Two-dimensional Left Ventricular Deformation During Systole Using Magnetic Resonance Imaging With Spatial Modulation of Magnetization

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Background Myocardial tissue tagging with the use of magnetic resonance imaging allows noninvasive regional analysis of heart wall motion and deformation. However, any evaluation of the effect of disease or treatment requires a baseline reference of normal values and variation. We studied the two-dimensional motion of material points imaged within the left ventricular wall using spatial modulation of magnetization (SPAMM) in 12 normal human volunteers.

Methods and Results Five parallel short-axis and five parallel long-axis slices were acquired at five times during systole. SPAMM tags were generated at end diastole using a 7-mm grid. Intersection point data were analyzed for displacement, rotation, and torsion, and triangles of points were analyzed for local rotation and principal strains. Short-axis displacement was the least in the septum for all longitudinal levels (P<.001). Torsion about the long axis was uniform circumferentially because of the motion of the centroids used to reference the rotation. In the long-axis images, the base displaced longitudinally toward the apex, with the posterior wall moving farther than the anterior wall (13.4±2.2 versus 9.7±1.8 mm, P<.001) in this direction. The largest principal strain (maximum lengthening) was approximately radially oriented in both views. In the short-axis images, the minimum principal strain (maximum shortening) increased in magnitude toward the apex (P<.001) with little circumferential variation, except at midventricle, where the anterior wall showed greater contraction than the posterior wall (−0.21±0.03 versus −0.19±0.02, P<.02).

Conclusions Consistent regional variations in deformation are seen in the normal human heart. Displacement and maximum shortening strains are well characterized with two-dimensional magnetic resonance tagging; however, higher-resolution images will be required to study transmural variations. (Circulation. 1994;89:740-752.)

Key Word • magnetic resonance imaging

A precise quantitative description of regional wall motion and deformation is required to evaluate the extent of disease and model cardiac mechanics. To provide this information, a number of accurately identifiable landmarks must be located and followed through time. Landmarks have been supplied in the past by implanted radiopaque beads or ultrasonic crystals or by the bifurcations of the coronary arteries. These techniques are limited in the case of implanted markers by the invasiveness of the implantation and the small regions covered and in the case of natural landmarks by their sparse distribution. The use of magnetic tagging with magnetic resonance imaging (MRI) makes possible the noninvasive evaluation of regional heart wall motion. With the spatial modulation of magnetization (SPAMM) technique, a large number of trackable material landmarks can be created noninvasively throughout the heart wall in a short time interval by a sequence of nonselective radio frequency pulses separated by magnetic field gradients. Typically, two orthogonal families of parallel planes of magnetic saturation are created, each orthogonal to the image plane. The intersections of these tagging sheets with the imaging plane form stripes in the image that can be tracked throughout systole; the intersections of the tagging stripes with each other in the image serve as material markers. Small groups of markers (eg, marker triangles) can be analyzed with methods previously applied to implanted radiopaque beads and ultrasonic crystals. If we assume constant deformation within the triangle, the relative motions of the vertices can be decomposed into local rigid body motion (translation and rotation) and local deformation (strain).

To evaluate the extent of disease or the effect of treatment, a baseline reference of normal values and variation must be available. We hypothesize that such a database would contain systematic variations in regional wall motion and deformation in the normal heart and that these variations can be accurately assessed with MR tagging techniques. To build an initial database, we studied the two-dimensional deformations measured using SPAMM in 12 normal human volunteers. Current problems encountered are discussed, and possible directions for improvement are proposed.

Methods

Image Acquisition

Twelve normal human volunteers (age range, 19 to 37 years; 6 men) were imaged with a 1.5-T superconducting
magnet (Signa, GE). An initial scout series of coronal spin-echo images was obtained to identify cardiac landmarks. A long axis (LA) of the left ventricle (LV) was chosen to pass through a point at the left of the aortic root and the most anterior left point on the LV apex, as seen in the coronal images. End diastole (ED) was determined by detection of the rising R wave of the ECG. A set of rapid gradient-echo transverse images was acquired at midventricle for timing. End systole (ES) was defined as the instant of least cavity area in the midventricle. Multislice, multiphase SPAMM tagged images were then obtained in a short-axis view perpendicular to the LA in five locations at five times spaced uniformly throughout systole, starting 13 milliseconds after detection of the R wave and triggered prospectively. Time 1 refers to ED, time 2 to early systole, and time 5 to ES. Slices were 5 mm thick, had to 15 mm between slice centers, and had a 24-cm field of view with a 256×128 image matrix (interpolated to 256×256). These images used a 130° flip angle, TE (echo time) of 27 milliseconds, flow compensation, and two signals averaged. The SPAMM tagging pulse sequence was performed at detection of the R wave and generated a tagging grid with 7-mm stripe spacing at an approximate thickness (Fig 1). A similar series of tagged images was then obtained parallel to the LA and approximately perpendicular to the interventricular septum. Each of the short-axis (SA) and LA image series required about 20 minutes to acquire, depending on the heart rate.

Conventional echocardiograms were also performed on each subject to evaluate the contractile state and verify the normal cardiac function of the study population. Two-dimensional echocardiograms were qualitatively normal in each case. Quantitative results of the two-dimensional guided M-mode echocardiograms are summarized in Table 1. None of the subjects had evidence or prior history of cardiac disease, and LV mass index, end-diastolic dimension, end-systolic dimension, wall thickening, and fractional shortening were within normal limits.

Image Analysis

Images were analyzed using a custom-written interactive display and analysis package running on standard graphics workstations. The inner and outer contours of the LV were traced manually on each image so as to enclose the LV free wall and septum (Fig 1). In the ED image, the location of the inner contour could not be determined because of signal from blood (before the tags in the cavity are washed away), so the contour was inferred from the subsequent image, relative to the tags in the surrounding muscle. Tag stripes within the muscle were located and tracked from ED to ES using an automated tracking procedure based on an active contour model. In this procedure, each stripe was modeled as a thin, flexible beam with a small inherent resistance to stretch and bending. Stripes were joined at the intersection points, and the entire mesh was deformed to minimize the corresponding intensity values in the image. The inherent continuity of the structure gave the model some immunity to image noise. Bilinear interpolation between pixels enabled subpixel resolution in stripe tracking. Stripe intersections were then extracted from the tracked mesh. Right ventricular inner contours were also defined at ED to aid in registration. In addition, the locations of the LV apex and the center of the mitral/aortic valve ring at the base were chosen manually on the most central LA image at ED. These were subsequently used to register data on a regional basis, as follows.

Data Registration

To register data between hearts, a fixed cylindrical polar reference frame was defined using anatomic reference points defined in the ED images. A central axis of the LV was defined by connecting the centroids of the most basal and most apical SA outer contours. Apex and base markers, as defined on the most central LA image, were projected onto this axis (Fig 2, top). The longitudinal coordinate (z) of any SA or LA tag point p was determined by the length between the point's projection onto the central axis and the apex projection, expressed as a proportion of the apex-base projection distance (ie, z=0 for the apex and z=1 for the base; Fig 2, top). The circumferential coordinate (θ) was given by the angle around the central axis from the centroid of the SA right ventricular contours to the point p (antidirectionally positive as seen looking from the apex to the base; Fig 2, bottom). The radial coordinate (r) of each SA point was defined as the distance from the point to the inner contour along the line joining the point and the centroid of the outer contour, divided by the distance between the outer and inner contours along the same line (Fig 2, bottom). This provided a normalized estimate of the position of the point within the wall: 0 for endocardium and 1 for epicardium.

These coordinates were divided into regions for comparison, as follows. Three longitudinal regions were defined as base (z>0.55), midventricle (0.35<z<0.55), and apex (z<0.35). The most basal “third” was chosen to be somewhat more than one third of the distance from base to apex at ED to allow for the motion of the base toward the apex during systole. Circumferential regions were divided equally (Fig 2, bottom) into septal (7π/4<θ≤π/4), posterior (π/4<θ≤3π/4), lateral (3π/4<θ≤5π/4), and anterior (5π/4<θ≤7π/4). Radial regions were also divided equally into endocardial (r≤1/3), midwall (1/3<r≤2/3), and epicardial (r>2/3). Data in each region were averaged to give a single value for each heart and then pooled for all 12 hearts according to region. Data from the papillary muscles were excluded from the analysis.

Displacement and Torsion

Each tag intersection point's displacement D from the position at ED was calculated for each subsequent time. Rotation (θ in Fig 3) about the LA was calculated for the SA image points as the angle between the radial line joining the point and the centroid of the outer contour at time 1 and the same line at each subsequent time. Thus, the rotation was about the moving centroid for each slice. A displacement direction (y) was also calculated as the angle between the displacement vector and the radial line at ED (Fig 3). Torsion in the SA images was calculated after the points were averaged into regions and was defined as the average rotation of a region relative to the corresponding region at the base (antidirectionally positive looking from the apex). For all LA images, the displacement was calculated in the longitudinal direction only, i.e., parallel to the central axis at ED.

Homogeneous Strain Analysis

As described previously, tagged intersection points were used to create a triangular tiling of the heart wall. Where possible, the points were triangulated so as to create the greatest number of triangles across the wall. The deformation within each triangle relative to the configuration at ED was assumed to be locally homogeneous, characterized by the deformation gradient tensor F. This quantity relates any material line segment dX of the undeformed state to the corresponding deformed line segment dx:

\[
\begin{align*}
dx &= F \, dX \\
F &= R \cdot U \\
E &= 0.5(U \cdot U - I)
\end{align*}
\]

where R describes the rotation of the triangle about its centroid, and U (the right stretch tensor) and E (Green's strain tensor) describe the state of strain within the triangle. The eigenvalues of E (λ₁ and λ₂, with λ₁≥λ₂) are the principal strains (λ₁>0 for extension and λ₁<0 for contraction), corresponding to the maximal extensions and contractions experienced by the triangle. The associated eigenvectors e₁ and e₂ are
Fig 1. This and facing page. Spatial modulation of magnetization images at midventricle with left ventricular inner and outer contours (lines) and automatically tracked intersections (squares). Top left, Short-axis time 1 (end diastole). Bottom left, Short-axis time 5 (end systole). Top right, Long-axis time 1 (end diastole). Bottom right, Long-axis time 5 (end systole).

the directions in which they act. The principal angle \( \beta \) was defined by the orientation of \( e_1 \) relative to the radial direction (joining the triangle centroid with the centroid of the outer contour for that slice). Deformation parameters examined were the angle of rotation \( \alpha \) associated with \( R \) (anticlockwise positive looking from the apex), the eigenvalues \( \lambda_1 \) and \( \lambda_2 \), the principal angle \( \beta \), and the relative change in area of the triangle \( \Delta A \). For triangle deformation parameters in the LA images, only the most central slices were analyzed because these were approximately normal to the septal and lateral walls and were easier to interpret than the other, more tangential, LA slices.

Reproducibility and Statistical Analysis

The reproducibility and interexamination variation of the deformation parameters obtained were investigated by comparing two SA image series of 25 images each (five locations imaged five times) taken sequentially for one subject. Comparisons were made between manual and automatic point tracking methods, and errors were calculated between images of corresponding location and time.

For the 12 normal subjects, kinematic parameters were compared at multiple regions within the LV wall using repeated-measures ANOVA. When testing for circumferential or longitudinal variation, the radial regions were pooled.
If a significant difference was found, the Scheffé test was applied to compare individual regions.

**Results**

**Reproducibility**

Tag intersection points were tracked in two SA images series (25 images each) of the same subject both manually (by an experienced observer) and by the automatic procedure. At each slice location, corresponding points and triangles were found by matching $x$ and $y$ image coordinates to within 1 mm on the initial (ED) image. Differences between displacement, rotation, and strain measures were compared between series (examinations) using both manual and automatic tracking methods. Manual versus automatic comparisons were also made and averaged over the two series. These results are summarized in Table 2. At time 5 (ES), no significant differences were found in any parameter between methods or examinations by ANOVA except for displacement in the second series, where a difference was found between manual and automatic methods. The root-mean-square discrepancies between deformation measures derived from different series were approximately halved in the automatic method compared with manual tracking (Table 2).

**SA Displacement and Torsion**

The SA image points displayed clear regional variations in displacement and rotation. Fig 4 shows the loci of tag points from ED to ES in apex, mid, and base
TABLE 1. Summary of Two-dimensional Guided M-Mode Echocardiographic Data

<table>
<thead>
<tr>
<th>Subject</th>
<th>Sex</th>
<th>BSA (m²)</th>
<th>LVMI</th>
<th>EDD (cm)</th>
<th>ESD (cm)</th>
<th>TF (%)</th>
<th>FS (%)</th>
<th>HR (bpm)</th>
<th>BP (mm Hg)</th>
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<tr>
<td>1</td>
<td>F</td>
<td>1.68</td>
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<td>5.0</td>
<td>2.7</td>
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<td>49</td>
<td>120/74</td>
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<td>2</td>
<td>M</td>
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<td>5.6</td>
<td>3.7</td>
<td>63</td>
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<td>63</td>
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<td>F</td>
<td>1.64</td>
<td>106.7</td>
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<td>39</td>
<td>89</td>
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<td>80.0</td>
<td>5.0</td>
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<td>33</td>
<td>75</td>
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<td>3.1</td>
<td>71</td>
<td>39</td>
<td>65</td>
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<tr>
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<td>16.2</td>
<td>0.4</td>
<td>0.4</td>
<td>12</td>
<td>6</td>
<td>11</td>
<td></td>
</tr>
</tbody>
</table>

BSA indicates body surface area (m²); LVMI, left ventricle mass index (g/m²); EDD, end-diastolic dimension (cm); ESD, end-systolic dimension (cm); TF, wall thickening fraction (%); FS, fractional shortening (%); HR, heart rate (beats per minute); and BP, blood pressure (mm Hg).

levels from ED to ES in a typical subject, looking from the apex. The contours are drawn at ED for reference. A number of qualitative trends can be seen. There is a larger displacement of the lateral wall compared with the septum at all levels. At the apex, the posterior wall moves more than the anterior, whereas at the base, the anterior displaces more than the posterior. The septum in all levels appears to rotate anticlockwise initially, becoming more radially directed at subsequent times. There is an overall anticlockwise rotation of the apex and clockwise rotation of the base. However, in anteroseptal regions of the mid and apical levels and the posteroseptal regions of the base, a pronounced hook-like motion is observed because of a reversal of rotation during contraction. In particular, the posterior base initially rotates anticlockwise in early systole but then reverses to a clockwise rotation by ES.

Displacement is plotted in Fig 5 (all graphs of this type show values averaged for each region across 12 hearts with error bars indicating SD). At ES (Fig 5, top), displacement was the least in the septum for all longitudinal levels (P<.001 compared with the lateral...
TABLE 2. Interexamination Errors Between Two Short-Axis Series

<table>
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<th></th>
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<td>0.60</td>
<td>1.00</td>
<td>0.84</td>
</tr>
<tr>
<td>δ, °</td>
<td>1.8</td>
<td>1.7</td>
<td>2.4</td>
<td>1.9</td>
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<tr>
<td>λ₁, °</td>
<td>0.121</td>
<td>0.212</td>
<td>0.228</td>
<td>0.064</td>
</tr>
<tr>
<td>λ₂, °</td>
<td>0.049</td>
<td>0.042</td>
<td>0.064</td>
<td>0.038</td>
</tr>
<tr>
<td>α, °</td>
<td>4.0</td>
<td>3.8</td>
<td>5.4</td>
<td>2.7</td>
</tr>
<tr>
<td>β, °</td>
<td>10.1</td>
<td>8.0</td>
<td>12.37</td>
<td>5.7</td>
</tr>
<tr>
<td>ΔA</td>
<td>0.103</td>
<td>0.144</td>
<td>0.163</td>
<td>0.078</td>
</tr>
</tbody>
</table>

Root-mean-square differences are shown averaged over the five locations for deformation parameters calculated between end diastole and end systole.

wall at each level). At the apex, displacement was greatest in the lateral and posterior regions, whereas the base experienced maximum displacement in the lateral and anterior walls. There was a steady decrease in posterior wall displacement from apex to base (8.0±1.5 to 6.2±1.6 mm, *P* < .001) together with an increase in the anterior wall displacement from apex to base (5.4±1.9 to 6.8±1.9 mm, *P* < .01). At each circumferential location, endocardial regions displaced more than epicardial regions, as shown in Fig 5, middle (*P* < .02, except the posterior wall [*P* = NS]). In early systole (time 2, Fig 5, bottom), the anterior wall showed the greatest motion and the posterior showed the least at all levels (*P* < .01).

The rotation about the moving centroid (Fig 6, top) increased from base to apex (*P* < .001 between each
Torsion, in the 5 Time

During early systole, the greatest torsion was 4° to 4° higher in endocardial regions than in epicardial regions (Fig 6, middle; P<.01 at each level). Within each level, the lateral and anterior walls rotated the most and the septum rotated the least (being more negative at the base), although this was significant only at midventricle (P<.001). Absolute rotation was 2° to 4° higher in endocardial regions than in epicardial regions (Fig 6, middle; P<.01 at each level). During early systole (Fig 6, bottom), the anterior wall showed the greatest rotation. There was a marked variation in the development of rotation during systole between longitudinal levels (compare Fig 6, top, with Fig 6, bottom). At the apex, more than 50% of the rotation is developed in the first 25% of systole (time 1 to time 2). At the base, all regions initially rotated anticlockwise (time 2) and then clockwise to ES.

Rotation relative to the base (torsion) is plotted at ES in Fig 7. No significant circumferential variation was found at apex or midventricle. Similar to rotation, the torsion was 4° to 6° greater in endocardial regions than in epicardial regions (16.5±2.7° versus 10.3±2.2° at the apex, P<.001; 8.5±3.6° versus 4.2±1.7° at midventricle; P<.001). Torsion, unlike rotation, increased steadily with frame number in all circumferential regions.

Of critical importance in the calculation of rotation and torsion is the motion of the centroid used as the reference axis of rotation. Regions that appear to have markedly different rotations with respect to a fixed centroid (eg, of the outer contour at ED; see Fig 4) have similar rotations with respect to a moving centroid. To quantify the motion of the centroids, we defined a coordinate system in each SA slice with the origin at the centroid of the outer contour at ED, the x axis pointing toward the septum, and the y axis pointing toward the posterior wall. In this fixed coordinate system, the centroid of the most apical slices displaced 0.0±1.3 mm in the x direction and −2.5±1.4 mm in the y direction (ie, in the anterior direction) from ED to ES, whereas the centroid of the most basal slices displaced 0.8±1.3 mm in the x direction and 1.3±1.3 mm in the y direction (ie, in the posterioseptal direction) (P<.01 for y displacement apex versus base). The direction of displacement from ED to ES relative to the ED centroid (in the same fixed coordinate system) is plotted in Fig 8. The septum displaces almost normal to itself at all levels, in contrast with the anterior apex and posterior base, which displace more tangentially. Thus, the motion of the contour centroids serves to equilibrate the apparent torsion around the wall, eg, by increasing the apparent rotation of the apical septum because of the anterior motion of the centroid.

LA Displacement

Fig 9 shows the loci of tag points for the most central LA image plane of a typical study. The motion of the valve plane toward the apex is clearly seen. Fig 10 shows the variation of longitudinal displacement around the ventricle. In all levels, the lateral and posterior walls displaced more than the septal or anterior walls, with maximum in the posterior wall and minimum in the anterior wall (eg, 13.4±2.2 mm at basal posterior versus 9.7±1.7 mm at basal anterior, P<.001). This variation was also present at early systole (time 2; 4.2±1.4 mm in basal posterior versus 3.5±1.1 in basal septum; P<.05), in contrast to the SA displacement, which was greatest in the anterior regions.

SA Local Deformation Analysis

The maximum principal strain λ1 (greatest extension), as seen in the SA images referenced to ED, is shown in

![Figure 6](image1.png)  
**Fig 6.** Bar graphs of short-axis rotation about the centroid. Top, Time 5 (end systole). Middle, Time 5 transmural variation. Bottom, Time 2 (early systole).

![Figure 7](image2.png)  
**Fig 7.** Bar graph of torsion at end systole.

![Figure 8](image3.png)  
**Fig 8.** Bar graph of displacement direction (end systole).
Fig. 9. Plot of loci of tag points in the most central long-axis image from end diastole to end systole in a typical subject. S indicates septal; L, lateral. Contours shown at end diastole for reference.

Fig. 11. At the base, the lateral wall showed greater extensions than the septum or anterior walls (0.29±0.05 versus 0.21±0.08 or 0.21±0.06, respectively; P<.01 for each). Extension was greater at the base than at the apex (0.24±0.07 versus 0.15±0.07, P<.001). The minimum principal strain (greatest contraction) λ2, shown in Fig. 12 for ES, was greater in magnitude at the apex than at the base (−0.22±0.02 versus −0.19±0.03, P<.001). No significant circumferential variation was found except at midventricle, where the anterior wall showed slightly greater contraction than the posterior wall (−0.21±0.03 versus −0.19±0.02, P<.02). The calculation of relative change in area of the triangles (the product of the eigenvalues of U in Equation 1) showed a net loss in area from ED to ES of approximately 15% at the apex, 10% at the midventricle, and 5% at the base.

The direction of greatest principal stretch was aligned approximately radially in all locations (Fig. 13). This is consistent with the hypothesis that the maximum lengthening strain is primarily associated with wall thickening. However, there was a consistent variation from negative angles at the apex to positive angles at the base (referred to the radial direction, P<.001). This deviation from radial orientations is a measure of the circumferential/radial transverse shear caused by endocardial points rotating more about the centroid than the epicardial points. Thus, at the apex, the deviation is positive (anticlockwise—the direction of rotation at the apex), whereas at the base, it is negative (clockwise—the direction of rotation at the base). In addition, apical regions showed greater shear in the septal and anterior walls than in the lateral wall (P<.05). The rigid body rotation of the triangles (α, Fig. 14) increased toward the apex (5.5±2.5°) from the base (0.1±2.1°, P<.001). This is consistent with the anticlockwise rotation of the apex. Values of α are reduced from those of the rotation about the centroid resulting from the presence of transverse shear. At the apex, the lateral wall (6.9±2.6°) showed greater rotation than the septum (4.5±1.5°) or anterior wall (4.2±2.5°) (P<.05), although this might be the result of increased shear in the septal and anterior walls (Fig. 13).

LA Local Deformation Analysis

The greatest principal strain λ1 in the most central LA images is shown in Fig. 15 (septal and lateral walls only). Differences between septal and lateral walls were not significant because of the high variation between hearts; however, λ1 increased toward the base (P<.01 base versus apex). The greatest shortening (λ2) showed small variations between septal and lateral walls (Fig. 16). Shortening in the septum increased toward the apex (P<.05), whereas the lateral wall showed no significant longitudinal variation. At the base, the lateral wall contracted more than the septum (P<.05). The direction of maximal lengthening was again oriented approximately in the radial (transmural) direction (Fig. 17). These directions are again relative to the ray joining the triangle centroid to the centroid of the LV contour. This direction deviates from perpendicular to the walls at the base and apex, so there is a positive/negative variation between septum and lateral walls in Fig. 17. The change...
in triangle area also showed a net loss between ED and ES in the LA images ranging from approximately 14% at the apex to 4% at the base.

**Discussion**

The SPAMM technique provides a large number of material landmarks from which a number of kinematic parameters can be calculated. Alternative tagging procedures use selective radio frequency pulse sequences, but these do not readily generate many tag planes.

Without two-dimensional grid tagging, trackable landmarks have been primarily limited to the intersections of the epicardial and endocardial contours with the tag plane. In the SPAMM technique, the trackable points are the intersections of the tag planes and have two major advantages: (1) more points can be tracked across the wall, and (2) the imaged motion is a projection of the true three-dimensional motion onto the image plane. The latter is due to the fact that the tag planes are generated orthogonal to the image plane. This is not true for the intersection with the epicardial and endocardial surfaces because the curvature of these surfaces combines with through-plane motion to produce apparent motion in the image plane.

**Displacement**

Little emphasis has been placed on the measurement of the displacement field of the LV because it is made up of a number of complicated effects such as circumferential and longitudinal shortening, wall thickening, and LV torsion. However, it is an important indicator of the contractile state and loading conditions experienced by the myocardium. The reduced displacement in the septum in both SA and LA images may be due to the influence of the right ventricle on LV motion. Subjects whose right ventricle did not extend as far toward the LV apex as others appeared to have greater motion in the apical septal region, although these effects await further study.

Rogers et al. used selective tags and radial LA image orientations to measure longitudinal displacements. An average longitudinal translation of 12.8±3.8 mm at the base and 1.6±2.2 mm at the apex was found, with a similar circumferential variation around the base as seen in Fig 10 (15.2±3.5 mm at the posterior base versus 12.8±4.5 mm at the anterior base). This variation in displacement, combined with the posterior motion of the basal SA centroid and the anterior motion of the apical SA centroid, implies a rocking motion of the LV about a septolateral axis. The large motion of the valve plane toward the apex (Fig 10) suggests that the expansion of the atria and their loading of the LV base should be considered in mathematical models of LV mechanics.

These results are difficult to compare with echocardiography because the latter technique does not track material points. Ingels et al. have studied the motions of radiopaque tantalum screws implanted in transplanted hearts. Viewed from an external frame of reference, the transplanted heart behaved as a "double bellows," with a relatively immobile septum toward which the free walls of both ventricles contract. However, their results indicate abnormal motions, eg, 3- to 4-mm displacement of septal markers directed away from the LV interior, with characteristics similar to right ventricular pressure overload. On the basis of echocardiographic data, they estimated a normal systolic excursion of the septum midwall of 3.4 mm compared with 12 mm in the posterolateral free wall. These estimates are qualitatively similar to the circumferential variation shown in Fig 5, top.

**Rotation and Torsion**

The torsion of the LV during systole has been well documented in studies involving bead implantation and MR tagging and has been proposed as a mechanism for equalizing the transmural gradient of fiber
shortening. Buchalter et al.10 used four radially oriented tagging planes placed parallel to the central axis of the LV and tracked the intersections of the tag stripes with the epicardial and endocardial contours in SA images to obtain rotation relative to the base. Torsion in posteroseptal regions was found to be less than in anterolateral regions; however, no significant variation was found in the present study. This difference might be the result of the method of centroid calculation. Buchalter et al.10 used separate centroids for the inner and outer marker sets to calculate endocardial and epicardial torsions. Because of the increased error in locating the endocardial boundary on the images, we used the same centroid for all markers. The definition of the centroid is important because its motion served to reduce the circumferential variation in torsion. The transmural gradient (8° between epicardial and endocardial contours10) was greater than that observed in the present study (4° to 6°). This difference is probably the result of the nature of the tags tracked: Buchalter et al. used the intersections of the tags with the epicardial and endocardial boundaries, whereas we used internal tags to avoid errors resulting from the location of the endocardial boundary and effects due to wall curvature and through-plane motion on calculated displacements. The transmural gradient of rotation thus appears to be similar in both studies because our endocardial and epicardial measurements were averaged over the inner and outer thirds of the wall, respectively.

Hansen et al.17 measured relative rotation using midwall radiopaque markers in human cardiac allograft recipients and found greater torsion in the lateral (23.4±10.7°) and posterior (18.8±6.3°) walls than in the anterior wall (13.3±6.0°) at the apex. Some of the difference between the bead implantation and the MR results might be because of the calculation of the centroid about which the points rotate. Implanted bead studies17-19 often use apical and basal beads to define a moving LA, whereas MRI studies must use the LV boundaries. The relative rotations of the more apical markers were calculated relative to a minor axis defined by posterior and anterior basal beads17,18 as opposed to the MR studies, in which apex and base rotations were compared on a regional basis (septum with septum, etc.). In the MR studies, endocardial contours are often very difficult to determine, especially at the apex. Our results indicate that the concept of torsion, although a useful first approximation to the true motion, is an oversimplification. The true motion cannot be described by a rotation of all points about a unique axis of symmetry (Fig 4). If a unique axis of rotation exists, it is likely to be closer to the interventricular septum.20

Strain
In the SA images, the direction of greatest stretch (λ1) is approximately radial in all regions. The maximum stretch is thus associated with wall thickening. Circumferential shortening is associated with the maximal contraction λ2. The variance in the maximum principal strain λ1 is larger than that of the minimum principal strain λ2 in both SA and LA images. This is also true of Waldman et al.'s three-dimensional bead study of canine hearts,5 in which the SD of the maximum principal strain was 0.13 compared with 0.05 in the minimum principal strain. One source of error is the assumption of homogeneous strain within marker triangles. Transmural strains are the least well estimated because the strain is varying most in this direction.21 Also, simulations of the effect of marker noise on the error in λ1 and λ2 show that the error in the former is always higher than the latter.7

There was a consistent variation in principal angle β from negative values at the apex to positive at the base. This angle between the radial direction and the direction of the largest principal strain is related to the radial-circumferential transverse shear Eθr by

\[
\sin 2\beta = \frac{2E_{\theta r}}{\lambda_1 - \lambda_2}
\]

β has been shown to change (become more circumferential) with ischemia.22 The longitudinal variation seen in the present study implies that in the normal heart, the Eθr shear changes sign from apex to base.

In the images studied, λ1 appeared to increase in magnitude toward the apex, but λ2 showed the opposite trend. This led to an increase in the amount of area lost by the triangles during systole in both SA and LA views. It is possible that endocardial trabeculation is insufficiently resolved and shortening at the endocardium is measured across a combination of muscle and cavity in some areas, producing a greater apparent reduction in area. It is also possible that because of the increased curvature of the wall through the slice near the apex, the three-dimensional maximum principal strain becomes more out of plane, leading to an apparent decrease in λ1. These effects require further study using high-resolution images.

Circumferential shortening was measured by Clark et al.23 in a one-dimensional analysis of SPAMM images. They measured the dimension changes parallel to the endocardial and epicardial surfaces between closely spaced (7 mm) radially oriented stripes. Basal slices showed less circumferential shortening between ED and ES than apical slices (28±9% versus 34±13% at the midwall, corresponding to strain values of −0.24 and −0.28, respectively). In the case of shortening strains, the strain λ is related to the percent segment length change (%S) by the following equation:

\[
%S = (1 - \sqrt{\lambda - 1}) \times 100\%
\]
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shortening
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shortenings
(13%,
sites
values
shortening
dimensional
shortening,
increasing
(LV)
in
approximately
in
longitudinal
during
longitudinal
shortening,
the
LV
LA
and
LV
endocardium.
Similar
trends
were
found
in
this
study
for
\( \lambda_1 \),
which
is
oriented
greatly
in
the
longitudinal
direction.
At
the
base,
longitudinal
shortening
was
greater
in
the
free
wall
than
in
the
septum:
25\pm5\%
varying
17\pm4\%,
or
strains
of
\(-0.22\)
versus
\(-0.16\),
comparable
to
\( \lambda_1 \) values
of
\(-0.19\pm0.03\)
versus
\(-0.14\pm0.03\)
in
the
present
study.
Also,
there
was
a
longitudinal
dradient
of
shortening,
increasing
from
apex
to
base
in
the
lateral
free
wall
but
decreasing
from
apex
to
base
in
the
septum,
as
in
Fig
16.

Ultrasonic
crystals
have
been
implanted
in
the
midwall
parallel
to
the
epicardium,
showing
a
direction
of
maximal
shortening
oriented
50°
below
longitudinal.7
These
planes
are
not
normally
imaged
with
MRI,
and
comparisons
await
three-dimensional
reconstructions
from
the
SA
and
LA
images.27,28

Ingels
et
al18
measured
circumferential
and
longitudinal
shortenings
in
human
heart
transplant
recipients
by
using
implanted
midwall
radiopaque
markers.
Circumferential
shortening
at
the
midventricular
level
was
the
least
in
the
septal
and
posterior
walls
(11\pm4\% and
12\pm5\%)
and
greatest
in
the
anterior
wall
(16\pm5\%).
Fig
12
shows
the
same
trends,
although
the
shortening
values
are
higher
(\(-0.2\)
strain
corresponds
to
22\%
shortening).
This
may
be
attributed
to
a
reduced
contraction
in
the
transplanted
hearts19
and
to
the
fact
that
\( \lambda_1 \)
was
not
aligned
exactly
in
the
circumferential
direction.
In
the
longitudinal
direction,
shortening
was
similar
for
all
circumferential
sites
(13%,
or
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strain),18
again
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value
of
\(-0.17\)
found
in
this
study
(Fig
16).
Similar
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LeWinter,26
circumferential
shortening
was
found
to
be
greater
than
longitudinal
shortening
in
the
anterior
wall
but
not
in
the
posterior
wall.18

Comparison
With
Disease

To
illustrate
the
usefulness
of
a
database
of
normal
LV
motion,
we
applied
the
same
techniques
to
a
patient
with
a
basal
anteroseptal
infarct.
Displacement
of
points
in
a
basal
SA
slice
are
shown
in
Fig
18.
Deformation
parameters
are
summarized
in
Table
3
for
the
basal
regions
of
the
LV.
It
can
be
seen
that
displacement
is
reduced
from
normal
values
in
the
septal
and
anterior
walls
(cf,
Fig
5,
top),
rotation
in
the
lateral
and
anterolateral
regions
is
increased
in
magnitude
(cf,
Fig
6,
top),
and
principal
strains
are
decreased
in
magnitude
in
the
anterior
wall
(cf,
Figs
11
and
12).
The
principal
direction
in
the
lateral
and
anterolateral
walls
has
shifted
away
from
the
radial
direction
(cf,
Fig
13),
and
the
triangle
areas
remain
essentially
unchanged
in
the
septal
and
anterior
walls
(cf,
4%
decrease
in
the
normal
hearts).

Study
Limitations

The
two-dimensional
nature
of
the
image
data
places
a
number
of
limitations
on
the
analysis.
First,
there
is
a
potential
error
in
the
location
of
points
within
the
heart
because
initial
locations
of
the
material
points
that
are
imaged
in
subsequent
frames
are
not
known.
For
example,
because
of
motion
through
the
SA
planes,
the
locations
of
later
eytes
SA
points
are
more
basal
than
those
calculated
on
the
basis
of
the
location
of
the
image
plane.
This
error
was
smaller
than
the
size
of
the
regions
in
which
data
were
pooled;
the
basal
level
was
therefore
chosen
to
be
relatively
large
(\( z>0.55 \))
because
of
the
high
degree
of
through-plane
motion.
Second,
three-dimensional
dero-

![Fig 18. Plot of loci of tag intersections in a basal short-axis location for a patient with an anteroseptal infarct.](http://circ.ahajournals.org/)

<table>
<thead>
<tr>
<th>TABLE 3. Short-Axis Basal Deformation Parameters in a Patient With a Basal Anteroseptal Infarct</th>
</tr>
</thead>
<tbody>
<tr>
<td>Parameter</td>
</tr>
<tr>
<td>-----------</td>
</tr>
<tr>
<td>D, mm</td>
</tr>
<tr>
<td>( \delta_0 )</td>
</tr>
<tr>
<td>( \lambda_1 )</td>
</tr>
<tr>
<td>( \lambda_2 )</td>
</tr>
<tr>
<td>( \alpha_1 )</td>
</tr>
<tr>
<td>( \beta_1 )</td>
</tr>
<tr>
<td>( \Delta A )</td>
</tr>
</tbody>
</table>
tions were not calculated. Thus, the principal shortening strains imaged are less than the maximum shortening experienced by the tissue because the three-dimensional principal shortening direction is approximately parallel to the obliquely oriented subepicardial muscle fibers. Despite this, the analysis is a true reflection of the two-dimensional motions undergone because the tagging lines are all initially orthogonal to the imaging planes. A full three-dimensional treatment requires a greater complexity in the data analysis. It is therefore likely that two-dimensional parameters may prove to be more useful in the clinical setting than their three-dimensional counterparts, at least in the short term.

The measurement of torsion depends on an accurate estimate of the moving centroid of each contour. At the apex, the inner contour was often difficult to determine because of confusion with the tags, trabeculation, and blurring resulting from respiration, as well as partial volume effects resulting from the oblique intersection of the wall with the imaging plane. Better contours are obtainable on untagged bright-blood spin-echo images or rapid gradient-echo breathhold images. Therefore, the geometry of the heart could be better determined by an initial untagged image series, and the motion by a tagged sequence. A measure of torsion that does not rely on the location of the contour is the circumferential-longitudinal shear strain. However, this requires a full three-dimensional analysis. A measure of rotation that does not rely on the location of the centroid is \( \alpha \), the local rotation of the triangle.

The main limitation of the current implementation is the limited transmural resolution of the tags. With a 7-mm stripe spacing, only one or two triangles can be formed across the wall, making transmural variations difficult to resolve. It is also possible that endocardial trabeculation and interstices are insufficiently resolved and that shortening at the endocardium is measured across a combination of muscle and cavity in some areas. Higher resolution is obtainable with breathhold images, in which the blurring of the tags is reduced, in conjunction with cardiac coils. Faster imaging techniques would also allow the acquisition of more locations within a given heart, with radially oriented LA image planes. It is also possible to extend the analyses to higher-order finite elements, thereby reconstructing the continuous strain field and reducing the statistical scatter present in homogeneous strain analyses. These higher-order elements cover several tags and allow information about the deformation of the stripes connecting the intersections to be included in the deformation analysis. This would render the technique more accurate as more data are used from each image.

In conclusion, we have attempted to characterize normal heart wall motion as seen in two-dimensional MR images to build a database with which to compare hearts with disease or impaired function. Displacement, rotation, and circumferential and longitudinal strains are estimated to sufficient accuracy to resolve consistent regional variation. Transmural gradients of deformation currently are not well defined and require higher resolution of tag and endocardial contour definition.

**References**

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A A Young, H Imai, C N Chang and L Axel

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