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Precision of Myocardial Contour Estimation from Tagged MR Images with a “Black-Blood” Technique¹

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Abstract

Rationale and Objectives—The authors determined whether blood presaturation of tagged magnetic resonance (MR) images affects identification of left ventricular endocardial borders.

Materials and Methods—Three healthy volunteers underwent MR imaging performed with a breath-hold segmented spoiled gradient-recalled-echo sequence with tissue tagging. Two saturation pulses (in the basal and apical regions) were used to generate black-blood images. Manual segmentation of endocardial contours on black-blood and white-blood images was performed independently by five observers.

Results—Endocardial borders were better identified on black-blood images compared with white-blood images, especially in the early systolic phases. Interobserver variability in contour estimation was significantly higher for white-blood images ($P < .001$) and was twice that for corresponding black-blood images during early systole. Contour variability appeared to be affected mainly by tag-to-myocardium contrast ($P = .009$) and myocardium-to-chamber contrast ($P = .05$).

Conclusion—Blood presaturation of tagged MR images improves reliability of contour segmentation.

Keywords

Heart, function; heart, MR; magnetic resonance (MR), comparative study; magnetic resonance (MR), pulse sequences

Magnetic resonance (MR) imaging of the heart is a noninvasive method of quantitatively assessing cardiac function by obtaining spatially registered images of the entire heart throughout its contractile cycle. In the past, estimates of local cardiac function have been made based on changes in the shape of the heart that occur throughout the contractile cycle. This approach, as well as clinical ultrasonographic imaging methods (1–5), relies on myocardial border identification (ie, endocardium and epicardium) for the calculation of gross myocardial wall thickening. More recently, myocardial tagging by means of MR myocardial tagging has been implemented to record unambiguously the intrinsic motion within the myocardial wall. MR tagging produces localized perturbations of magnetization across the image (the tags). The deformation of the tags can then be tracked to assess the motion of myocardial tissue itself (6,7). MR tags can be localized more accurately than contours (8), and the tags also provide transmural information.

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Contour segmentation is still necessary, however, to define a volume of interest in which to calculate strain from the MR tags (9). This volume of interest is also used in three-dimensional display of the reconstructed heart model. Therefore, it is important that accurate contour segmentation be performed on the tagged images. It is possible to acquire a separate set of images without tags specifically for contour detection, but obtaining this separate set of images would lengthen the examination time and would provide an opportunity for misregistration to occur between image sets.

Several techniques have been proposed for improving image acquisition and segmenting the myocardial contours. A high-speed partial k-space gradient-recalled acquisition has been implemented to reduce motion artifacts while a cine loop of a section is obtained during a breath hold (10). Many researchers have developed methods for segmenting the left ventricular contours with varying degrees of automation (11–23). These automated methods reduce analysis time as well as human bias in contour estimates.

Although use of manual editing techniques has been minimized by such refinements, manual editing is still necessary for the optimization and correction of automatically detected contours. Gradient-recalled-echo pulse sequences used to obtain a cardiac cine loop produce a high signal intensity for moving spins and are referred to as “white-blood” techniques. Applied in conjunction with pulsed magnetic field gradients, the tagging pulses produce saturated tags that appear on the image as a dark pattern that reduces the overall signal intensity of ventricular blood after the tags are “mixed in.” This decreased signal intensity leads to greater difficulties in identifying the boundary between the endocardium and the ventricular cavity.

“Black-blood” imaging has been previously described as a technique used to presaturate inflowing spins and thereby reduce the image brightness of blood in the left ventricular cavity (24–27). Black-blood imaging methods are not adapted, however, for the breath-hold cine acquisitions that are necessary to measure regional cardiac function accurately. To solve this problem, we designed a presaturation pulse that is used in conjunction with tagging pulses to produce black-blood images. An inversion pulse is applied at end systole in the atria in combination with an apical saturation pulse to saturate blood before image acquisition.

The purpose of this study was to investigate whether the saturation used to create black-blood tagged images substantially affects the identification of endocardial borders and the variability in manual contour editing compared with white-blood tagged images.

MATERIALS AND METHODS

Imaging Protocol

Three healthy volunteers (two men, one woman; age range, 29–35 years) were examined with a 1.5-T MR imager (Signa; GE Medical Systems, Milwaukee, Wis). The breath-hold cine MR imaging protocol consisted of an electrocardiogram-triggered segmented k-space spoiled gradient-recalled-echo pulse sequence (10) with a surface radio-frequency flex coil used as a receiver. Sequential and contiguous stacks of short-axis images in double obliquity were prescribed to image the entire heart from base to apex. Six breath holds were necessary to acquire a complete set of images. The maximum number of cardiac phases was determined from the heart rate. Twelve phases were necessary to image the systole period and were acquired in 23 heartbeats. The following imaging parameters were used: echo time of 2.3 msec, repetition time of 6.5 msec, α (flip angle) = 15°, one signal acquired, field of view of 36 cm, matrix size of 256 × 110, and section thickness of 10 mm. Parallel-line tissue tagging was triggered by the up-slope of the QRS complex of the electrocardiogram, immediately before the imaging pulse.

Two selective inversion pulses were used to generate the black-blood images (Fig 1). The first pulse was applied at end systole after the end of imaging, and it inverted the magnetization in a 10-cm slab near the base of the imaging section. This pulse saturated the blood in the atria and pulmonary veins just before diastolic filling. The second pulse was applied to a 10-cm slab at the apex of the imaging plane immediately after the QRS complex. This pulse further saturated blood in the ventricle before imaging. For the white-blood images, the amplitude of the saturation pulses was set to zero. All other parameters were identical for black-blood and white-blood imaging and were optimized for maximum tag contrast and image-acquisition speed.

Each volunteer was successively examined, first with the white-blood sequence and then with the black-blood sequence. The examination lasted approximately 30 minutes. The protocol, in agreement with National Institutes of Health guidelines, was approved by our institutional committee on human research.

Image Sets

Sections at three different levels—basal, midventricular, and apical—were selected to constitute the image reading sets used for analysis. Those sections that were immediately contiguous with the most basal and apical sections were deemed basal- and apical-level sections to avoid partial-volume effects with surrounding regions. Midventricular sections were selected to include both superior and inferior papillary muscles. To observe temporal changes in image contrast, we selected eight time frames for each section level from the total of 12 frames; the time between frames was 32 msec. All contiguous time frames (images 1 to 5) were selected in the first half of systole, but only images 7, 9, and 11 were retained during the second half. The beginning of the cardiac cycle was more closely sampled to take into account the rapid changes in contrast that occur at this time. The same locations and time frames were used for white-blood and black-blood image stacks in each of the three volunteers.

Each reading set of black-blood and white-blood images was divided into three subsets according to anatomic level (basal, midventricular, apical). Within each subset, the sequence of images was randomly assigned for analysis. Five trained observers were asked to use manual editing techniques to perform endocardial-border segmentation independently with a customized contouring software package developed on Silicon Graphics workstations (Mountain View, Calif). To display corresponding black-blood and white-blood images, we used the same standardized region of interest, image center, and window and level settings on each section selected for all five observers. Cinematic display was available to observers to facilitate the identification of endocardial borders. Observers were not permitted to use time or space interpolation to improve their contour estimates. The amount of time allowed for segmentation of each case was not restricted. Only one subset of images was segmented at a time to prevent fatigue. A minimum delay of 24 hours was required between each subset. To avoid recognition of corresponding images, a time interval of at least 2 weeks was required between readings of black-blood and white-blood image sets.

Analysis Protocol

For analysis of contour variability, we analyzed the contour positions given by the five observers by measuring the distances from the center of the left ventricular cavity to 16 equiangular points on the endocardial boundary of each image. We calculated the average position at the 16 equiangular points to determine the average contour.

We assessed interobserver contour variability by means of statistical comparison with the average contour. Results were categorized by anatomic level, as well as whether the images originated from black-blood or white-blood data sets. We tested for equality of variance

between black-blood and white-blood groups and between anatomic levels by using two-tailed *F* tests ($\alpha = 0.01$) (28). For each group, we analyzed changes in variability over time by means of regression analysis.

We examined the respective locations of average contours on corresponding black-blood and white-blood images and used paired-sample *t* tests to determine whether average positions were comparable. We further assessed the statistical significance of mean differences over time and between anatomic levels by using analysis of variance.

To understand better the factors that affect variability of contour editing, we used a multiple-regression model to explore the relationship between contour variability and covariates that may affect border conspicuity. For this analysis, measurements were collected from two locations at the basal level to limit partial-volume effects with endocardial trabeculations or papillary muscles. The first location was on the posterior wall where the tag lines were perpendicular to the endocardial border. The second location was on the septal wall where the tag lines were parallel to the endocardial border. At the two selected locations and within a 6-mm window centered on the mean contour position, we measured contour variability, maximum image signal intensity gradient, myocardium-to-chamber contrast, and tag-to-myocardium contrast. The dependent variable chosen in our model was the standard deviation of contour position among the five observers. We assessed the effects of the continuous independent variables (maximum gradient, myocardium-to-chamber contrast, tag-to-myocardium contrast) and coded two additional variables as dummy variables: nature of blood signal intensity (1 = white blood, 0 = black blood) and tag orientation relative to endocardium (1 = perpendicular, 0 = parallel). We further assessed second-order interaction terms between the independent variables.

Statistical significance was inferred when *P* was less than or equal to .05, and all reported *P* values were two tailed. Statistical analysis was carried out with commercially available software (Stata 4.0; Stata, College Station, Tex).

RESULTS

Figure 2a shows black-blood and corresponding white-blood tagged images of the same volunteer; short-axis views at the basal level at three different phases during systole are shown. Intraventricular structures (superior borders of papillary muscles) and myocardial boundaries are clearly depicted on all of the black-blood images. By comparison, visualization of intracavitary structures and endocardial borders, especially early in systole, is more difficult on the corresponding white-blood images. The effect of presaturation of flowing blood on black-blood imaging compared with white-blood imaging in terms of signal intensity is shown in Figure 2b. A total of 72 white-blood and 72 black-blood short-axis images were analyzed by each of the five observers. Even though no time restriction was applied during contour segmentation, observers consistently reported greater difficulties with white-blood images than with black-blood images.

Interobserver variability in contour estimation was always significantly lower with black-blood images than with white-blood images ($P < .001$) (Fig 3). Although the distribution of deviations from the estimate of mean endocardial position was not normal, it was nearly symmetric with zero mean. The variability for black-blood images was about half that for white-blood images at basal and midventricular levels. Variability significantly increased at the apical level for black-blood images ($P = .007$). No changes in variability were reported for white-blood images as a function of anatomic level.

Temporal changes in variability were markedly different for black-blood and white-blood images (Fig 4). For black-blood images, variability remained unchanged during most of the

systolic portion of the cycle (from 0.94 mm at t_0 to 0.98 mm at t_6) and increased after end systole. Variability for white-blood images was more than twice that for the corresponding black-blood images during early systole, and after a significant decrease during midsystole ($P = .01$) it consistently remained at a level that was at least 50% that for black-blood images.

Overall, average contours on black-blood images appeared significantly larger than those on white-blood images for all anatomic levels ($P < .001$) and all time frames ($P < .001$) (Fig 5). The difference between black-blood- and white blood-derived contours was greatest at the apical level (mean difference, 2.08 mm; standard error, 0.12; $P < .001$). This finding remained the same over time ($P = .82$). The amplitude of the difference decreased for the basal level (mean difference, 1.6 mm; standard error, 0.10; $P < .001$) and for the midventricular level (mean difference, 0.93 mm; standard error, 0.14; $P < .001$). For basal and midventricular levels, the difference was the highest at end diastole ($P < .001$), and it became nonsignificant at end systole ($P = .11$ and $P = .82$, respectively). The extent of these differences was the same for all three volunteers.

Regression analysis showed that contour variability was significantly affected by tag-to-myocardium contrast ($P = .009$) but not by the value of the maximum gradient ($P = .26$). The average effect of myocardium-to-chamber contrast was significant when controlling for blood signal intensity ($P = .05$). But the fact that there was an “inter-action,” or “interaction effect” to be more accurate statistically speaking, between myocardium-to-chamber contrast and blood signal intensity indicated that the effect of myocardium-to-chamber contrast was strongly dependent on the nature of the blood signal intensity ($P = .001$). The orientation of the tagging pattern could not help explain changes in contour variability ($P = .41$).

DISCUSSION

The results of this study suggest that there is significantly less interobserver contour variability for black-blood tagged images compared with white-blood tagged images ($P < .001$), regardless of anatomic location. Variability in contour editing was reduced when black-blood images were used rather than white-blood images. We assumed that the quality of a contour estimate was inversely proportional to the variance of the estimate. Contour estimates must be not only accurate but also reproducible to provide clinically useful results. Knowledge of interobserver variability also provided an index of reliability of the measurements obtained with each technique.

Difficulties with edge detection on images obtained with breath-hold gradient-recalled-echo sequences relate largely to signal intensity of blood flow. Fast-flowing blood typically appears bright, whereas slow-flowing regions, such as along the myocardial wall, usually have intermediate signal intensity because they are partially saturated by multiple section-selective excitation pulses. Differentiation between slow-flowing regions and stationary structures becomes difficult and is subject to a large degree of subjectivity, especially when regions of heavier trabeculation are encountered, such as in the apical region. Presaturation of flowing blood, which was used in the black-blood technique, improved contrast between flowing blood and stationary tissues, especially in slow-flowing regions. Slow-flowing blood, which tends to cause the greatest amount of flow-related signal intensity, will be subject to the greatest amount of presaturation. The overall efficiency of presaturation also depends on the amount of blood being saturated. This is evident in the apical regions where less blood passes through the presaturation region. This finding, along with partial-volume effects, explains the greater variability of apical black-blood contours compared with those at other levels.

Contour variability also depends on the cardiac phase during systole. In our study, the greatest improvement in black-blood versus white-blood images was observed during early systole. At

this time, detection of wall contour is dependent on interruption of the tag lines at the wall boundaries and on the overall contrast between wall and blood signal intensity. In early systole, saturated tag lines remain visible across all regions of the images and in the cavity in particular. Locally saturated blood regions are indeed not yet entirely mixed with the nonsaturated regions. Depending on the nature of the signal intensity of the inflowing blood determined by the presaturation pulse, the dark tags have to mix with either black blood or white blood. Mixing locally saturated dark regions (eg, tagged blood) with white blood resulted in a grayish decreased blood signal intensity that decreased the contrast between myocardium and cavity and contributed to the difficulties encountered in recognizing endocardial boundaries on white-blood images.

Variability increased on black-blood images during the late systolic phases to a level similar to that of white-blood images. This finding was related to the increase in ventricular blood signal intensity observed on late systolic black-blood images, as well as the decrease in tag contrast due to the longitudinal relaxation of magnetization.

The size of the left ventricular cavity was underestimated and wall thickness was overestimated when endocardial contours were analyzed at end diastole on white-blood images. The magnitude of the difference was dependent on the cardiac phase and was statistically significant only during the first part of the cardiac cycle, except at the apical level ($P < .01$). Because functional indexes are commonly normalized by using end-diastolic measurements, wall thickening tends to be underestimated on white-blood tagged images compared with black-blood images. In our experience, a difference of 2 mm between black-blood and white-blood contours at end diastole led to a 25% underestimation of systolic wall thickening (black-blood contours were 10 mm at end diastole and 15 mm at end systole; white-blood contours were 12 mm at end diastole and 15 mm at end systole).

To quantify more objectively the observational errors and to determine the major factors that influence detectability of myocardial borders, we investigated the factors that most directly influence this method of detection: contrast between the contiguous structures, myocardial edge sharpness, and contrast and direction of tissue tagging. As expected, contour variability was closely related to myocardium-to-chamber contrast, and higher contrast values were associated with black-blood images. This finding is illustrated by the strong interaction found between those two factors in the multiple-regression model. The maximum gradient at the edge is an indicator of edge sharpness but did not appear to influence border conspicuity or contour reliability. Tissue tagging significantly influenced border conspicuity because tag-to-myocardium contrast was strongly associated with better reproducibility of contours ($P = .009$). The direction of the tagging pattern, however, did not statistically significantly affect contour reliability.

The two saturation pulses are very short and are simple to prescribe because they have the same orientation as the short-axis imaging sections. One drawback of this method is that the basal saturation pulse must be used at end systole to invert the blood before it fills the ventricles. This method does not present a problem for the spoiled gradient-recalled-echo segmented k-space sequence, but this pulse may interrupt the steady-state condition of magnetization during cine phase-contrast sequences.

In this study, we evaluated only manual contour editing and focused on observer variability. Manual detection methods have largely been replaced, however, by semiautomated contour-detection algorithms, and user intervention is now limited to correcting errors in the contours produced. Therefore, our results probably amplify the effect of observer subjectivity and cannot be directly extrapolated to the clinical arena. Because an improvement in the identification of the endocardial border on black-blood images compared with white-blood images was found

with manual editing, we would also expect an improvement in automated segmentation results with this method. Such an improvement would result in less time being spent manually editing contours and less errors occurring due to human bias in regions that are edited.

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Figure 1. Illustration of presaturation scheme for black-blood imaging. *SPGR* = spoiled gradient recalled echo. (Reprinted, with permission, from reference 28.)

a.

b.

Figure 2.

(a) White-blood (upper) and black-blood (lower) midpapillary-level short-axis images of a 30-year-old volunteer (electrocardiogram trigger delay was 46, 176, and 311 msec). Endocardial borders are well delineated by the saturated blood from the first time frame on black-blood images and throughout systole. These borders are difficult to detect on first-time-frame white-blood images because the inflowing spins have not yet mixed with the saturated tags. (b) Graph illustrates time course of absolute signal intensity of black-blood (*BB*) and white-blood (*WB*) images (ventricular blood, tag lines, and myocardium signal intensity) at the basal level in same subject as in a. Note that the contrast between myocardium and ventricular blood is always greater over time on black-blood images than on white-blood images.

Figure 3.

Graphic illustration of interobserver variability of contour estimation. Distribution of deviation from the mean estimate is shown for black-blood (*BB*) and white-blood (*WB*) images at (top) basal, (middle) midventricular, and (bottom) apical levels (all time frames included). *SD* = standard deviation.

Figure 4. Graphic illustration of temporal pattern of segmentation variability. Variability for black-blood (*BB*) images is compared with that for white-blood (*WB*) images at all anatomic levels.

Figure 5. Temporal pattern shows the average of the magnitude of the difference between black-blood (*BB*) and white-blood (*WB*) contours.