indicate that the myocardium in the border zone underwent a paradoxical systolic stretch in the circumferential direction. Similarly, three animals showed large positive longitudinal strains in the border zone, which suggest that the border myocardium was also stretched in the longitudinal direction during active contraction (Figure 8-7). There was a significant transmural gradient between the epicardium and endocardium in E_{rr} , E_{cc} and E_{ll} in the remote zone ($P < 0.05$), and the transmural gradient was lost in both the infarct zone and the border zone.

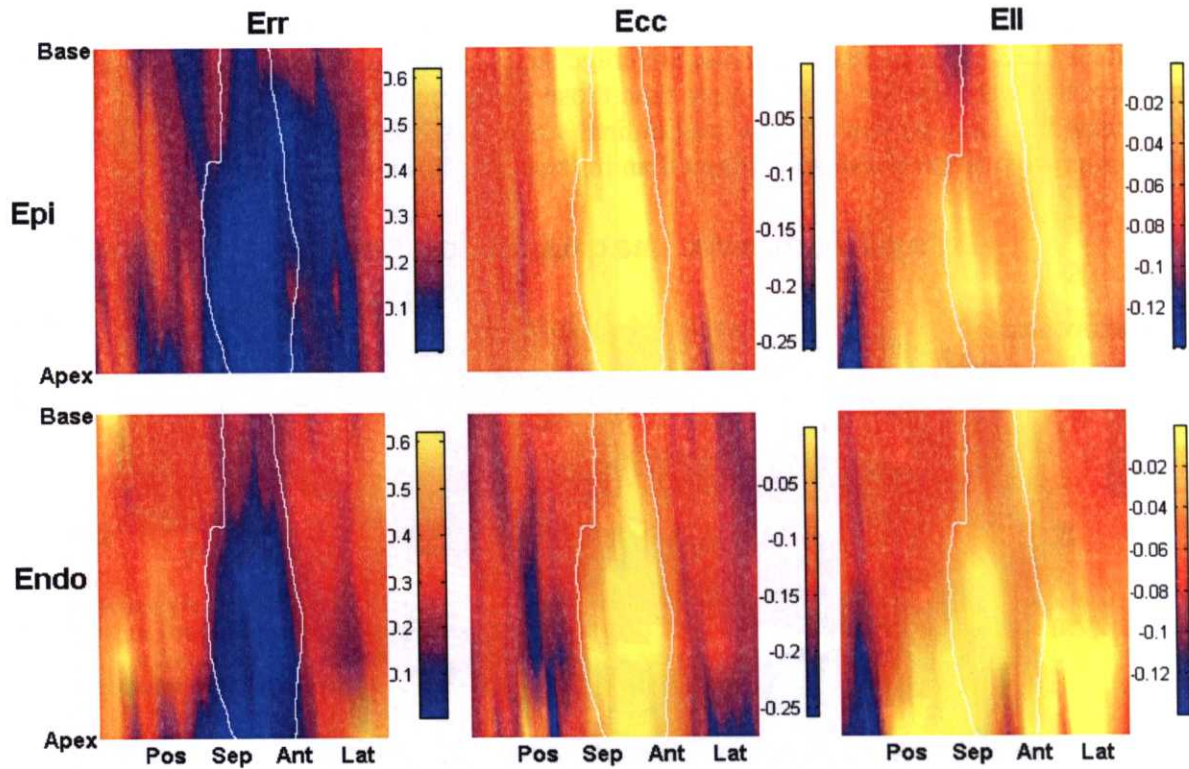


Figure 8-6: 3-D Strain map from a single animal. The area circumscribed by a solid white line represents the infarct zone. E_{rr} : radial, E_{cc} : circumferential, and E_{II} : longitudinal strains. The regions of abnormal strains, particularly the longitudinal strain, extended far beyond the infarct zone, and the strains were larger in the epicardium than in the endocardium. Epi: epicardial layer, Endo: endocardial layer.

In summary, the infarct zone was characterized by delayed electrical activation and abnormal mechanics, including loss of transmural gradient and reduction of strain magnitude or abnormal stretch. In the border zone, the abnormal mechanics similar to that of the infarct zone was observed, however, the electrical activation time was not different from that of the remote zone.

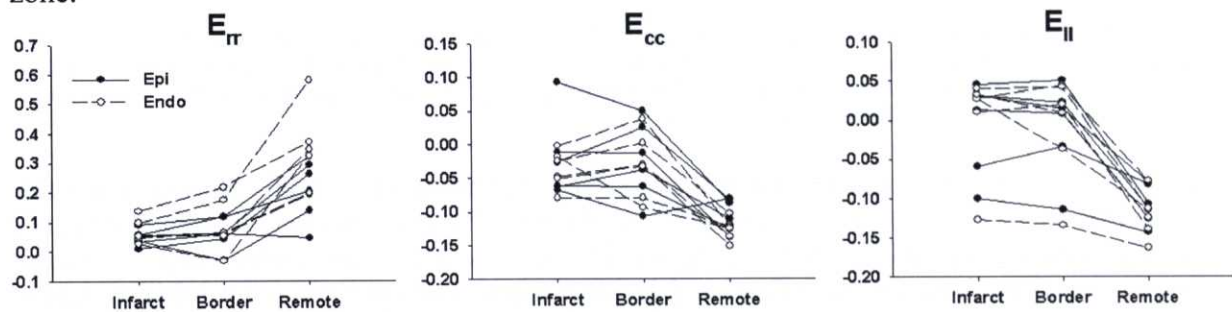


Figure 8-7: Quantitative analysis of finite strain in each zone ($n=6$). E_{rr} , E_{cc} and E_{II} in both the infarct zone and the border zone were significantly smaller than those in the remote zone ($P<0.05$), and there was no significant difference between the infarct zone and the border zone ($P=n.s.$). There was a significant transmural gradient between the epicardium and endocardium in E_{rr} , E_{cc} and E_{II} in the remote zone ($P<0.05$), and the transmural gradient was lost in both the infarct zone and the border zone.

8.5 Discussion

To examine the electrical activation in the infarct border zone with depressed systolic function in chronic MI, the present study combined epicardial electrical recording and high-resolution MR-based imaging techniques to examine the anatomical correlation of local depolarization, myocardial scar, and systolic deformation of the heart.

8.5.1 Impact of infarct geometry on electromechanical properties in the infarct zone

Recent studies have demonstrated that the DHE MRI technique allows accurate assessment of infarct extent and geometry. The spatial extent of DHE was the same as that of myocyte necrosis shown by TTC-stained pathology, independent of wall motion and infarct age (138), and the clinical reproducibility of DHE for determination of infarct size and distribution is highly comparable to that of routine clinical SPECT (160). In addition, DHE is superior to SPECT in detection of subendocardial infarcts (243).

The high-resolution DHE MRI results show that the infarct geometry on the border is highly complex (Figure 8-2). The infarct was mostly non-transmural, and the infarct size was approximately 30% of the LV, both observations are consistent with previous reports using a similar occlusion-reperfusion infarct model in canine (78, 95, 96, 129, 248). Traditionally, the infarct zone has been associated with delayed electrical activation and slow conduction velocity (69, 78, 97, 235). As a result of the relatively small size and nontransmural geometry of the infarct, delayed electrical activation in a limited region of the infarct zone was observed (Figure 8-3). The nontransmural geometry of the infarct in this model is not only due to reperfusion, but also due to rich collateral circulation intrinsic to the canine heart, as compared to humans and other species such as swine. Permanent coronary ligation in swine tends to result in a well-demarcated, transmural scar, whereas the canine MI model in the present study is characterized by a relatively large volume of viable myocardium over the infarct zone. Holmes *et al.* (115) studied systolic deformation of the infarct zone in a small region in the LV free wall by permanent occlusion of obtuse marginal branches of LCx in swine. Despite impaired systolic shortening, they found significant systolic wall thickening (E_{π}) still present at 3 weeks, which is suggestive of passive deformation in the regions composed almost entirely of collagen (114). In contrast, the results indicate significant impairment of systolic deformation and loss of transmural gradient in E_{π} , E_{cc} and E_{ll} including systolic wall thickening in the infarct zone (Figure 8-4A and 8-5B). Therefore, the difference in the infarct geometry appears to account for distinct characteristics in mechanics as well as electrical activation.

8.5.2 Mechanism of impaired systolic function in the infarct border zone

It was hypothesized that impaired systolic function in the infarct border zone may be partially accounted for by delayed electrical activation. However, the electrical activation in the infarct border zone was not delayed compared with that of the remote zone (Figure 8-3 and 8-4), whereas the systolic function of the border zone was depressed relative to that of the remote zone (Figure 8-6 and 8-7). These findings indicate that electrical factors do not contribute to the impaired systolic function in the border zone, but the border zone dysfunction most likely results from mechanical interaction between the ischemic and normal myocardium. Although the details of this interaction are not fully understood, mechanical ‘tethering’ due to high wall stress appears to be a predominant mechanism (93, 94, 156, 163, 210, 238). Besides, recent studies have demonstrated that the LV myocardial shortening is regionally heterogeneous (271), and the timing and the peak of shortening may be controlled by nonuniform prestretch from atrial

contraction via a regional Frank-Starling effect (272). Although prestretch during atrial contraction in the border zone was not quantified because the time course of finite strain was not examined, it is speculated that a higher wall stress in the border zone may lead to a smaller diastolic prestretch, which may contribute to generating delayed and little shortening.

Of note, the high-resolution motion tracking MRI revealed a paradoxical systolic stretch (eccentric contraction) in the border zone of some animals (Figure 8-7). This systolic stretch is clinically important because it may generate a stretch-activated ectopy to trigger reentry ventricular arrhythmia in patients with ischemic heart. Whether the border zone undergoes little shortening (isometric contraction) or stretch is most likely determined by the loading conditions, the local material property during ventricular tension development, and the ventricular geometry which dynamically changes as structural remodeling progresses. Anatomy-based description of mechanics in the myofiber direction may identify the geometrical and hemodynamic factors that contribute to systolic stretch in the border zone.

8.5.3 Limitations

In the present study, the 3D finite deformation of the LV wall was examined in open-chest, anesthetized dogs. Therefore, the results may not precisely reflect the cardiac mechanics in closed-chest, awake animals. The spatial registration error in this electromechanical mapping technique has been previously reported to be 2.1mm on average, with a precision of marker localization in the images and on the excised heart of 1.0 and 0.7mm, respectively (88). The temporal resolution of electrical activation was 1ms, and this means that the temporal error was within approximately one sample point (≈ 1 ms). Although this mapping technique assumes that the heart be undeformed between the *in situ* and excised states without controlled perfusion fixation *in situ*, previous studies have reported high precision in registering electrode locations over the epicardial surface (88). Due to MR compatibility issues, the electrical mapping in this study was limited to the epicardium. At present, commercially available basket catheters contain ferromagnetic material and would severely degrade image quality. Clinical applicability of this technique to the study of cardiac electromechanics may be limited by surgical intervention and mechanical restriction from the sock. Due to a highly elastic property of the sock material, its mechanical restriction on the heart was minimal, although this effect has not yet been quantified.

8.6 Conclusions

Using a high-resolution electromechanical mapping system, it was demonstrated that abnormal electrical activation is observed in a limited region in the infarct zone, whereas the region of abnormal mechanics extends far beyond the infarct zone in chronic MI. The infarct border zone is characterized by abnormal mechanics directly coupled with normal electrical depolarization. These findings indicate that electrical factors do not contribute to the impaired systolic function in the border zone, but the border zone dysfunction most likely results from mechanical interaction between the ischemic and normal myocardium.

Chapter 9

Global electromechanics in infarct heart with substrate for sudden cardiac death

9.1 Abstract

Alteration of action potential duration (APD) induced by abnormal myocardial stretch (“prestretch”) may increase the risk for sudden cardiac death (SCD) in chronic myocardial infarction (MI) by increasing the dispersion of repolarization and thus susceptibility to ventricular tachyarrhythmia. However, the effect of prestretch on APD in chronic MI *in vivo* has not been quantitatively characterized. The anatomical correlation between prestretch and activation recovery intervals (ARI) was examined in a swine model of chronic MI *in vivo* where sustained monomorphic ventricular tachycardia is inducible with programmed stimulation. Four weeks after antero-septal MI was created (n=8), high temporal resolution (9msec) tissue-tracking MRI was used to characterize prestretch, and a 247-lead epicardial sock was subsequently placed over the ventricular epicardium to measure ARI under thoracotomy. Prestretch appeared in early systole (30msec), peaked at 70-100msec and persisted until 240-260msec. Prestretch was highest in the posterolateral viable region opposite to the MI ($10.8 \pm 7.4\%$), and lowest in the septal region ($0.4 \pm 0.2\%$) ($P < 0.05$). Prestretch significantly correlated with peak shortening rate ($P < 0.001$). The magnitude of prestretch was similar in the viable and infarct myocardium, but the mechanical function in the infarct myocardium was significantly depressed. There was a strong and consistent correlation between prestretch and ARI in all animals ($P < 0.001$). In conclusion, prestretch in chronic MI prolongs local ARI in a magnitude-dependent manner in swine with a substrate for SCD. Early interventions to reduce prestretch may help reduce SCD in patients with chronic MI.

9.2 Introduction

Recent prospective multicenter clinical trials demonstrate a major survival benefit with the implantable cardioverter-defibrillator (ICD) for sudden cardiac death (SCD) in high-risk subgroups with LV dysfunction due to prior myocardial infarction (MI) and nonischemic cardiomyopathy (149). However, these high-risk subgroups constitute only a small fraction of all the SCDs, thus interventions in these subgroups do not have a major impact on the general public health problem of SCD (183). The majority of the potential SCD victims are not protected by the current guidelines, primarily due to lack of specific markers with high predictive value (269).

SCD mainly results from ventricular tachyarrhythmia (30), thus is an electrical phenomenon. Inducibility of sustained monomorphic ventricular tachycardia (VT) during the invasive electrophysiological study with programmed electrical stimulation (PES) is clinically equivalent to having an arrhythmic substrate, thus has been used as a prognosticator for subsequent mortality (47). However, its prognostic accuracy is relatively low (47, 48, 70). Moreover, its invasive nature does not warrant its use in the “unprotected” yet low-risk subgroups.

Recent studies demonstrate that the mechanical and electrical phenomena in cardiomyocytes are interdependent (146). For example, mechanical myocardial stretch alters action potential duration (APD) (204). Because myocardial scar and structural remodeling confer significant heterogeneity to the material properties of the LV, the regions of greater compliance that experience abnormal stretch, or “prestretch”, may have altered APD (170), which may increase the dispersion of repolarization within the ventricles and thus the susceptibility to ventricular tachyarrhythmia (27, 91, 204). In fact, strategies to suppress prestretch in patients with LV dysfunction have consistently proven to be successful. For example, beta-blockers, which likely diminish prestretch due to the negative inotropic effect, reduce mortality among patients with heart failure due to a significant reduction in fatal arrhythmias (59). Furthermore,

cardiac resynchronization therapy (CRT) aims to cancel the prestretch in the lateral LV by electrically stimulating the region of prestretch, and improves morbidity and mortality in patients with intraventricular conduction delay (40, 58)

Stretch is a mechanical phenomenon, thus can be noninvasively quantified with MRI (194). Quantitative characterization of the effect of prestretch on APD may establish the link between the mechanical and electrical phenomena in the heart *in vivo*, and allow an MR-guided, noninvasive risk assessment of SCD in these “unprotected” subgroups. The hypothesis of this work is that the prestretch in chronic MI with a substrate for SCD alters the local APD *in vivo*. To test this hypothesis, the anatomical correlation between prestretch and activation recovery intervals (ARI) in a swine model of chronic MI was quantitatively examined where sustained monomorphic VT is inducible with PES. A high-resolution cardiac electromechanical mapping technique (88) allowed acquisition of anatomical, electrical, and mechanical data with high precision.

9.3 Materials and Methods

All studies were performed according to the *Position of the American Heart Association on Research Animal Use* (2). All animal protocols were reviewed and approved by the Animal Care and Use Committee of the National Heart, Lung and Blood Institute, which is accredited by the American Association for Accreditation of Laboratory Animal Care.

9.3.1 Creation of myocardial infarction

Eight domestic swine (34–47kg) were anesthetized with intravenous sodium thiopental (10-20mg/kg), intubated, and mechanically ventilated with a mixture of oxygen, medical air and isoflurane (1-3%). The surface ECG and the arterial pressure were recorded throughout the procedure. A modified version of a catheter-based MI procedure was used that has been described previously (213). Briefly, a balloon angioplasty catheter (2.7Fr) was advanced to the middle portion of LAD through a 6Fr guiding catheter via a carotid artery. The balloon was then inflated to 6atm to occlude LAD, and the occlusion was maintained for 150min. After completion of balloon occlusion, the catheters were removed, the surgical wound closed, and the animals extubated and recovered.

9.3.2 *In vivo* MRI

Four weeks after MI, the animals were anesthetized as above to undergo *in vivo* MRI in a 1.5T scanner (Espree, Siemens) with an 8-channel surface coil (Figure 9-1B). LV geometry (end-diastolic and end-systolic volumes) and global function (stroke volume and ejection fraction) were evaluated with a cine steady state free precession (SSFP) sequence (bandwidth (BW) $\pm 930\text{Hz/pixel}$, echo time (TE) 1.78msec, repetition time (TR) 3.56msec, readout flip angle 35° , FOV $233 \times 311\text{mm}$, image matrix 192×256 , spatial resolution $1.2 \times 1.2 \times 8.0\text{mm}$, 6 slices). LV regional function was evaluated with a 2-D myocardial tissue tracking sequence (137) (cine-DENSE, BW $\pm 1,008\text{Hz/pixel}$, TE/TR 4.52/9.0msec, readout flip angle $12\text{--}20^\circ$, FOV $400 \times 200\text{mm}$, image matrix 128×64 , spatial resolution $3.1 \times 3.1 \times 8.0\text{mm}$, 6 slices). The slice positions and the orientations of the cine-DENSE images were set to the same as those of the cine SSFP images to facilitate registration processes. A total of 45-65 images were acquired through systole with temporal resolution 9.0msec. LV short-axis image slices from LV base to apex were acquired in a consistent manner based on predetermined anatomical landmarks without gaps between slices. Each image acquisition was ECG-gated, and the image was acquired during a single breath hold (40–60sec) by manually holding mechanical ventilation at end-expiration.

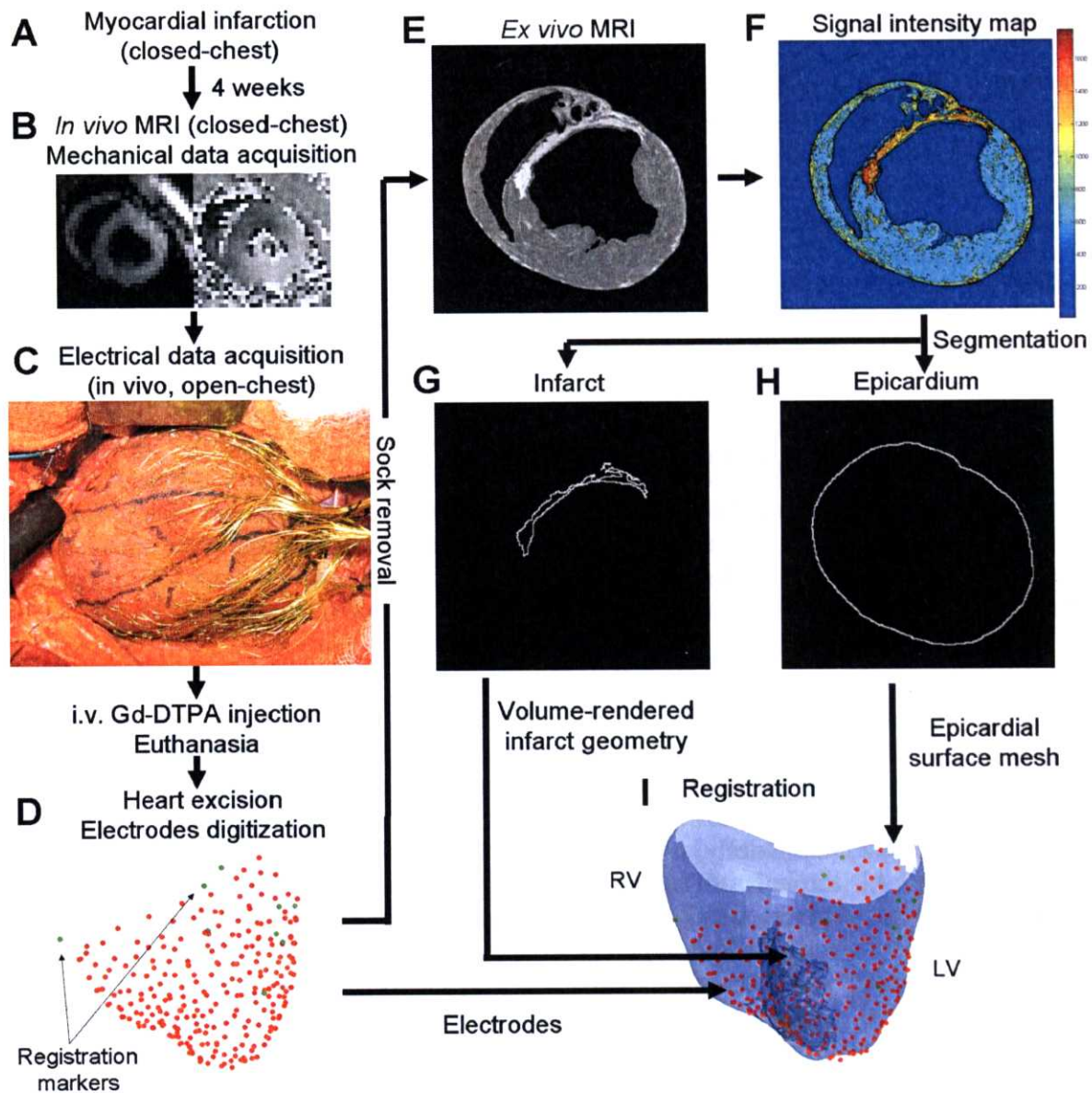


Figure 9-1: Overview of data acquisition and processing.

9.3.3 Electrical recording system

After MRI, the animals were transported to the operating room and underwent median sternotomy. A multi-electrode epicardial sock, consisting of a nylon mesh fitted with 247 silver electrodes (electrode spacing = 2-5mm) attached in an ordered fashion, was placed over the ventricular epicardium (Figure 9-1C) (19). The sock was placed in a consistent and predetermined orientation and secured with several sutures. A ground reference electrode was sewn onto the fat pad at the aortic root. Electrical signals from the 247 electrodes were amplified (gain=1,000) and were acquired using a 16-bit analog-to-digital converter (PXI-6225x4, National Instruments) at a minimum sampling rate of 1,000Hz for 10sec (29). Electrical recording was conducted during normal sinus rhythm and sustained monomorphic ventricular tachycardia (VT).

9.3.4 Induction of monomorphic ventricular tachycardia

A pair of bipolar stimulating wires was directly attached to the RV outflow tract (RVOT) using an alligator clip, and a modified version of a swine PES protocol (213) was conducted to test the inducibility of sustained monomorphic VT. Briefly, following regular ventricular pacing (S1) at a cycle length of 250, 300 and 350 msec, single (S2), double (S2, S3), or triple (S2, S3, S4) premature stimuli were introduced with a pulse width of 2msec. The endpoints of the protocol were (1) completion of the protocol without inducing VT or ventricular fibrillation (VF); or (2) induction of two episodes of sustained monomorphic VT or VF. Sustained VT was defined as monomorphic VT lasting more than 15sec, and was confirmed in multiple leads. VT or VF was electrically defibrillated using an internal defibrillation pads, and the cables to the electrical recording system were temporarily disconnected before defibrillation to protect the system. When either endpoint was reached, the stimulating protocol was completed.

9.3.5 Postmortem studies

Upon completion of PES, heparin 5,000IU and Gd-DTPA 0.20mmol/kg were administered intravenously and the animals euthanized 15-20min after Gd-DTPA. The heart was removed, and was filled with vinyl polysiloxane, maintaining end-diastolic shape (88). As markers for registering the MR and the sock data, eight to fifteen 10mm x 1mm glass tubes filled with Gd-DTPA (5 mM) were placed in the myocardium, and the locations of the sock electrodes and the glass tubes were digitized (Microscribe 3DLX, Immersion, San Jose, CA) (Figure 9-1D) (88). The sock electrodes were removed from the heart. To evaluate the extent of MI with high spatial resolution, ex vivo contrast-enhanced MR imaging(109) was performed in the 1.5-T MR scanner. A 3D Gradient recalled echo (GRE) sequence was used to visualize MI and to locate the glass tube markers (BW ± 170 Hz/pixel, flip angle 20°, TE/TR 3.71/9.28msec, FOV 120x120x100mm, image matrix 256x256x192, spatial resolution 0.5mmx0.5mmx0.5mm) (Figure 10-1E).

9.3.6 Data processing

Custom programs in MATLAB (Mathworks, Inc.) and C⁺⁺ were used for data analysis. The infarct geometry was extracted from the 3D GRE images using a signal intensity threshold to visualize a volumetric image of infarct in the ventricles (Figure 9-1F) (18). The MI was segmented as the hyperenhanced region that was seen in >1slice in the 3D GRE images. The hyperenhanced region was defined as >6 standard deviation (SD) of the signal intensity above the mean intensity of the circular region (diameter ~10mm) in the remote, viable myocardium (Figure 9-1G) (186). The values of the MI depth at the cine-DENSE image slice positions were determined using bicubic interpolation (19). The ventricular epicardial surface was extracted from the MR images, and the spherical harmonic mesh (117) was created to subsequently visualize electrical data (Figure 9-1H). The locations of the glass tube markers were determined from the MR images, and the electrodes were spatially referenced to the MR images using rigid-body transformation (Figure 9-1I) (88).

The local activation recovery interval (ARI) was measured between times of electrical activation ($=dV/dt_{\min}$ of the QRS) and recovery. Electrical recovery was defined as the dV/dt_{\max} of the T wave for negative T wave, the dV/dt_{\min} of the T wave for positive T wave, and at the mean time between dV/dt_{\max} and dV/dt_{\min} for biphasic T wave (259). ARI between electrodes was calculated using bicubic interpolation.

The LV endocardial border was manually segmented in the LV short-axis cine SSFP images to calculate global LV function using MIPAV (NIH) (164). The cine-DENSE sequence

used for tissue-tracking encodes one-dimensional (1-D) displacement of each pixel from the reference time point as stimulated echoes in the phase image (Figure 9-2) (4).

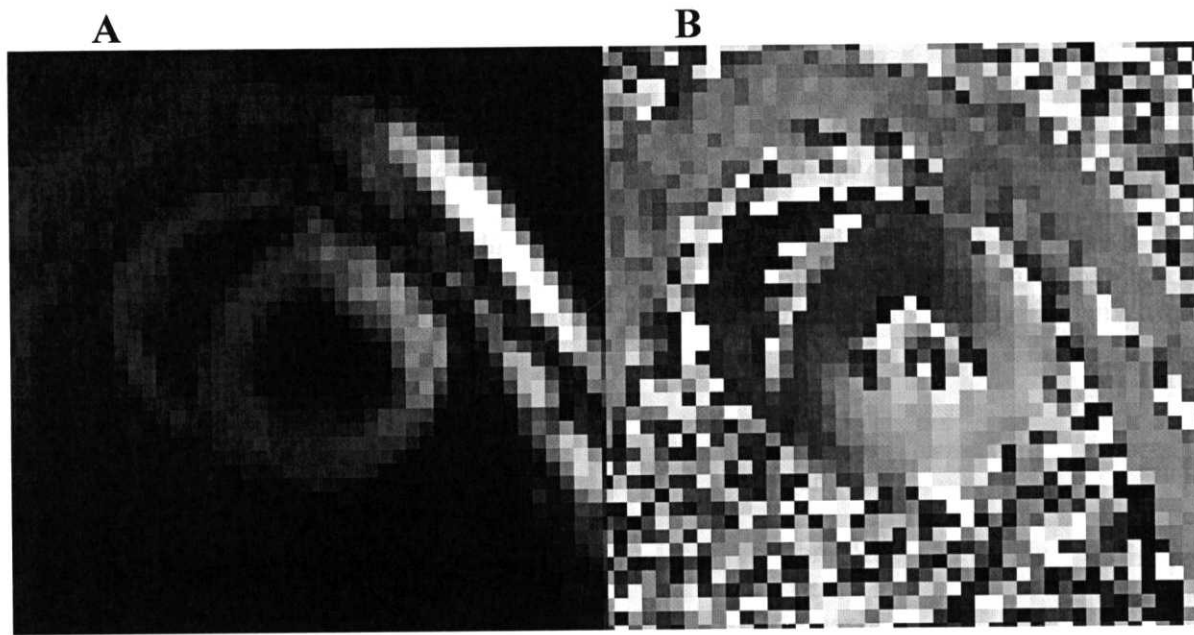


Figure 9-2: The magnitude (A) and the phase (B) images from the cine-DENSE sequence.

A previously published protocol of image acquisition, displacement and strain calculation was followed (137). Briefly, two separate image acquisitions were performed consecutively to obtain 1-D displacement information in two orthogonal directions (Figure 9-3).

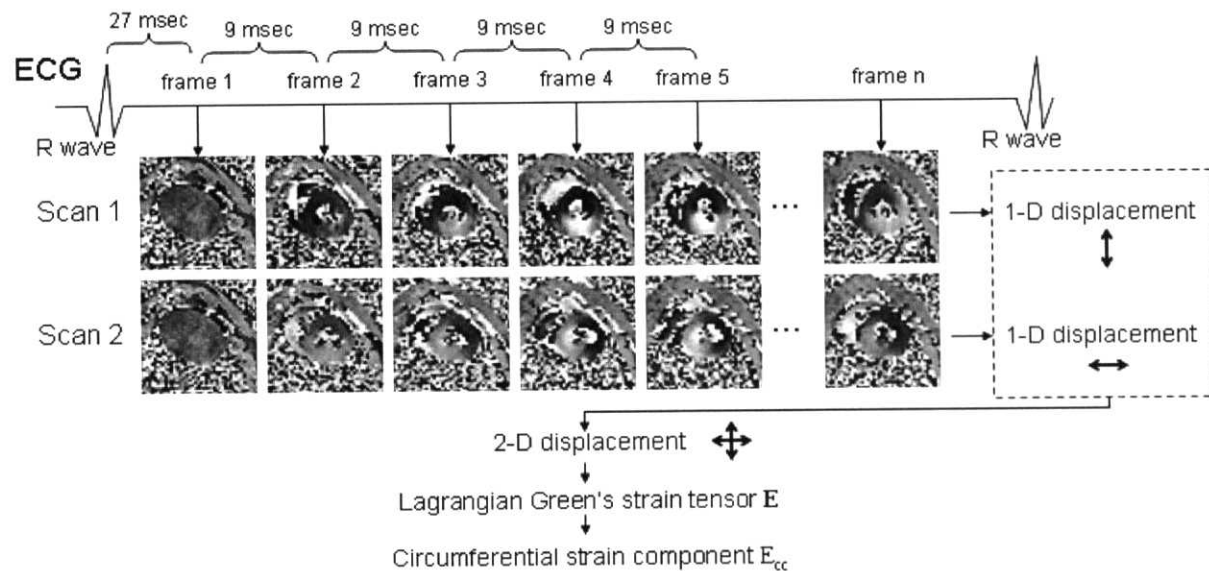


Figure 9-3: Image acquisition, displacement and strain calculation using the cine-DENSE sequence.

The LV myocardium was manually segmented and the 2-D phase information in each pixel was unwrapped and was converted to 1-D displacement. The 2-D displacement of each pixel was computed by means of vector addition of the two orthogonal 1-D displacement. Regional Lagrangian Green's strain tensor \mathbf{E} was computed by means of isoparametric

formulation with quadrilateral elements (175). Each quadrilateral element was a square 4-pixel neighborhood of myocardium where the final position of each element of myocardium is known (the pixel location), and the initial 2D position of each pixel is measured. After diagonalization of \mathbf{E} , the directions of the first and second principal strains and the corresponding eigenvalues were found. Relative to the center of mass of the left ventricle, the circumferential strain component E_{cc} were computed by means of projection of \mathbf{E} into the circumferential directions.

The circumferential strain ϵ_c (Figure 9-4A) was calculated from the Lagrangian Green's strain tensor as the percent change in length of a small line segment in the circumferential direction (272):

$$\epsilon_c = (\sqrt{1 + 2E_{cc}} - 1) \times 100\%$$

where E_{cc} is the Lagrangian Green's strain component in the circumferential direction. By convention in continuum mechanics, a negative strain represents shortening (e.g. -15%), whereas a positive strain represents stretch (e.g. +15%) in the circumferential direction.

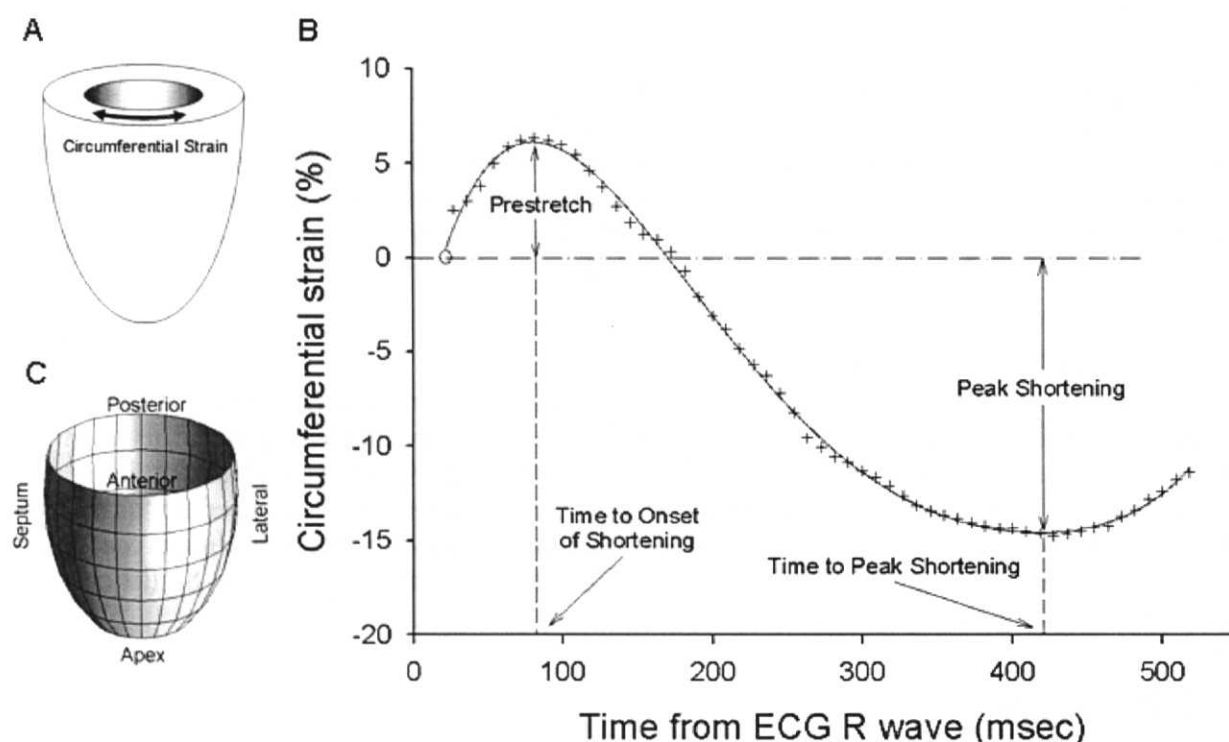


Figure 9-4: *A. Circumferential strain.* Circumferential strain ϵ_c describes myocardial shortening or stretch in the circumferential direction. *B. Sample fit of raw circumferential strain data at imaging times (+) to 7th-order polynomials (solid line).* Prestretch was defined as the maximum strain (=stretch) prior to systolic shortening, and the onset of shortening was defined as the time of prestretch. Peak shortening was defined as the minimum strain (= shortening) before diastolic stretch. *C. Mesh reconstruction of LV.* Mesh vertices indicate locations of displacement and strain calculations. There are a total of 144 data points per heart (=24 sectors x 6 short-axis slices).

The short-axis LV circumference was divided into 24 sectors, and the average circumferential strain in each sector was calculated. The reference configuration (strain=0%) was defined at the time of the tag encoding gradient, which occurred at 22msec after the peak R wave, and the first cine-DENSE image took place at 27msec after the peak R wave. The strains were interpolated across time using a seventh-order polynomial fit to all of the data points (Figure 9-

4B). The seventh-order polynomial was chosen over other orders of polynomial fits and local cubic splines because it provided a good trade-off between accuracy of the fit and noise reduction of the data (252).

Abnormal stretch, or “prestretch”, was defined as the maximum strain (= stretch) prior to systolic shortening, and peak prestretch rate was defined as the positive peak of the time derivative of the circumferential strains during prestretch. The onset of shortening was defined as the time of prestretch, and peak shortening was defined as the minimum strain (=shortening) before diastolic stretch. Peak shortening rate was defined as the negative peak of the time derivative of the circumferential strains during shortening. Strain data in sectors where prestretch or peak shortening was not observed were excluded from analysis.

Each LV had 144 mechanical data points (= 24 sectors x 6 short-axis slices), and a 3-D prolate ellipsoid mesh described by Bovendeerd *et al* (38) was used to present the data (Figure 9-4C). A bicubic surface through existing data points were fitted to the mechanical data, where the value of an interpolated point is a combination of the values of the sixteen closest points. The circumferential length of each sector in the figures was shortened or lengthened based on ϵ_c to visually demonstrate the dynamic motion of the LV with reference to the undeformed configuration.

9.3.7 Electromechanical data registration

To evaluate the correlation between electrical and mechanical indices involved in prestretch, electrical data were analyzed in the regions of prestretch. The electrical data in the LV was spatially referenced to mechanical data based on pre-determined anatomical landmarks. A bicubic surface fit was used to calculate the electrical indices at the mechanical data points in the LV that was covered by the epicardial sock electrodes.

9.3.8 Statistical analysis

Values are means \pm SD (n=8). One-way analysis of variance (ANOVA) was used to compare maximum and minimum mechanical indices, and mechanical indices between the viable and infarct myocardium. If a statistically significant result was obtained, individual locations were compared by a two-tailed *t* test. Linear regression was used to assess the correlation between prestretch and other indices, and expressed with their correlation coefficient (*r*) and *P* value. A value of *P*<0.05 was considered statistically significant. Statistics were performed with SigmaStat 3.0 (SPSS, Chicago, IL).

9.4 Results

9.4.1 General parameters

All animals survived the infarct procedure, which resulted in anteroseptal MI (Figure 9-5). The animals underwent MR studies and VT induction 29 \pm 5days after MI. Sustained monomorphic VT was induced in all animals, which indicate that all animals had a substrate for ventricular tachycarrhythmia (Figure 9-6). All induced VTs were hemodynamically unstable, likely due to relatively short cycle length (CL) (133 \pm 22 msec). Hemodynamic parameters during the MRI study are summarized in Table 9-1. The time course of circumferential strain by tissue-tracking MRI covered 86 \pm 21% of cardiac cycle (Table 9-2). The heart rate during the open-chest electrical measurements was 129 \pm 13bpm, and this value was significantly larger than that of the *in vivo* MRI study (97 \pm 28bpm, *P*=0.019).

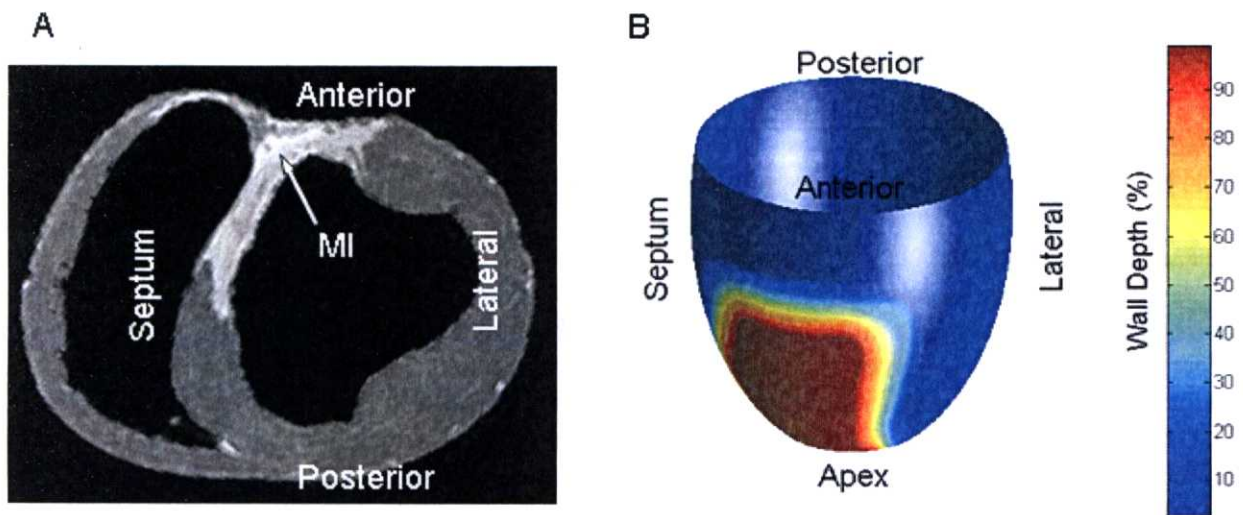


Figure 9-5: A. Short-axis view of the contrast-enhanced MR image. Regions of high intensity (= light gray) corresponds to MI. B. Mesh presentation of the MI wall depth (%) in one animal. Note the MI was mostly transmural (= dark red), and was located in the anteroseptal region.

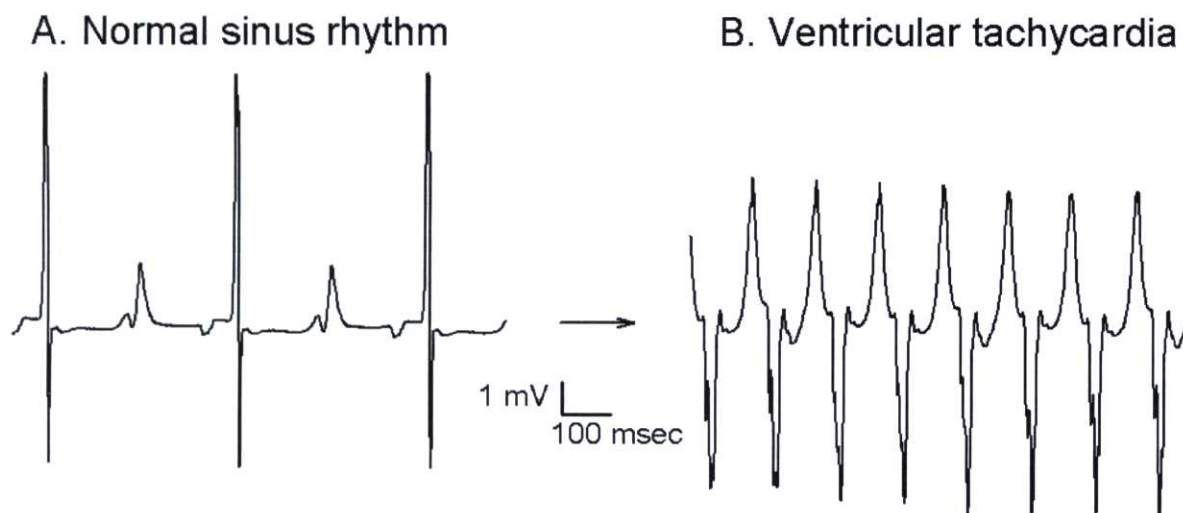


Figure 9-6: Unipolar electrical recordings from an epicardial sock electrode. A. Normal sinus rhythm, B. Sustained monomorphic VT.

Parameters	Mean \pm SD (n=8)
Heart rate (bpm)	96.8 \pm 28.4
LV end-diastolic volume (mL)	59.7 \pm 8.7
LV end-systolic volume (mL)	36.5 \pm 5.3
LV ejection fraction (%)	38.7 \pm 4.7
Stroke volume (mL)	23.2 \pm 4.7
Cardiac output (mL/min)	2,205 \pm 708.2

Table 9-1: Hemodynamic parameters at the time of MRI studies.

Parameters	Mean \pm SD (n=8)
Cycle length (msec)	669 \pm 198
Cine-DENSE coverage (msec)	546 \pm 92
Time coverage per cardiac cycle (%)	86 \pm 21

Table 9-2: Time coverage by cine-DENSE sequence per cardiac cycle.

9.4.2 Infarct geometry

The MI was mostly transmural due to little collateral circulation intrinsic to the swine heart, and was non-transmural only at its anterior and septal edges (Figure 9-5A). On the anterior edge, the MI usually involved the anterior segment of the anterolateral papillary muscle on the endocardial border. The MI also expanded to the anterior portion of the right ventricle (RV). The myocardial wall in all portions of the MI exhibited substantial thinning. The mesh presentation of the MI wall depth (%) in one animal is shown in Figure 10-5B. On average, the MI covered 23 \pm 4% of the LV surface (n=8).

9.4.3 Characteristics of prestretch

Figure 9-7 shows circumferential strains over time in the mesh representation of the LV in one animal during normal sinus rhythm. The time course of mechanical mapping showed substantial heterogeneity in strain distribution. The viable myocardial tissue that occupied the majority of the LV underwent normal systolic deformation sequence, shown in light green (0%), cyan (-5%), blue (-10%), then dark blue (-15%). In contrast, the MI region (circumscribed by a solid line) underwent substantial systolic stretch, shown in yellow (5%) as viable myocardium shortens (blue). The viable myocardium in the lateral region exhibited significant prestretch (yellow-red) during early systole. Significant prestretch (red) appeared around 30msec in the lateral, viable myocardium when the majority of the LV has not started contraction yet (light green). Prestretch peaked at 70-100msec, and persisted until 240-260msec. Regional heterogeneity of prestretch-related mechanical indices is shown in Figure 10-8 and Table 9-3. Comparison of mechanical indices between the viable vs. infarct myocardium is summarized in Figure 9-9.

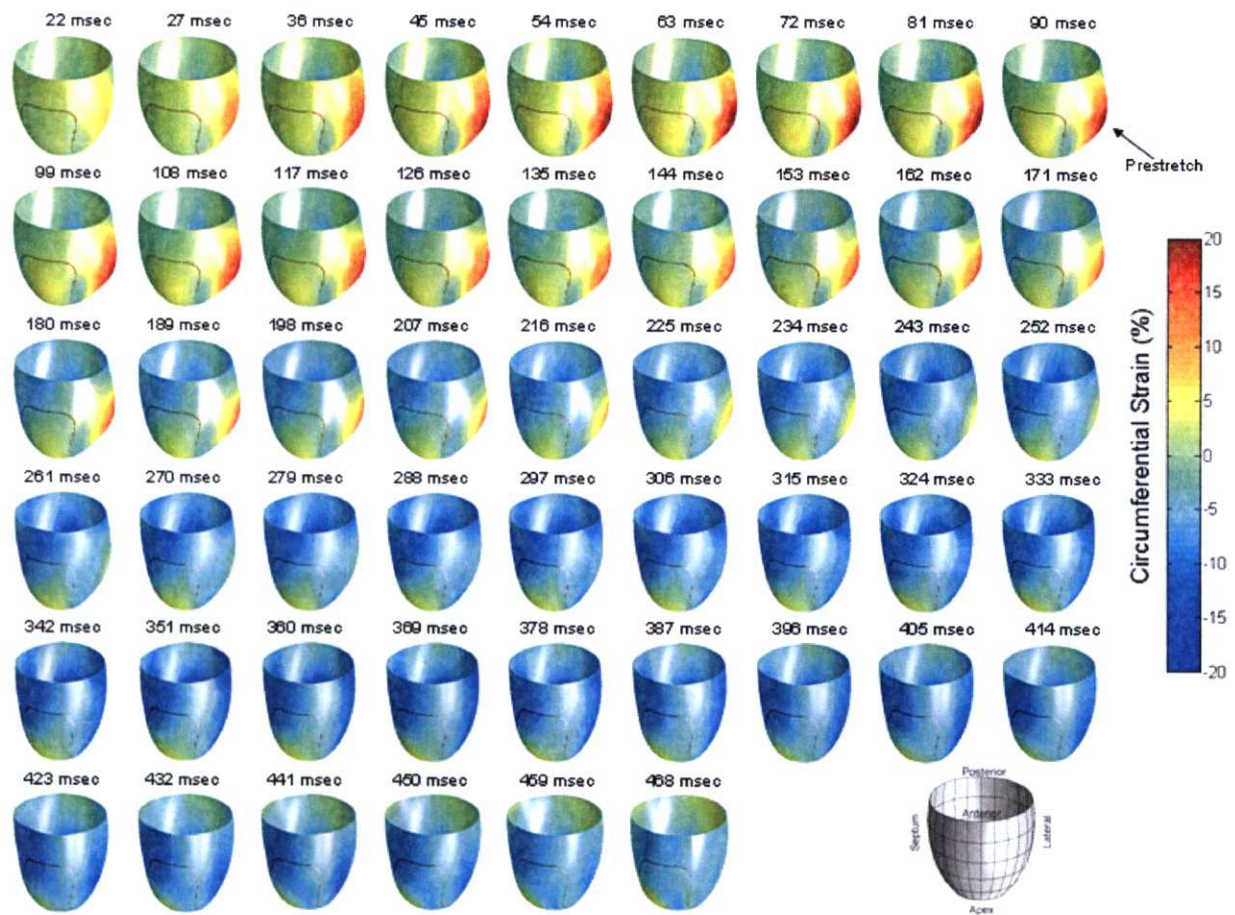


Figure 9-7: Time course of circumferential strain in one animal. Note that prestretch (shown in yellow-red) appears at around 30msec in the lateral region when the majority of the LV myocardium has not started contraction yet (shown in light green). Prestretch peaked at 70-100msec, and persisted until 240-260msec. The area circumscribed by a solid line represents MI (>50% wall depth). Note the circumferential length change appears greater than actual change *in vivo*.

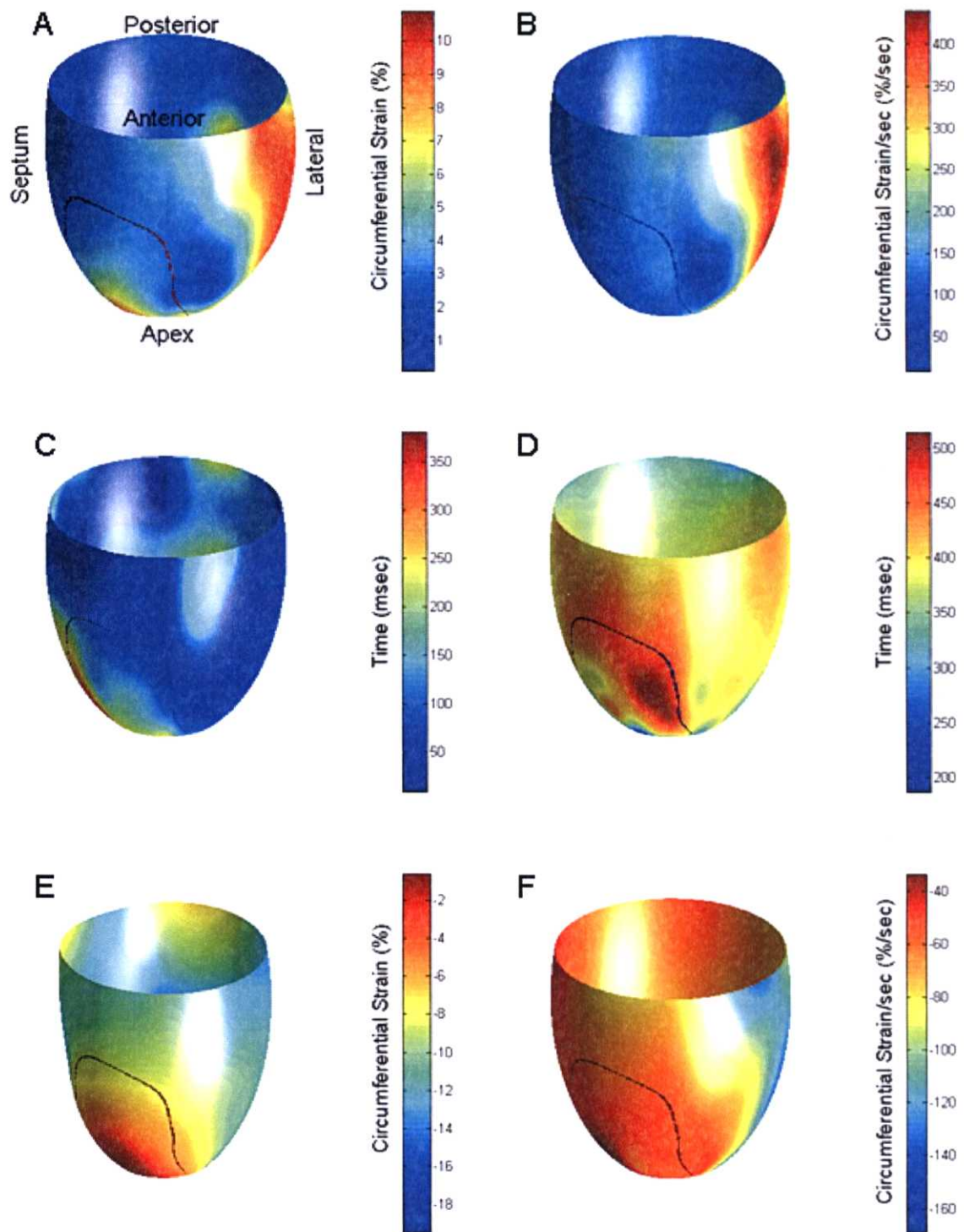


Figure 9-8: *Average mechanical indices by region (n=8)*. A. Prestretch (circumferential strain, %), B. Peak prestretch rate (circumferential strain/sec, %/sec), C. Time to onset of shortening (time, msec), D. Time to peak shortening (time, msec), E. Peak shortening (circumferential strain, %), F. Peak shortening rate (circumferential strain/sec, %/sec). The area circumscribed by a solid line represents MI (> 50% wall depth).

Parameters	Average	Minimum	Maximum
Prestretch (%)	4.6 ± 5.3	0.4 ± 0.2	10.8 ± 7.4*
		Septal (viable)	Posterolateral (viable)
Peak prestretch rate (%/sec)	176.3 ± 199.0	7.1 ± 3.3	442.3 ± 324.6*
		Septal (viable)	Lateral (viable)
Time to onset of shortening (msec)	112.7 ± 150.4	34.2 ± 10.1	392.9 ± 310.4*
		Posterolateral (viable)	Anteroseptal (MI)
Time to peak shortening (msec)	392.3 ± 91.4	199.9 ± 299.1	523.1 ± 79.0*
		Septal (viable)	Anteroseptal (MI)
Peak shortening (%)	-11.3 ± 4.7	-19.6 ± 4.5	-0.7 ± 1.3*
		Posterolateral (viable)	Anteroseptal (MI)
Peak shortening rate (%/sec)	-80.0 ± 43.9	-170.5 ± 91.7	-33.6 ± 12.0*
		Lateral (viable)	Anteroseptal (MI)

Table 9-3: *Minimum vs. maximum mechanical indices by region.* Values are means±SD (n=8). *: P<0.05 vs. Minimum.

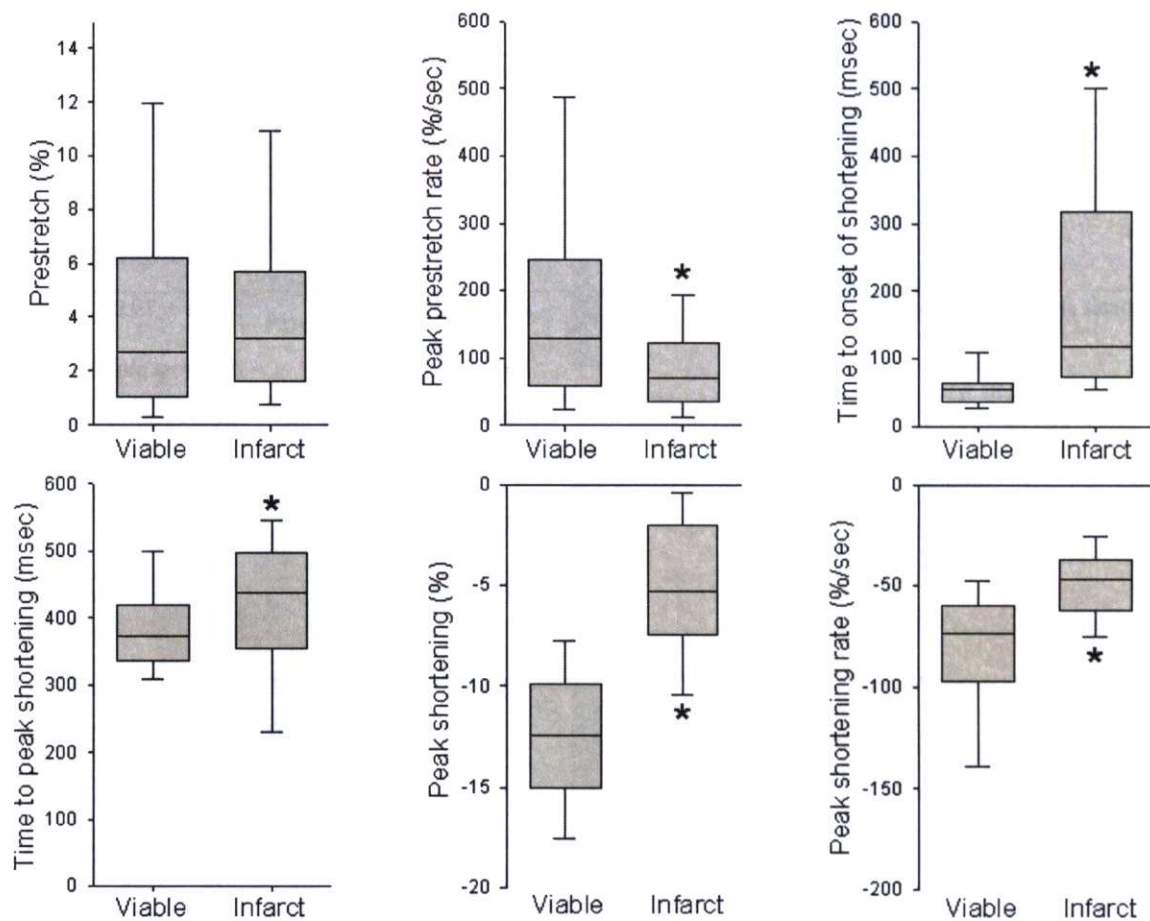


Figure 9-9: Comparison of mechanical indices between viable vs. infarct myocardium. These plots contain the pooled data from all the data points from all animals. The boundary of the box closest to zero indicates the 25th percentile, a line within the box marks the median, and the boundary of the box farthest from zero indicates the 75th percentile. Whiskers above and below the box indicate the 90th and 10th percentiles. *: P<0.05 vs. viable myocardium.

9.4.4 Correlation between prestretch and electromechanical indices

There was a significant correlation between prestretch and peak shortening rate in the viable myocardium (Figure 9-10).

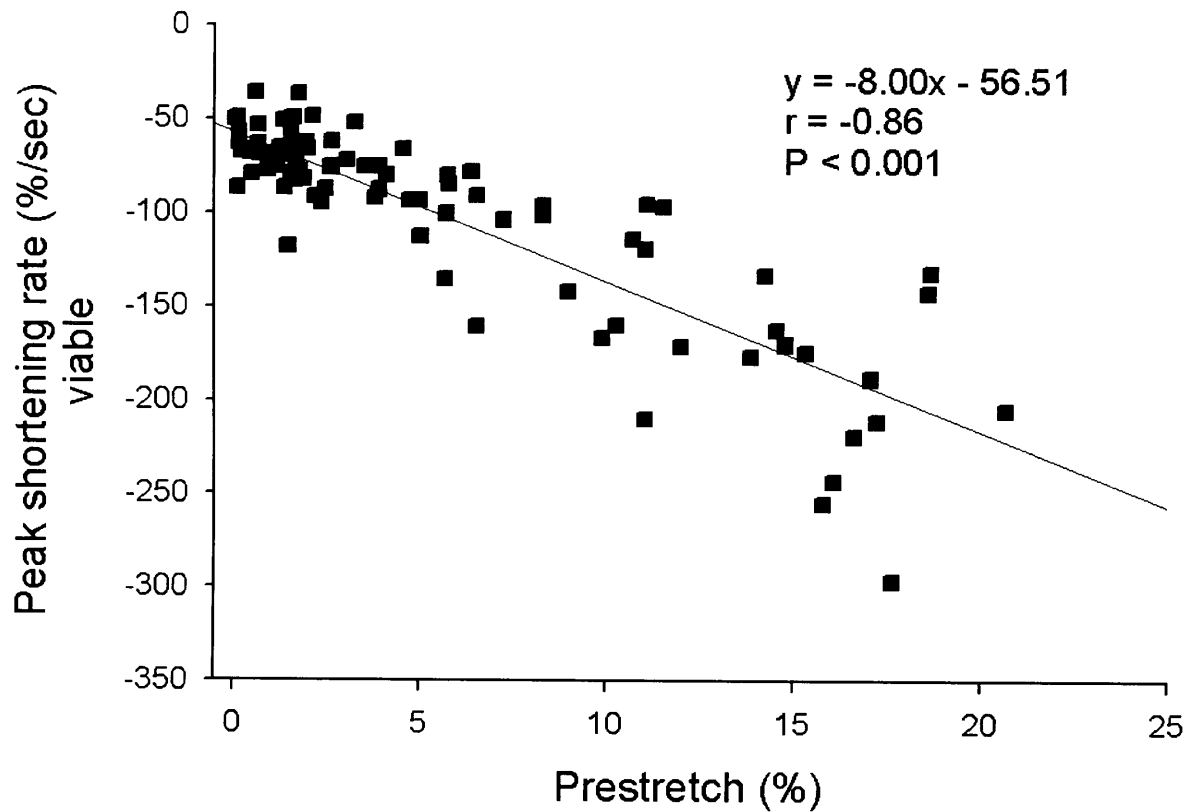


Figure 9-10: Correlation between prestretch and peak shortening rate in the viable myocardium in one animal ($r=-0.75$, $P<0.001$).

The correlation between prestretch and no other mechanical indices were consistently significant (Figure 9-11). The correlation between prestretch and ARI was analyzed in the prestretch region, which was the viable myocardium mostly in the lateral LV. The spatial distribution of greater prestretch was very similar to the distribution of longer ARI (Figure 9-12). A total of 24 ± 6 data points out of 144 in each heart were used for analysis. Figure 9-13 shows a strong correlation between prestretch and ARI in one animal ($r=0.89$, $P<0.001$). This correlation was consistently significant among all animals (Figure 9-11).

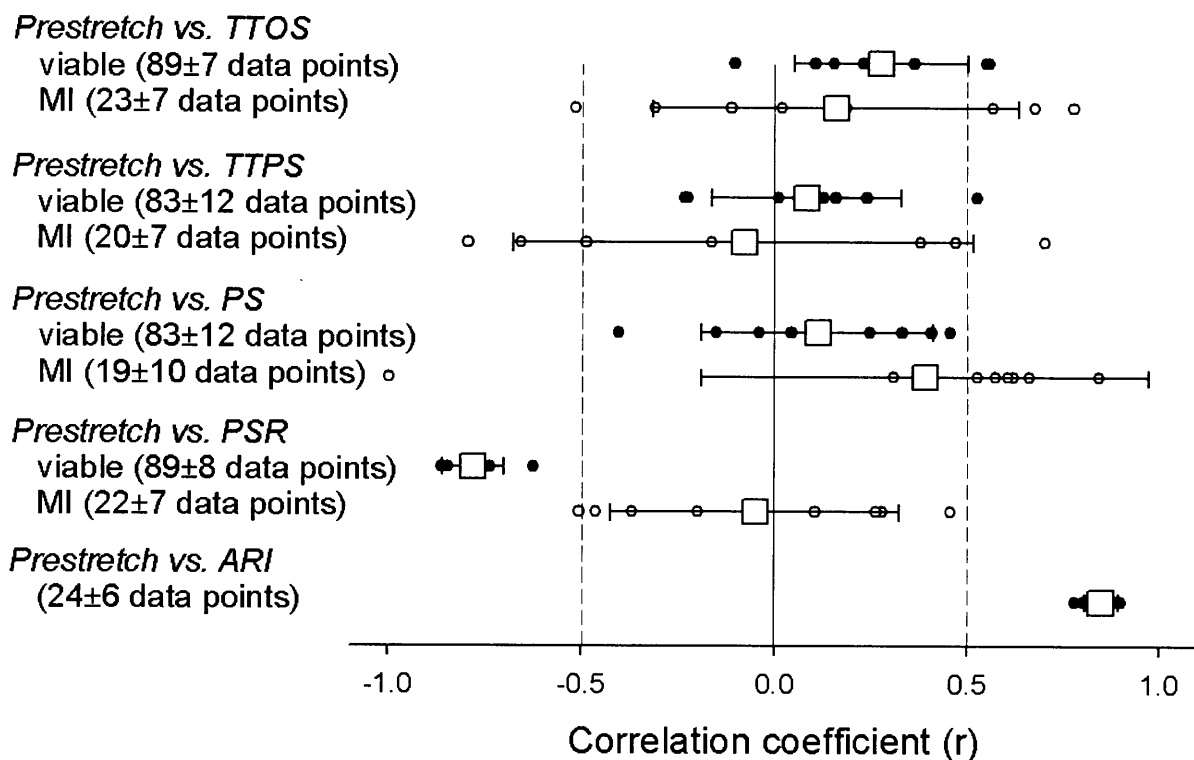


Figure 9-11: *Correlation coefficients for each animal.* Individual correlation coefficients are shown in closed (viable myocardium) and open (infarct myocardium) circles, and the open squares indicate mean \pm SD of the correlation coefficient in each animal. The number of data points per animal (mean \pm SD) to calculate each correlation coefficient is also shown. The horizontal dashed lines indicate $|r|=0.5$. TTOS, time to onset of shortening; TTPS, time to peak shortening; PS, peak shortening; PSR, peak shortening rate; ARI, activation recovery interval.

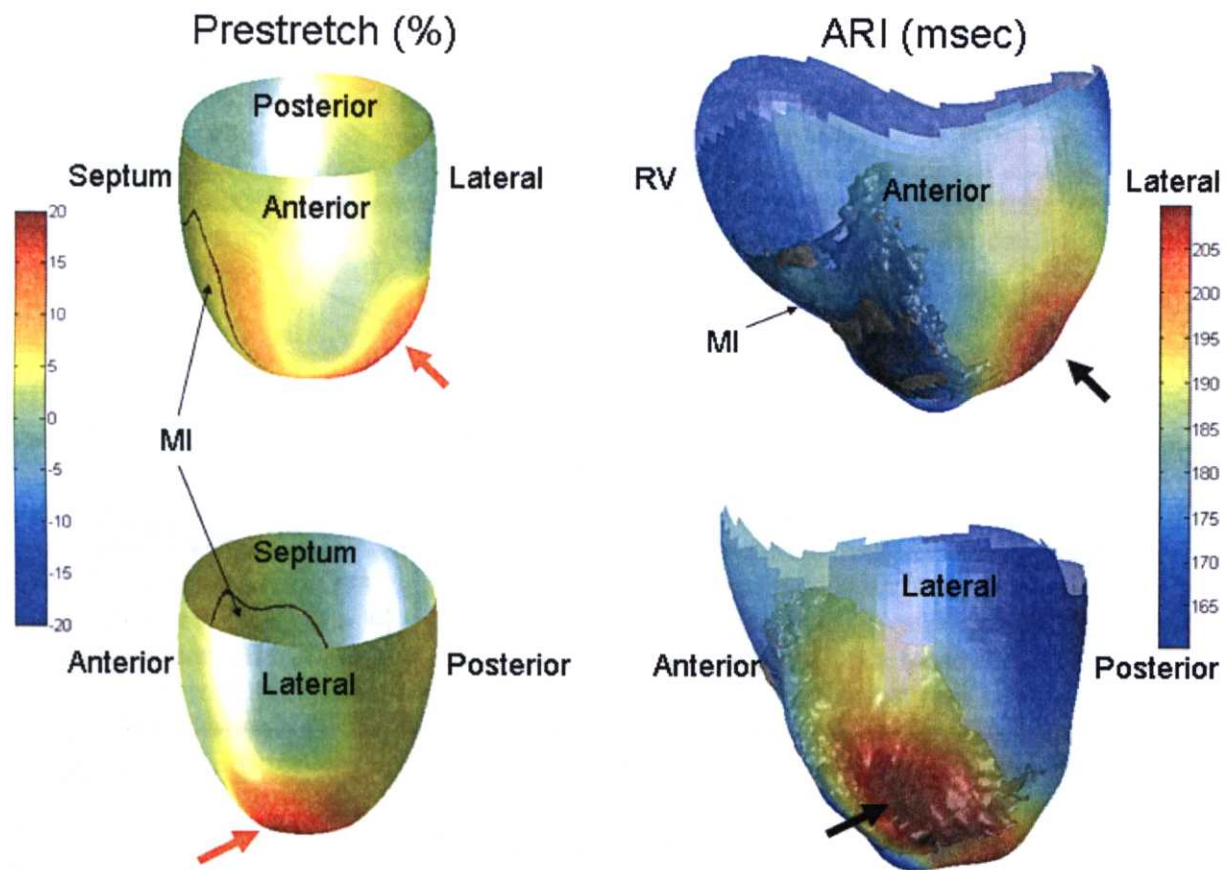


Figure 9-12: *Spatial distribution of prestretch and ARI in one animal.* The prolate ellipsoid mesh on the left represents prestretch (color-coded in %) derived from the cine-DENSE images, whereas the epicardial mesh on the right represents ARI (color-coded in msec) derived from the electrical sock measurement. The spatial distribution of greater prestretch (red arrow) was very similar to the distribution of longer ARI (black arrow).

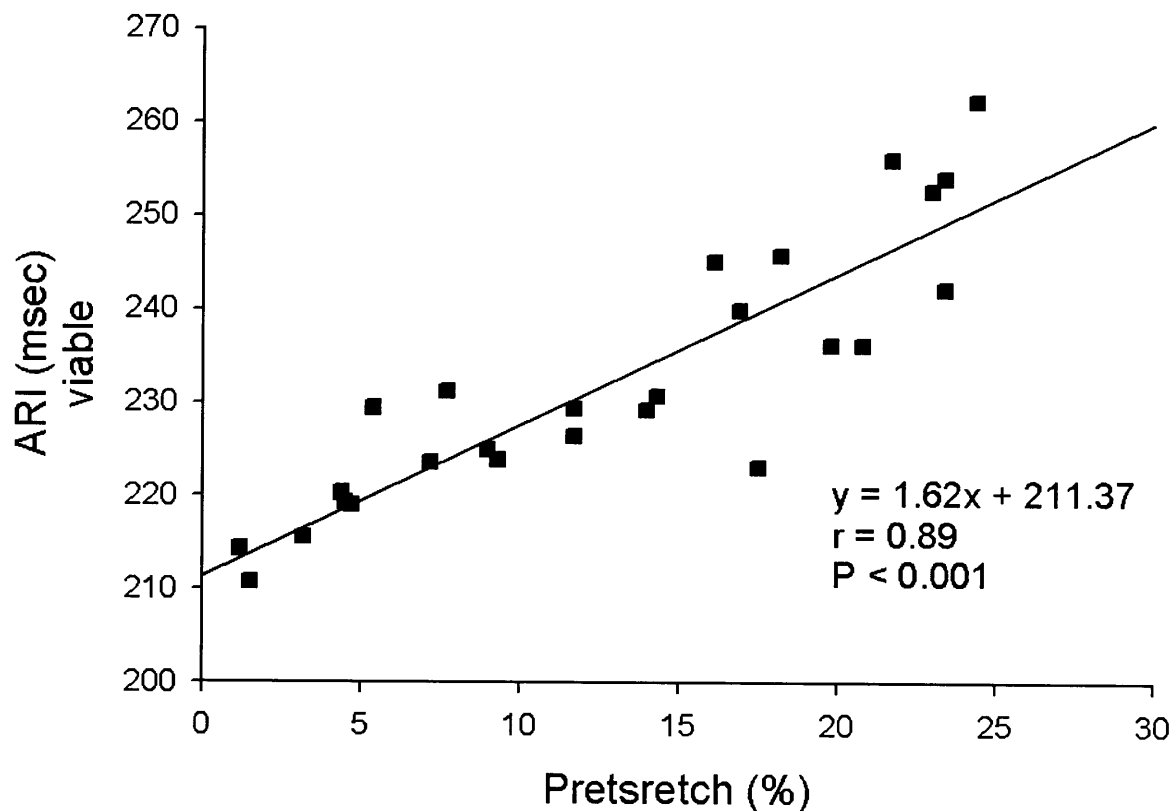


Figure 9-13: Correlation between prestretch and ARI in the prestretch region in one animal ($r=0.92$, $P<0.001$).

9.5 Discussion

9.5.1 Overview

The present study combined noninvasive MR-based techniques and invasive electrical measurements to examine the anatomical correlation of abnormal myocardial stretch and ARI in chronic MI. The quantitative mechanics using the cine-DENSE MRI demonstrates excellent agreement with that of tagged MRI (137), and cine-DENSE MRI requires less user input for post-processing than tagged MRI (252). Furthermore, high temporal resolution (9ms) allowed characterization of the timing, location, and magnitude of prestretch with high fidelity.

Although alteration of APD in response to acute loading has been studied at the global ventricular level, this is the first study to quantitatively demonstrate the magnitude-dependent mechano-electrical feedback (MEF) at a local level *in vivo*. The regionally heterogeneous prestretch likely generates a heterogeneous distribution APD, which results in an increased dispersion of repolarization and thus increased susceptibility to ventricular tachyarrhythmia. This finding is consistent with a recent report that the presence of left bundle branch block and resultant mechanical dyssynchrony in patients with heart failure is an independent risk factor of all cause mortality and sudden death (28)

9.5.2 Animal model of chronic myocardial infarction with substrate for sudden cardiac death

The occlusion-reperfusion MI model in the present study reflects a clinical scenario where acute coronary artery occlusion is followed by revascularization procedures, which is clinically associated with inducible VT (136). A closed-chest animal model of chronic MI within the first month after MI was used, because the risk of SCD is highest in this time period (220). Sustained monomorphic VT was inducible in all the animals studied, which confirms that those hearts have a substrate for SCD.

9.5.3 Mechanical characteristics of prestretch in chronic myocardial infarction

Prestretch appeared in early systole when the majority of the LV myocardium has not started contraction yet (Figure 9-8). This suggests that prestretch is not initiated by contraction of other ventricular regions, but by ventricular filling, likely due to atrial contraction. Diastolic expansion of the MI tissue is limited due to its 'stiff' material property relative to that of the viable myocardium, thus the region of largest expansion coincides with the more elastic, viable myocardium opposite to the MI. As a result, the regional pattern of prestretch was similar to that of normal human heart, being lowest in the septum and highest in the posterolateral region (82, 272). The posterolateral region opposite to the MI experiences largest and quickest prestretch, and quickest shortening (Figure 9-8A, B, F). Prestretch significantly correlated with peak shortening rate (Figure 9-10) which may reflect the regional Frank-Starling mechanism (272). However, no significant correlation was observed between prestretch and other mechanical indices (Figure 9-11). This implies disruption of the regional Frank-Starling mechanism in chronic MI, due to significantly depressed mechanical function in the infarct myocardium (Figure 9-9).

9.5.4 Prestretch and stretch-induced arrhythmia

The magnitude of prestretch in the data was substantially higher than those reported in normal human heart (10% vs. 4-6%) (82, 272) (Table 9-3). Because the magnitude of stretch is important in the induction of stretch-induced arrhythmia (55, 91, 128, 139, 263), these results support the involvement of prestretch in triggering fatal arrhythmia in chronic MI. In addition, the relatively long time period of prestretch (Figure 9-7) sufficiently covers the vulnerable window for stretch-induced arrhythmia (142), thus provide a mechanistic support for SCD in this population.

The results suggest that prestretch in chronic MI prolongs local ARI in a magnitude-dependent manner (Figure 9-10, 9-11, 9-12). This finding is consistent with a recent report by Nishimura *et al* using a carbon fiber technique and voltage-sensitive dye in single ventricular myocytes in rat (190). In contrast, earlier studies show that acute loading can shorten (204, 262) or prolong (170, 227) APD, possibly mediated by SACs (118), beta-adrenergic receptor activation (153) or actin cytoskeleton (190). One possible factor to explain the finding in light of these studies is the timing of stretch. Stretch results in shortening of APD during the early plateau phase and prolongation of APD at later stages of repolarization (90, 141, 146, 261). The results show that prestretch persists until 240-260msec, and the relatively long period of prestretch may cause consistent prolongation of APD in the region of prestretch. Another possible factor is that structurally remodeled hearts with chronic MI *in vivo* was used. Stretch-induced MEF was mostly studied in isolated normal rabbit heart (26, 80, 153, 204, 227, 262), and electrophysiological properties, including conduction velocity and MEF, of the structurally

remodeled myocardium are most likely different from those of the normal myocardium. For example, the myofiber and laminar structure of the myocardium are the structural bases of the orthogonal anisotropy of electrical propagation (116, 150), and these cardiac microstructures undergo significant alterations in the structurally remodeled hearts in a regionally heterogeneous manner (20, 110).

9.5.5 Clinical implications

The findings have two important clinical implications. First, the MR-guided, noninvasive quantification of prestretch may help identify the population at risk for SCD, along with other promising noninvasive diagnostic techniques including signal-averaged ECG (62), T-wave alternans (TWA) (36) and heart rate variability (HRV) (49). Further studies are needed to assess the risk of SCD in human patients with prestretch.

Second, the MR-guided quantification of prestretch may guide early interventions as primary prevention of SCD. Early interventions to modulate prestretch by multisite pacing combined with pharmacological interventions to suppress stretch-activated arrhythmia may be beneficial in reducing SCD. In addition, as demonstrated in CRT trials, suppression of prestretch would also prevent and even reverse structural remodeling in the viable myocardium, which would further reduce mortality in the longer period (99).

9.5.6 Limitations

Control studies were not conducted, therefore the electromechanical indices were not compared between control and infarct animals. The heart rate during electrical measurements was significantly higher than that of mechanical measurements (129 ± 13 vs. 97 ± 28 bpm, $P=0.019$), most likely due to reduced preload associated with thoracotomy. Earlier studies suggest that ARI measurements are sensitive to loading conditions, thus the magnitudes of ARI may have been underestimated. The LV mechanics was examined in anesthetized, closed-chest swine. Therefore, the results may not precisely reflect the cardiac mechanics in conscious animals. Species difference (swine vs. human) is another factor to consider when these results are clinically extrapolated. The strain data in the present study describe the average transmural mechanics in the LV myocardium *in vivo*. Although significant transmural variations in timing and magnitude of prestretch have been described (16), the aim of the present study was to identify regional variations, not transmural variations, of cardiac strains. To maintain a sufficient signal-to-noise ratio (SNR), the tissue-tracking MRI sequence required an interval for T_1 recovery between the last phase in one cardiac cycle and the beginning of the next cardiac cycle. Therefore the mechanical indices for an entire cardiac cycle was not studied. However, the aim of the present study was to examine systolic shortening, and the 85% coverage of the cardiac cycle (Table 9-2) was more than sufficient for this objective.

9.6 Conclusions

Prestretch in chronic MI prolongs local ARI in a magnitude-dependent manner in swine with a substrate for SCD. The stretch-induced ARI prolongation may increase the dispersion of repolarization and thus susceptibility to ventricular tachyarrhythmia. Early interventions to reduce prestretch may help reduce SCD in patients with chronic MI.

Chapter 10

Thesis conclusions

The overall goal of the thesis was to assess anatomical correlation between cardiac mechanics and electrophysiology *in vivo*, and to investigate whether it is possible to predict electrical behaviors from mechanical behaviors of the heart. There were six specific aims to achieve this goal, and the conclusions from these aims are:

Aim 1: LV mechanics during early relaxation involves substantial deformation of fiber and sheet structures with significant transmural heterogeneity. Predominant epicardial stretch along myofibers during isovolumic relaxation appears to drive global torsional recoil to aid early diastolic filling (Chapter 4)

Aim 2: Enhanced filling in volume-overload hypertrophy is achieved by enhanced sheet shear early in diastole. These results provide the first evidence that changes in motion of radially oriented laminar sheets may play an important functional role in pathology of diastolic dysfunction in this model (Chapter 5).

Aim 3: Despite lack of evidence of significant transmural gradient in electrical repolarization *in vivo*, there is transmural dispersion of myofiber relaxation as well as shortening (Chapter 6).

Aim 4: Normal sheet extension and wall thickening immediately after activation may require normal transmural activation sequence, whereas sheet shear deformation may be determined by local anatomy. (Chapter 7).

Aim 5: The infarct border zone is characterized by abnormal mechanics directly coupled with normal electrical depolarization. This indicates that impaired function in the border zone is not contributed by electrical factors but results from mechanical interaction between ischemic and normal myocardium (Chapter 8)

Aim 6: Prestretch in chronic MI prolongs local ARI in a magnitude-dependent manner in swine with a substrate for SCD. The stretch-induced ARI prolongation may increase the dispersion of repolarization and thus susceptibility to ventricular tachyarrhythmia. Early interventions to reduce prestretch may help reduce SCD in patients with chronic MI (Chapter 9).

These results underscore the fact that cardiac anatomy (e.g. fiber, sheet structures), electrophysiology (e.g. activation sequence) and mechanics (e.g. shear, prestretch) are indispensably intertwined, thus interdependent. Although this thesis focused on prediction of the susceptibility to SCD as a clinical output, the interdependence of anatomy, electrophysiology and mechanics could be further utilized in future research to extract clinically relevant information.

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List of Abbreviations and Acronyms

AF	atrial fibrillation
ANOVA	analysis of variance
AoP	aortic pressure
APD	action potential duration
ARI	activation-recovery interval
AV	atrioventricular
bpm	beats per minute
BW	bandwidth
CHD	coronary heart disease
CL	cycle length
CRT	cardiac resynchronization therapy
DENSE	displacement encoding of stimulated echo
EC	early contraction
ECG	electrocardiogram
ECM	extracellular matrix
ED	End diastole
EDP	End-diastolic pressure
EDV	End-diastolic volume
EF	ejection fraction
EPI	echoplaner imaging
FOV	field of view
Gd-DTPA	Gadolinium diethylene triamine pentaacetate
GRE	gradient recalled echo
ICD	implantable cardioverter-defibrillator
IHD	ischemic heart disease
IVC	isovolumic contraction
IVR	isovolumic relaxation
LA	left atrium
LAA	left atrial appendage
LAD	left anterior descending coronary artery
LAP	left atrial pressure
LBBS	left bundle branch block
LCx	left circumflex coronary artery
LV	left ventricle
LVP	left ventricular pressure
MEF	mechanoelectrical feedback
MI	myocardial infarction
MRI	magnetic resonance imaging
MVO	mitral valve opening
PES	programmed electrical stimulation

PPR	peak prestretch rate
PS	peak shortening
PSR	peak shortening rate
RV	right ventricle
RVOT	right ventricular outflow tract
SAC	stretch-activated channel
SCD	sudden cardiac death
SD	standard deviation
SNR	signal-to-noise ratio
SSFP	steady state free precession
TdP	torsade de pointes
TDR	transmural dispersion of repolarization
TDRelax	transmural dispersion of myofiber relaxation
TDSHORT	transmural dispersion of myofiber shortening
TE	echo time
TR	repetition time
TTOS	time to onset of shortening
TTPS	time to peak shortening
VF	ventricular fibrillation
VT	ventricular tachycardia

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Chapter 4

Regional mechanics during relaxation in normal heart

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Chapter 5

Regional mechanics during relaxation in hypertrophied heart

Figure 5-1. A. Schematic representation of the heart. X_1 : circumferential axis, X_2 : longitudinal axis, X_3 : radial axis, LAD: left anterior descending coronary artery, LCx: left circumflex coronary artery, LV: left ventricle, LA: left atrium. A precalibrated micromanometer pressure catheter was inserted into apex to monitor LV pressure. A 9-Fr silicone catheter was inserted into LA to monitor LA pressure, which was used in every imaging study to correct for baseline drifting of the LV micromanometer. **B. Bead set** consisted of three transmural columns of 4-6 gold beads (0.8mm) and a surface gold bead (1.7mm) above each column. Finite deformation of the myocardium was calculated from displacement of each material point (= bead) in space. **C. Diastolic filling.** ECG: surface electrocardiogram, LVP: left ventricular pressure, LAP: left atrial pressure. Diastolic filling was defined as the period beginning at mitral valve opening (time=0%, pressure crossover of LVP and LAP) and ending at end diastole (time=100%, peak of ECG R-wave). White squares in the figure represent longitudinal strain (E_{22}) at 80% wall depth in a control animal. **D. Fiber-sheet coordinate system.** Each cylinder represents a myofiber. Myofibers are organized into laminar “sheets”, which are approximately four cells thick and roughly stacked from apex to base (150). Sheet angle (β) is measured with reference to the positive radial axis (X_3).

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Figure 5-3. Transmural fiber and sheet angles during diastolic filling. Values are means \pm SE (n=6). Closed square: mitral valve opening, open square: end diastole, Epi: epicardial surface, Endo: endocardium. Both in control and hypertrophy, transmural fiber angles did not significantly change from mitral valve opening to end diastole ($P=n.s.$). Similarly, transmural sheet angles did not change during diastolic filling in control ($P=n.s.$). In contrast, transmural sheet angles significantly decreased from mitral valve opening to end diastole in hypertrophy (*: $P<0.05$). See text for details.

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Chapter 6

Regional electromechanics in normal heart

Figure 6-1. Schematic representation of the heart. The transmural bead set was implanted between the first (D_1) and the second (D_2) diagonal branch of the left anterior descending coronary artery (LAD) to measure finite deformation of the myocardial tissue across the wall. Epi, epicardium; Endo, endocardium; LV, left ventricle; LCx, left circumflex coronary artery.

Table 6-1. Hemodynamic parameters. Values are mean \pm SD (n=14). LVP_{max} , maximum left ventricular pressure; dP/dt , first derivative of pressure over time; dP/dt_{max} , peak positive dP/dt ; dP/dt_{min} , peak negative dP/dt .

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Chapter 7

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Figure 7-1. A: Schematic representation of the left ventricle. X_1 : circumferential axis, X_2 : longitudinal axis, X_3 : radial axis, LAD: left anterior descending coronary artery, D_1 , D_2 : first and second diagonal branch of LAD, respectively. **B: Schematic representation of local fiber-sheet axes.** Fiber angle (α) was measured in the circumferential-longitudinal (X_1 - X_2) plane at each wall depth with reference to the positive circumferential axis (X_1). Sheet angle (β) was measured in the plane perpendicular to the fiber angle at each wall depth with reference to the radial axis (X_3). X_f : fiber axis, X_s : sheet axis, X_n : axis oriented normal to the sheet plane. The X_f , X_s , and X_n axes present a Cartesian system (For details, see (17)).

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Chapter 8

Global electromechanics in infarct heart

Figure 8-1. Epicardial electrical signals during a cardiac cycle (cycle length = 544 msec, HR = 110 bpm) registered to a short-axis delayed hyperenhancement (DHE) MR image of the LV split into 32 sectors. Regions of bright intensity correspond to MI. In this image, abnormal or delayed activation is seen from sectors 4 through 9, whereas the infarct ranges from sectors 3 through 14.

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Figure 8-6. 3-D Strain map from a single animal. The area circumscribed by a solid white line represents the infarct zone. E_{rr} : radial, E_{cc} : circumferential, and E_{ll} : longitudinal strains. The regions of abnormal strains, particularly the longitudinal strain, extended far beyond the infarct zone, and the strains were larger in the epicardium than in the endocardium. Epi: epicardial layer, Endo: endocardial layer.

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Chapter 9

Global electromechanics in infarct heart with substrate for sudden cardiac death

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to reflect the reference configuration. **C. Mesh reconstruction of LV.** Mesh vertices indicate locations of displacement and strain calculations. There are a total of 144 data points per heart (=24 sectors x 6 short-axis slices).

Figure 9-5. A. Short-axis view of the contrast-enhanced MR image. Regions of high intensity (= light gray) corresponds to MI. **B. Mesh presentation of the MI wall depth (%) in one animal.** Note the MI was mostly transmural (= dark red), and was located in the anteroseptal region.

Figure 9-6. Unipolar electrical recordings from an epicardial sock electrode. A. Normal sinus rhythm, B. Sustained monomorphic VT.

Figure 9-7. Time course of circumferential strain in one animal. The numbers indicate the time from the peak of ECG-R wave. The area circumscribed by a solid line represents MI (>50% wall depth). Note the circumferential length change appears greater than actual change *in vivo*.

Figure 9-8. Average mechanical indices by region (n=8). A. Prestretch (circumferential strain, %), B. Peak prestretch rate (circumferential strain/sec, %/sec), C. Time to onset of shortening (time, msec), D. Time to peak shortening (time, msec), E. Peak shortening (circumferential strain, %), F. Peak shortening rate (circumferential strain/sec, %/sec). The area circumscribed by a solid line represents MI (> 50% wall depth).

Figure 9-9. Comparison of mechanical indices between viable vs. infarct myocardium. These box plots contain the pooled data from all the data points from all animals. The boundary of the box closest to zero indicates the 25th percentile, a line within the box marks the median, and the boundary of the box farthest from zero indicates the 75th percentile. Whiskers above and below the box indicate the 90th and 10th percentiles. *: P<0.05 vs. viable myocardium.

Figure 9-10. Correlation between prestretch and peak shortening rate in the viable myocardium in one animal (r=-0.75, P<0.001).

Figure 9-11. Correlation coefficients for each animal. Individual correlation coefficients are shown in closed (viable myocardium) and open (infarct myocardium) circles, and the open squares indicate mean±SD of the correlation coefficient in each animal. The number of data points per animal (mean±SD) to calculate each correlation coefficient is also shown. The horizontal dashed lines indicate |r|=0.5. TTOS, time to onset of shortening; TTPS, time to peak shortening; PS, peak shortening; PSR, peak shortening rate; ARI, activation recovery interval.

Figure 9-12. Spatial distribution of prestretch and ARI in one animal. The prolate ellipsoid mesh on the left represents prestretch (color-coded in %) derived from the cine-DENSE images, whereas the epicardial mesh on the right represents ARI (color-coded in msec) derived from the electrical sock measurement. The spatial distribution of greater prestretch (red arrow) was very similar to the distribution of longer ARI (black arrow).

Figure 9-13. Correlation between prestretch and ARI in the prestretch region in one animal (r=0.92, P<0.001).