Published in final edited form as:

J Electrocardiol. 2002 ; 35(Suppl): 61–64.

Electromechanical Mapping With MRI Tagging and Epicardial Sock Electrodes

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Abstract

Methods currently exist for the precise measurement of local three-dimensional myocardial motion noninvasively with magnetic resonance imaging tagging. From these motion estimates, strain images representing the local deformation of the myocardium can be formed to show local myocardial contraction. These images clearly show the sequence of mechanical events during the activation and relaxation of the heart, making them ideal to visualize abnormalities caused by asynchronous electrical activation or ischemia. Coupled with the near simultaneous mapping of electrical depolarization with a sock electrode array, we can investigate the relationship between electrical activity and mechanical function on a local level in the *in vivo* heart. Registered fiber angle maps can also be obtained in the same heart with diffusion magnetic resonance imaging to assist in the construction of the electromechanical model of the whole heart.

The ability to measure the precise mechanical function, the electrical function, and the underlying fiber architecture in the same heart *in vivo* may uncover the interactions of these constituents in normal and abnormal cardiac function. The detailed relationship between local muscle shortening, electrical activation, and myocardial fiber angle has been difficult to appreciate due to the inability to measure these features of the heart in the same preparation.

The measurement of the electrical activity of the heart can be accomplished with numerous techniques including intracavity electrodes (1), multi-electrode baskets, optical techniques (2), monophasic action potentials (3) and body surface potential mapping (4–6).

MRI has made contributions to the understanding of myocardial mechanics by providing methods for measuring local three dimensional myocardial motion non-invasively using presaturation tagging patterns (7–9). Recently, methods for measuring the diffusion tensor with MRI have led to a method for measuring the myocardial fiber angle in soft tissue (10,11).

We have developed an experimental protocol in which electrical mapping, myocardial strain mapping, and fiber angle mapping can be achieved in the same heart (12). The data are registered so that local correlations can be made among these 3 features.

In MRI tagging a set of saturation pulses placed in the tissue provides a spatially varying signal intensity pattern in the tissue; *the change in shape of the intensity pattern in the image reflects the change in shape of the underlying body containing the intensity pattern*. Originally demonstrated by Zerhouni et. al. (7) with saturation pulses, and by Axel (8) with SPAMM pulses, these techniques have become a mainstay for imaging myocardial function (13). Figure

1 shows a short axis slice of the heart at two time points in the heart cycle; just prior to pacing the right ventricle (RV) apex, and just after pacing. The ability to measure local mechanics from the deformation of the tagging pattern is very clear.

The objective of the analysis of tagged images is to track the three-dimensional motion of each material point in the heart, and then to compute the six components of the strain tensor at each point for a sequence of time points throughout the heart cycle. The strain tensor characterizes the *local* deformation of the heart wall, and components of this tensor, such as that in the circumferential direction are an excellent index of local myocardial contraction. To obtain precise quantification of the regional strains, the position of the tags must be measured with a tag detection algorithm (14). Once the relative positions of the tags have been determined as a function of time, these data can be used to estimate the strain tensor at each point in the myocardium. One very effective method for doing this is a displacement field model based on B-splines (15,16).

Measurement of Myocardial Function during Asynchronous Activation

The precise sequence of events during ectopic excitation is particularly evident on colorized images of the three-dimensional strain tensor. Figure 2 shows the evolution in time of the circumferential component of the three-dimensional strain tensor evaluated at the mid-wall for two ventricular pacing sites; because the fiber direction is essentially circumferential at the midwall, this component of the strain tensor closely matches muscle fiber shortening. During atrial pacing, the ventricle is activated through the normal Purkinje pathway and muscle shortening evolves relatively homogeneously over the ventricle; this is demonstrated with the uniformly increasing blue color over the ventricle in the top row of Figure 2. With epicardial ventricular pacing, early mechanical activation was observed at the pacing site, followed by propagation of a contraction wave-front to the opposite side of the heart. There was also a significant “prestretch” of the late activated myocardium on the opposite side of the heart from the pacing site, shown as a bright yellow color. This prestretch was quite pronounced (15%–20% in some cases) and occurred in the first 100 ms after the ventricular pacing pulse.

An alternative way of visualizing the contraction pattern is to graph the time course of strain for each material point of the heart. Each graph can be mapped to a position in an array that corresponds to a position in the heart. Examples of such strain curves are shown in Figure 3, in which the sequence of circumferential stretching and shortening for selected subepicardial sites in the LV freewall and the RV freewall is plotted as blue curves. When the blue curve is rising, that means the tissue is lengthening; when the blue curve is falling, that means the tissue is shortening. With the electrode sock on the same heart we are able to measure local unipolar electrograms at the same locations and overlay those electrograms on the mechanical data. The electrograms for this heart are shown as red curves. From these data, we can study the relationships between mechanical function, such as the onset of contraction (shortening) and relaxation (stretching), and the associated electrical phenomena such as the depolarization and repolarization.

Fiber Angle Measurement and Modelling

Magnetic resonance imaging (MRI) diffusion imaging can yield high resolution maps of the local myocardial fiber geometry. Figures 4A and B show maps of fiber orientation for the heart measured *in vitro* with diffusion tensor imaging. These measurements are made after fixation to achieve very high resolution (0.7 x 0.7 x 0.9 mm voxels) using long imaging times (days). The diffusion of water in the muscle was sampled with diffusion encodings oriented in 28 different directions, each of which had a diffusion length of ~0.7 μm . The map shows a clear transmural gradient in fiber angle from the epicardium to the endocardium. This nondestructive method produces the fiber map while keeping the geometry of the heart essentially intact. Only

small geometric distortions need to be corrected to overlay this fiber information back onto the *in vivo* data obtained with the MRI tagging studies. The structure of this mapping will be key in determining the activation pathways for the propagation of electrical activation, and it may also have a bearing on the mechanical performance of the tissue.

Acknowledgements

Many of the results and conclusions reported here are from the collaborative efforts of the Cardiac MRI Research Group at Johns Hopkins University (JHU) and the Medical Imaging Group at NHLBI. The authors would especially like to acknowledge the efforts of Scott Chesnick, Michael Guttman, Frits W. Prinzen, Joni Taylor, Brad Wyman, Joshua Tsitlik, Henry Halperin, Bill Hunter, and Elias Zerhouni. The fiber angle mapping is a joint project with Dr Rai Winslow at JHU. Electrical mapping techniques were taught to us, and the electrode sock was provided by Bob Lux and Rob MacLeod at the CVRTI, University of Utah.

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Fig. 1.

Two images of a short axis slice in a heart that has been tagged with a grid pattern. (A) The heart just before the application of a pacing stimulus. (B) The mechanical deformation of the myocardium in response to the early shortening in the septum. The precise local strain can be measured using the deformation of the tagging pattern as shown by the colored squares.

Fig. 2.

These colorized surfaces show the circumferential stretch (yellow) and contraction (blue) of the midwall of a single canine LV myocardium under different pacing protocols. Each row is a different pacing protocol, and time goes from left to right. The septum extends from 3 o'clock to 7 o'clock, and the location of the two pacing electrodes is marked with arrows. The early onset of contraction during ectopic pacing is shown in the second column, with significant prestretch of passive regions of myocardium shown as yellow in the third column. (LVb Pace, Left Ventricular base; RVa, Right Ventricular apex; BiV, biventricular)

Fig. 3.

The circumferential strain vs. time over a 350 ms time period during electrical depolarization/repolarization and mechanical prestretch and contraction. The strain curves (circumferential stretch/shortening) are shown in blue, the electrograms are shown in red, the time synchronized LV pressure and pacing stimulus are also shown. The location of the 4 electromechanical pairs of curves are overlaid on a volume rendering of a postmortem three-dimensional MRI data set of the canine heart. Notice the early electrical depolarization on the RVFW (point A) followed by the anterior (B) and posterior (C) points, then the late depolarization of the LVFW (point D).

Fig. 4.

(A) A “fiber angle map” in a short axis plane. The color represents the z component of the principle eigen-vector from the diffusion tensor at each location. Blue means the vector is in the plane, red means the vector is orthogonal to the plane. (B) A finite element model of the LV surface with the raw data of the fiber direction overlaid on top (left). This shows the density of the data obtained with three-dimensional diffusion MRI. The fiber orientation data can also be smoothed using a finite element fit; this is shown in the plot on the right.