Quantification of and Correction for Left Ventricular Systolic Long-Axis Shortening by Magnetic Resonance Tissue Tagging and Slice Isolation

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Background. Measurement of regional left ventricular (LV) function is predicated on the ability to compare equivalent LV segments at different time points during the cardiac cycle. Standard techniques of short-axis acquisition in two-dimensional echocardiography, cine computed tomography, and standard magnetic resonance imaging (MRI) acquire images from a fixed plane and fail to compensate for through-plane motion. The shortening of the left ventricle along its long axis during systole results in planar images of two different levels of the ventricle, leading to error in any derived functional measurements. LV systolic long-axis motion was measured in 19 normal volunteers using MRI.

Methods and Results. With a selective radio frequency (RF) tissue-tagging technique, three short-axis planes were labeled at end diastole and standard spin-echo images were acquired at end systole in the two- and four-chamber orientations. Persistence of the tags through systole allowed visualization of the intersecting short-axis tags in the long-axis images and allowed precise quantification of long-axis motion of the septum, lateral, anterior, and inferior walls at the base, mid, and apical LV levels. The total change in position along the long axis between end diastole and end systole was greatest at the base, which moved toward the apex 12.8±3.8 mm. The mid left ventricle moved 6.9±2.6 mm, and the apex was nearly stationary, moving only 1.6±2.2 mm (p<0.001). Having quantified the normal range of long-axis shortening, we developed a technique that isolates a slice of tissue between selective RF saturation planes at end diastole. Combining this with a wide end-systolic image slice, end-systolic images were acquired without contamination of signal from adjacent tissue moving into the imaging plane. This technique was validated in a moving phantom and in normal volunteers.

Conclusions. Significant LV systolic long-axis shortening exists, and this effect is seen the most at the base and the least at the apex. At a given ventricular level, shortening varied significantly according to location. A method using selective saturation pulses and gated spin-echo MRI automatically corrects for this motion and thus eliminates misregistration artifact from regional function analysis. (Circulation 1991;84:721-731)

Accurate measurement of regional left ventricular (LV) function relies on the ability to compare measures from the same LV segment throughout the cardiac cycle. Short-axis images at end diastole and end systole are compared to derive wall motion or thickening. During contraction and relaxation, significant cardiac motion exists, including translation, rigid body rotation, regional

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Presented in part at the American Heart Association 61st Scientific Sessions, November 1988, Washington, DC.
Supported in part by the McClure Fellowship, Applied Physics Laboratory, The Johns Hopkins University. M.B.B. was supported by the Dublin University Traveling Scholarship in Medicine.
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Received January 29, 1990; revision accepted April 9, 1991.
twist, and change in long-axis dimension. With natural anatomic landmarks, \(^1,^2,^4\) applied markers, \(^3,^4\) dimension gauges, \(^5\) or dynamic spacial reconstructor, \(^6\) global LV in-plane motion can be compensated for after acquisition. However, motion perpendicular to the imaging plane results in images from different ventricular levels and therefore cannot be corrected for after acquisition.

The fact that the base makes a substantial excursion toward the apex during systole while the apex remains virtually stationary has been well documented. \(^3,^5,^7-^10\) However, current short-axis planar imaging methods such as nuclear magnetic resonance (NMR), echocardiography, or cine computed tomography (CT) acquire data with an externally fixed orientation and do not account for motion along the long axis (through-plane). This misregistration results in the comparison of nonequivalent loci for regional function analysis.

Gated magnetic resonance imaging (MRI) offers excellent spatial resolution. Combined with its ability to acquire images in any plane, it has the potential for providing sensitive and specific noninvasive measurements of LV regional wall motion and thickening. \(^1,^11,^12\) MRI data are not generated in a continuous stream but rather occur only briefly after tissue irradiation with radio frequency (RF) energy. In cardiac imaging, this event is repeatedly synchronized to a specific point of the cardiac cycle to form an image. For a short period of time, the tissue protons retain a memory of the RF perturbation. This effect can be exploited by selective application of RF energy to visually mark or tag a piece of the myocardium. Such a technique was described by Zerhouni et al\(^13\) as a noninvasive method of measuring myocardial motion and deformation at different points along the cardiac cycle. Tags are regions where for a period of time the tissue signal has been reduced or eliminated. By eliminating the signal from entire planes of the left ventricle, specific myocardial regions can be isolated in a way that is insensitive to the previously mentioned through-plane motion.

The purposes of the present study were to 1) precisely quantify LV systolic long-axis motion, relative to a fixed external point, at three levels of the long axis in normal volunteers using noninvasive magnetic resonance tissue tagging; and 2) based on these results, to develop a new noninvasive acquisition technique permitting isolation of the same myocardial segment at both end diastole and end systole regardless of the amount of long-axis motion and without signal contamination from adjacent levels.

**Methods**

**Tissue-Tagging Technique**

Imaging was performed on a Resonex 0.38-T iron core resistive magnet. Acquisition included a spin-echo sequence with a time to echo (TE) of 30 msec and a repetition interval (TR) equal to the RR interval (RR). Four averages were performed over 128 phase and frequency levels. The minimum repetition interval was 56 msec for images and 12 msec for tags.

The specific details of tissue tagging have been described previously. \(^13\) Briefly, tissue tagging involves the selective saturation (tagging) of a plane orthogonal to the final image plane before the standard spin-echo sequence. If the imaging sequence occurs before the tagged region returns to equilibrium, the difference in magnetization will result in a difference in signal between the tagged and nontagged areas. Tagging requires modification only to the acquisition software. Tag planes are created in the same way as image planes but receive a single RF saturation or inversion pulse. To achieve the greatest flexibility, we select and position tag planes independent of standard image planes. The presence of tagging software on a system does not alter equipment performance. The basic capability of tagging and tissue isolation exists on all NMR systems; one only needs to consider different longitudinal relaxation times at various field strengths.

Tags move with the tissue into which they are placed and allow quantification of motion of specific areas of myocardium. The degree of signal reduction in the tag plane is dependent on the relation of the saturation flip angle to the longitudinal relaxation time (T1) of the tissue of interest, the experimental TR, and the delay between the tag and image sequence. In gated cardiac images, tags can be visualized for a period approximately equal to the tissue T1. The specific length of time depends on factors such as the amount of motion and blood flow occurring in the material being tagged as well as the relation of TR to T1. Assuming a negligible effect of TE, an approximation of the signal (S) in the tagged region is given by the following equation:

\[
S_{tag} = M[1 + (\cos \Phi - 1) \times \exp(-TD/T1) - \cos \Phi \exp(-TR/T1)]
\]  

(1)

where \(M\) is proton density, \(TD\) is delay between saturation pulse and imaging sequence, \(TR\) is repetition time between spin-echo sequences, and \(\Phi\) is the tip angle of the saturation pulse.

**Measurement of Long-Axis Motion**

To measure systolic motion, three parallel, equally spaced short-axis tags 3.5 mm wide were applied immediately and simultaneously after the electrocardiogram R wave using a three-lobe RF pulse. A 10-mm-thick systolic image was acquired separately in orientations equal to the standard echocardiographic two- and four-chamber views (Figure 1). Diastolic timing was derived directly from the electrocardiogram R wave; end systole was defined as the first high-frequency component of the second heart sound. A phonocardiogram was recorded using a nonferrous piezoelectric crystal attached to a strip-chart recorder operating at a paper speed of 100 mm/sec. During acquisition, the 90° and 180° RF
ECG

PHONO

PULSE

SEQUENCE

FIGURE 1. Acquisition timing of tagged, end-systolic image. Three equally spaced parallel short-axis tags are placed simultaneously using R wave synchronization. Temporal position of two- and four-chamber images is determined by the second heart sound. $S_1$, first heart sound; $S_2$, second heart sound. ECG, electrocardiogram.

pulses produced discrete artifacts on the phonocardiographic tracing and were used to verify the proper timing of systolic imaging. To ensure that the long-axis images included the center of the apex, the two- and four-chamber orientations were positioned using a group of five parallel short-axis images encompassing the entire left ventricle. Using on-line software, the center of the LV chamber was marked using a cursor for the most basal slice. After saving the coordinates of this point, the apical slice was displayed and similarly marked. The long axis was defined as the line connecting these two points. Displaying a midlevel slice and placing a third point outside the left ventricle would convert the line into a plane forced to pass through the center of the left ventricle. The third point was used to derive the spatial coordinates for the two- and four-chamber imaging planes. Each imaging plane was combined with the three short-axis tag planes for long-axis motion analysis. Systolic images therefore displayed the ventricular wall intersected by the three parallel tag lines, as shown in Figure 2. In addition, adjacent tissue such as the chest wall and spine showed the original diastolic position of the tags. Respiratory motion of the anterior chest wall has the effect of producing image artifacts seen as smearing or copies of the primary image along the phase encoding direction. The magnitude and position of these effects are dependent on the regularity, frequency, and displacement of respiration. The primary image accurately represents the anterior chest wall during its stationary, end-expiratory phase. Tags placed at end diastole can be well seen in both anterior and inferior extracardiac structures. Using on-line software, a line was placed between the anterior and posterior diastolic tag positions (Figure 2). The perpendicular distance from the diastolic reference line to the center of the systolic tag position in the midmyocardium of both walls of the ventricle was determined. This was repeated for the three levels

FIGURE 2. Systolic image of a human left ventricle in the two-chamber orientation. Diastolic position of midlevel tag is highlighted by line a. Systolic displacement of the tag toward the apex is indicated by line b.
and two orientations and represents the systolic displacement of each of the three short-axis levels of the heart with respect to a fixed external structure or imaging plane.

A total of 19 normal volunteers were imaged. There were 14 men and five women (age range, 26–83 years; mean±SD age, 56.1±17.0 years). All participants had been screened previously for the presence of hypertension or coronary disease by stress electrocardiogram and 201Tl scintigraphy. Informed consent was obtained in all cases.

**Tissue Isolation Technique**

By suppressing the signal from above and below a short-axis slice of interest (SOI) at end diastole followed by imaging a thick plane at end systole, the SOI is automatically captured without including signal from the adjacent levels (Figure 3). At end diastole, the SOI was “sandwiched” between two 2-cm-wide parallel planes of saturation. The systolic image was acquired using a 3-cm-thick slice that encompassed the original SOI and voided areas above and below, distributed in proportion to the amount of long-axis motion. Knowledge of the exact amount of shift resulting from long-axis shortening was not required.

When using tags to mark myocardium to be followed for motion analysis, even small differences in signal content between tag and image regions are sufficient to allow measurement of motion or distortion. However, tissue isolation requires maximal possible signal suppression at the time of imaging in that the final image will be a sum of signal across the 3-cm plane. Equation 1 showed that when the TI, TR, RR, and time between end diastole and end systole (R-S2) are fixed, the saturation flip angle alone determines the point of maximal signal suppression. By setting \( S_{sat} \) to equal zero (Equation 2), one can solve for \( \Phi \) :

\[
\text{Tip angle (\( \Phi \))} = \arccos \left( \frac{\exp(-D/T1) - 1}{\exp(-D/T1) - \exp(-TR/T1)} \right) \tag{2}
\]

where D is delay between application of diastolic tag and systolic image (T2 short relative to TR and D).

This angle (\( \Phi \)) is determined for every acquisition to achieve maximal suppression of signal in the saturation planes. In the case of long R-S2 intervals, full signal suppression may require synchronization to every other R wave. In the case of very long R-S2 intervals (more than 350 msec at 0.38 T), full saturation may not be possible. Most commercial imaging systems operate at field strengths of more than 0.38 T. In that T1 is proportional to field strength, full saturation could be achieved for longer R-S2 intervals when using systems with a stronger magnetic (B0) field. The basic principle of tissue tagging and slice isolation is independent of basic system software and individual field strength.

**Phantom validation of tissue isolation technique.** A reciprocating moving phantom was constructed to represent the left ventricle to illustrate the effects of long-axis motion on the measurement of LV phantom size and to test the method of tissue isolation. A phantom (Figure 4) was constructed from a 9.9×9.5×15.0-cm Plexiglas block. A hollow area was machined and filled with a signal-producing gel. The gel was in the form of a stylized left ventricle with a 6-cm-long cylindrical section having an outside diameter of 5 cm and a 4-cm-long conical section. The TI of the gel was estimated to be 340 msec using the method reported by McVeigh et al. The “wall” thickness was a constant 1 cm. To move the phantom along its long axis, a reciprocating electric motor was attached to the phantom by a solid nylon shaft. Both excursion distance and frequency could be adjusted. Frequency was adjustable from 0.0 to 0.83 Hz. Velocity was sinusoidal. Synchronization between the imager and phantom was accomplished by means of an adjustable trigger that produced a gateable signal at any point along the phantom’s cycle. Phantom validation studies were performed 1) to

**Table 1. Planarity or In-Plane Distortion as a Function of Slice Thickness and Radio Frequency Pulse Bandwidth**

<table>
<thead>
<tr>
<th>Slice thickness (mm)</th>
<th>FWHM*</th>
<th>Bandwidth†</th>
<th>Distortion (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td>3.5</td>
<td>3.4±0.3</td>
<td>0.625</td>
<td>0.95</td>
</tr>
<tr>
<td>10.0</td>
<td>9.6±0.5</td>
<td>0.625</td>
<td>1.40</td>
</tr>
<tr>
<td>10.0</td>
<td>9.5±0.3</td>
<td>1.250</td>
<td>1.40</td>
</tr>
<tr>
<td>20.0</td>
<td>17.8±0.9</td>
<td>0.625</td>
<td>1.90</td>
</tr>
<tr>
<td>20.0</td>
<td>19.3±0.3</td>
<td>1.25</td>
<td>1.50</td>
</tr>
<tr>
<td>30.0</td>
<td>25.7±0.2</td>
<td>0.625</td>
<td>2.40</td>
</tr>
<tr>
<td>30.0</td>
<td>28.4±0.4</td>
<td>1.25</td>
<td>1.90</td>
</tr>
</tbody>
</table>

Percent distortion was defined as the maximum through-plane curvature in tag plane/measurement distance. Measurement distance, field of view, and readout gradient were fixed at 26.3 cm, 32 cm, and 0.15 gauss·cm, respectively.

* Full width at half of the maximum signal intensity.
† Bandwidth reported relative to 0.5 gauss.

n=5.
confirm that the method could isolate a given cross section of the phantom independent of its systolic long-axis motion (in the presence of physiological ventricular velocities, images of the conical section of the phantom were recorded at various times in its cycle, and the diameters were compared) and 2) to determine whether the use of “thick” image and tag planes introduced significant geometric image distortion (a static phantom was imaged with tags of increasing thickness perpendicular to the image plane; the tags were then analyzed for proper thickness and planarity).

Standard spin-echo imaging with and without tissue isolation was performed in the phantom at a frequency of 0.83 Hz. Within the context of the phantom study, end diastole was defined as the point of maximum retraction of the phantom. It was at this point that the gating pulse was produced and saturation pulses were applied. Delays from the trigger of 0.0, 0.161, 0.235, 0.300, and 0.335 msec provided long-axis displacements of 0.0, 0.5, 1.0, 1.5, and 2.0 cm, respectively. Corresponding velocities at these points were 0.0, 5.84, 7.40, 7.85, and 7.40 cm·sec. The acquisition plane was positioned over the conical section of the phantom. Data taken with different long-axis displacement would thereby result in short-axis images having different diameters. Figure 5 displays the effect of systolic long-axis motion on the resulting short-axis phantom images. Conventional acquisition results in simulated end-systolic and end-diastolic images having different diameters as a result of noncompensated motion during the imaging sequence. Acquisition incorporating saturation pulses at end diastole effectively excludes signal from tissue adjacent to the SOI, allowing a wide image acquisition plane to capture an end-systolic image that represents signal from only the area defined at end diastole.

Slice thickness is determined by the relation of the RF pulse band width to the slice-selective gradient slope. In general, the bandwidth is fixed, with thicker slices produced by decreasing the slice-selective gradient slope. As this slope decreases, it may be overpowered by static field inhomogeneities and result in geometric distortion of the slice. Planarity or through-plane slice distortion was evaluated for plane thicknesses of 3.5, 10, 20, and 30 mm. A cylinder phantom (35x15 cm) was used to acquire standard images that were intersected by horizontal tags perpendicular to the image plane. Using system software, a thin, straight horizontal line was superimposed on one edge of the tag. Through-plane distortion was seen as a curvature of the tag as one moved from the center of the phantom. Images were enlarged so that individual pixels were visible, and the divergence of tag from the straight line was evaluated for each horizontal pixel over a 26.3-cm range using a 32-cm acquisition field of view. The maximum deviation (in pixels) that the tag plane made from a straight line was related to the length of

![Figure 4. Schematic of left ventricular phantom displaying the conical region used to validate tissue isolation. Diameter of short-axis slices taken in this region are directly related to their long-axis position.](image)

### Table 2. Site-Specific Systolic Long-Axis Displacement

<table>
<thead>
<tr>
<th>Level</th>
<th>Two-chamber view</th>
<th>Four-chamber view</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Anterior</td>
<td>Inferior</td>
</tr>
<tr>
<td>Base</td>
<td>12.8±4.5*</td>
<td>15.2±3.5</td>
</tr>
<tr>
<td>Mid</td>
<td>6.8±2.1</td>
<td>8.7±2.7</td>
</tr>
<tr>
<td>Apex</td>
<td>1.7±2.1</td>
<td>3.0±2.3</td>
</tr>
</tbody>
</table>

Given is total displacement occurring along the left ventricular long axis from end diastole to end systole for each of three levels (p<0.0001) in the two walls (p<0.01) as seen from the two- and four-chamber orientations (p<0.01).

*Motion expressed in mean±SD millimeters.

n=19.
the plane (also in pixels) and was expressed as percent distortion. Measurements were performed with both standard RF pulses and with those used in the tissue isolation technique, which use twice the standard spectral width, thereby allowing increased gradient slope and less geometric distortion. The readout gradient was held constant at 0.15 g/cm. The full width at half-maximum signal intensity was calculated from an image intensity plot across each tag region.

Tissue isolation in humans. The true LV short-axis plane was determined by an axial and long-axis "scout" image. The volunteer’s heart rate and S1-S2 interval were taken from a phonocardiogram immediately before final image acquisition. Using this information, the appropriate saturation flip angle was calculated. Imaging of a basal LV level was first performed at end diastole and end systole using a standard 30-msec gated spin-echo sequence. This was followed immediately by acquisition of an end-systolic image, including the tissue isolation technique. As with the phantom, the SOI was sandwiched between two 2-cm-thick saturation planes at end diastole, and after the appropriate delay to end systole, a 3-cm-thick short-axis image was acquired. In seven of the volunteers, the TR equaled 2 multiplied by RR; in all other cases, TR equaled RR. Images were acquired with 128 phase and frequency encoding steps with four averages and a 32-cm field of view.

Statistical Analysis

All values are reported as mean±SD. Differences and interactions between levels, walls, and views were tested by a three-way analysis of variance.

In the presence of significant differences within levels, a Newman-Keuls test was applied to compare individual levels.

Results

Long-Axis Motion

Table 2 presents long-axis motion relative to a fixed external point as a function of ventricular level, wall, and view. Differences in ventricular displacement were pronounced between levels (p<0.0001), ranging from 12.8±3.8 mm at the base to only 1.6±2.2 mm at the apex. Both the two- and four-chamber views produced significantly different degrees of long-axis motion between opposite walls (p=0.008). Figure 6 displays the amount of ventricular shift found in the 19 normal volunteers measured at three levels in the two- and four-chamber views. In all cases, motion was from the base to the apex. It can be seen in Figure 6 that the walls in the two-chamber view displayed greater displacement at all levels than those in the four-chamber view (p=0.006). Basal systolic displacement was greatest in the inferior wall, moving 15.2±3.5 mm, followed by the lateral, anterior, and septal walls, with systolic displacements of 13.3±3.7, 12.8±4.5, and 10.0±2.5 mm, respectively (p<0.01).

Phantom Validation Studies

The solid line in Figure 7 shows the theoretical signal-suppressing ability of 180° saturation pulse in a material with a calculated T1 of 541 msec at varying periods before the image is acquired. Open circles represent signals measured in a phantom using a long
TR (1,500 msec). Although maximal signal void occurs at T of approximately 0.693 T1 (where T is delay from 180° RF pulse), contrast between tagged and nontagged regions is present for more than twofold the tissue T1 period. The disparity between the calculated and observed tag signals at very short delays is the result of the normal heterogeneity of tip angles present in an RF pulse. The technique of tissue isolation was validated under controlled conditions simulating different physiological states. Figure 8 summarizes the effect of increasing amounts of systolic shift on proper registration of a moving phantom with and without isolation. In the absence of motion, both acquisition techniques produced the correct systolic image. As the amount of displacement increases, it can be seen that the systolic diameter-to-diastolic diameter ratio of standard imaging decreases progressively because of increasing misregistration, whereas it remains constant when isolation is used. Tissue isolation was tested at phantom velocities up to 7.85 cm sec without image-quality degradation or measurable misregistration.

Variations in the RR interval during acquisition in the presence of a fixed delay and saturation flip angle will result in incomplete signal suppression and therefore compromise tissue isolation. The magnitude of the effect is determined by how often and to what degree the RR interval varies during the study. Figure 9 displays the relative signal in the saturation region versus the observed RR interval for three RR intervals (600, 800, and 1,000 msec) with corresponding S1-S2 intervals of 250, 300, and 350 msec with tissue T1 assumed to be 450 msec. It can be seen that when ectopy results in a shorter-than-calculated RR interval, the signal-suppressing ability of the saturation pulse is reduced. In the case of the 1,000/350 msec RR/S1-S2 interval, 19% of the underlying tissue signal remains in the tag region at the calculated RR

**Figure 6.** Bar graph of degree of systolic long-axis displacement in the two- and four-chamber views for the base, mid, and apical left ventricular levels as measured in 19 normal volunteers by tissue tagging.

**Figure 7.** Plot of calculated tag signal expressed as the ratio of the signal in tagged regions to that of the signal in nontagged regions vs. the interval from tag to image for a 180° saturation pulse. Observed data are represented by open circles.
interval. By doubling the TR (RR), this could be reduced to 9%. In the case of shorter S1-S2 intervals, less signal is present in the tag at the calculated RR interval (14% in the 800/300-msec interval and 12% in the 600/250-msec interval). In both cases, complete signal suppression could be accomplished by gating on every other R wave. In addition to allowing better signal saturation at the expected heart rate, the shorter S1-S2 associated with shorter RR intervals maintains better signal suppression in the presence of shortened intervals resulting from ectopy. Tissue magnetization takes more than four T1 intervals to return to complete equilibrium; as the ratio between S1-S2 and RR interval increases, less time is available before the next pulse for complete relaxation.

**FIGURE 8.** Plot of effect of increasing amounts of through-plane motion (offset) occurring between the acquisition of simulated end-diastolic (ED) and end-systolic (ES) images in a conical phantom. As displacement increases, conventional magnetic resonance produces images from different levels of the cone. This results in a progressive decrease in the ED diameter-to-ES diameter ratio. The tissue-isolation technique tracks the same section of the cone phantom from end diastole to end systole, resulting in a diameter ratio that approaches unity.

**FIGURE 9.** Plot of effect of variations in heart rate during the acquisition period on signal saturation ability displayed for three typical RR/S1-S2 intervals. Calculated relative signal in the tag region is plotted against RR interval. As the RR interval is reduced from the calculated value in the presence of a fixed S1-S2 interval, there is a reduction in signal suppression. Within the physiological range of heart rates, shorter S1-S2 intervals allow better signal saturation and would therefore be less affected by changes in heart rate.
The results of tag planarity and thickness evaluation are presented in Table 1. Thin tags (3.5 mm) displayed less than 1.0% distortion over 26.3 cm. There was a gradual increase in through-plane distortion as tag thickness was increased. By using a wider bandwidth, 30.0-cm tags, the thickest used in this study, displayed only 1.9% distortion. Therefore, both phantom and human saturation tags were produced using the 1.25 spectral width.

Isolation of Normal Human Myocardium

Tissue isolation was applied in the short-axis plane at the base of the left ventricle in 15 normal volunteers. A selected example is shown in Figure 10. The top left panel displays a standard end-diastolic image; the top right panel displays the same position at end systole without tissue isolation; and the bottom left panel displays the same plane at end systole but acquired using the tissue isolation technique. In both examples, the end-diastolic image shows distinct and complete right ventricular (RV) and LV walls without inclusion of atria or great vessels. The second frame shows the effect of systolic motion. Without isolation, the systolic image includes portions of the left atrium and ascending aorta. In some cases, the septal segment of the wall is absent or markedly thinned. In the isolated ventricular slice in the bottom left panel, both left and right ventricles are clearly seen and have thickened circumferentially. The atria and aorta are absent. A papillary muscle is prominently seen in both the end-diastolic and isolated end-systolic images, verifying that similar levels have been imaged at the two cardiac phases.

Discussion

In the present study, systolic long-axis displacement at three separate levels of the left ventricle was...
quantified noninvasively in a group of 19 normal volunteers using a selective RF saturation technique known as tissue tagging. The base of the left ventricle moves nearly 13 mm during systole. The mid and apical levels show progressively less movement, displacing 6.9 and 1.6 mm, respectively. These findings underscore the need for a method of tissue isolation that will allow the identical myocardium to be visualized in both systole and diastole, particularly near the base where systolic displacement is greatest.

Previous studies of measuring contraction dynamics showed similar results for basal motion but could not describe motion elsewhere in the left ventricle because of a lack of identifiable landmarks. Zaky et al used echocardiography to follow the motion of the mitral ring throughout the cycle. They showed a 16.0±4.0-mm systolic displacement. In addition, by having the ability to make continuous measurements, they verified that the maximum displacement occurred at end systole and was well centered about the second heard sound. Mitral ring movement was also estimated using angiography by Dayem et al to be approximately 14 mm. Long-axis motion was determined by combining data taken from anterior and lateral projections. Epicardial and intramyocardial markers provide excellent reference points but are not applicable to normal human studies. Furthermore, the invasiveness of the procedure could alter the normal functional state.

Both magnetic resonance and ultrafast cine computed tomography provide exceptional spatial resolution. Gated short-axis slices 10 mm or less can clearly visualize structures such as the papillary muscles and larger trabeculations that may pass into or out of the imaging plane during cardiac contraction. This is another reason why slices must be properly matched throughout the cardiac cycle; the magnitude of functional heterogeneity may in part result from analysis of misregistered LV sections. The present study has shown that different positions around the short axis display differences in systolic shift, which at the base ranges from 15.2 mm in the inferior wall to 10.0 mm in the septum. The tissue isolation technique is ideally suited to this situation. Because no assumptions in regard to regional motion are made, this technique is able to capture the same myocardium between diastole and systole regardless of average as well as site-specific systolic shift. This would be especially valuable in the presence of ischemia or infarction, where variations in regional motion are exaggerated.

Limitations

Table 2 revealed that a basal LV slice does not move uniformly toward the apex at systole. Because of this, the short-axis plane, which was flat at end diastole, takes on a warped and tilted posture later in the cardiac cycle. Although the present technique captures the systolic plane regardless of nonuniformity, such distortion would result in an error in the estimation of wall thickness because of foreshortening. The magnitude of this error can be estimated by taking the cosine of the “tilt” angle; this would be the ratio of the observed wall thickness (foreshortened) to the actual wall thickness. In the case of a tilt of 20°, this would result in an error of 6%. The myocardial slice acquired by standard acquisition methods would by definition have parallel walls and not be affected by this error. However, the myocardium contained within the diastolic and systolic images would not be the same.

Because MRI requires a prolonged period of acquisition, patient movement may result in a degraded image. In the case of tissue saturation, variation in the RR interval results in incomplete signal suppression in the tag regions because the saturation flip angle is fixed before each image. The saturation pulse must also be prescribed for a specific tissue's T1 relaxation. The signal within tissues having different T1 values will not be as fully voided and will be added to the final image. Fat is an example of a tissue with a T1 much shorter than heart tissue that is included from within any region of the 3-cm image slice.

In that the presented technique was used on a low field resistive magnet, the thick slices used in tissue isolation were sensitive to geometric distortion. Use of spectrally wider RF pulses were combined with steeper slice-selective gradients to minimize this effect. As development of more “uniform” resistive magnets is achieved, the issue of slice geometry will become less of an issue. Presently, MRI is the only imaging modality to make use of the dynamic nature of tissue on a nuclear level. The ability to use the property of spin relaxation provides a simple way to both spatially and temporally choose the tissue to be seen as well as the tissue to be eliminated.

To determine LV function, images acquired at end diastole and end systole must be compared. Present standard MRI allows acquisition of either multiple time points at a fixed location (gradient echo sequences) or multiple time points and different ventricular levels using conventional spin-echo sequences. Either technique requires the number of scans to equal the number of LV levels to be analyzed. The present tissue isolation technique requires separate acquisition of the end-diastolic and end-systolic images at each LV level. Acquisition of a standard series of three LV short-axis levels requires 45 minutes, including scout images. The addition of three isolated systolic images would add an additional 15 minutes. This would make acquisition of more than three isolated levels or evaluation of temporal heterogeneity impractical until improvements allowing either faster acquisition or multiphase isolation are achieved.

In general, the isolation technique makes greater demands on the imaging system than standard image formation and requires a stable imaging period. As a result, although images are spatially correct, image quality is usually reduced compared with nonisolated images.
In summary, this study reports systolic long-axis motion of the LV base large enough to move a short-axis slice completely out of the acquisition plane in normal subjects. A slice-isolation technique is described that automatically images the identical short-axis section of myocardium between diastole and systole, even in the presence of regionally non-uniform long-axis shift. Use of this technique should therefore result in improved accuracy of analysis of regional LV function.

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KEY WORDS • magnetic resonance imaging • systolic motion • radio frequency
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Circulation. 1991;84:721-731
doi: 10.1161/01.CIR.84.2.721

Circulation is published by the American Heart Association, 7272 Greenville Avenue, Dallas, TX 75231
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Print ISSN: 0009-7322. Online ISSN: 1524-4539

The online version of this article, along with updated information and services, is located on the World Wide Web at:
http://circ.ahajournals.org/content/84/2/721

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