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Multi-Shot EPI for Improvement of Myocardial Tag Contrast: Comparison with Segmented SPGR

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Abstract

To assess the potential value of multi-shot EPI relative to segmented k -space SPGR for myocardial tagging, we measured tag contrast for both sequences in a phantom and human study and compared it with theoretical predictions. In the human heart, EPI tag contrast was three times that of SPGR at the end of systole. Tag duration was lengthened with EPI to at least 600 ms. In addition, the entire heart was examined in a total of 32 heartbeats with EPI versus 152 heartbeats with SPGR.

Keywords

magnetic resonance imaging; EPI; SPGR; myocardial tagging

INTRODUCTION

Myocardial tagging techniques can provide unique measurements of myocardial deformation (1-4). Because of misregistration caused by breathing, emphasis has recently been placed on the development of single breath-hold pulse sequences capable of acquiring multiple phases of the cardiac cycle in a reduced number of heartbeats. Segmented k -space tagged SPGR sequences have gained acceptance and can be implemented on standard hardware (5,6). These sequences, however, suffer from a poor signal-to-noise ratio and significant tag fading by end-systole. Furthermore, only one slice per breath-hold can be acquired with tagged SPGR. Echo-planar imaging (EPI) (7-9) is a method that provides not only the potential for imaging the entire heart in a single breath-hold, but also higher signal-to-noise ratio and taghissue contrast. A recently installed EPI system enabled us to explore EPI myocardial tagging. In this study, we compared the tag contrast in SPGR and EPI images.

METHODS AND THEORY

Pulse Sequence

Figures 1a and 1b show the SPGR and EPI tagging pulse sequences, respectively. Parallel tags were used in a direction perpendicular to the readout direction, which had higher spatial resolution than the phase encoding direction. Magnetization in the tagged regions was inverted to maximize the persistence of tag lines. The tagging pulse was ECG-triggered and followed by multiple RF excitations. For any one slice, multi-phase images were captured in a single breath-hold. For any one phase, multiple phase encoding views were acquired in the same R-R interval. The views were interleaved to reduce ghost artifacts due to signal modulation in

the phase encoding direction (5, 6, 10). In SPGR, a truncated RF excitation pulse and partial readout were used to reduce the echo time (TE) and the repetition time (TR), so that motion artifacts were minimized and high temporal resolution images were obtained. The phase cycling method (11) was used to spoil the nonprimary echoes. In EPI, the slices were interleaved such that each slice was excited only once per cardiac cycle. The effective repetition time was the length of the cardiac cycle. The effective repetition time was the length of the cardiac cycle. A spectral-spatial selective pulse (12) that excited water only was used to reduce chemical shift artifact.

Theory

In SPGR, a slice is excited repeatedly. The longitudinal magnetization just before the n th FW pulse can be expressed as

$$M_z(n) = 1 + [M_z(n-1) \cos \alpha - 1] e^{-TR/T_1} \quad [1]$$

where T_1 is the spin-lattice relaxation time, TR is the repetition time, α is the flip angle, and M_0 is defined as 1. The difference in longitudinal magnetization of spins in nontagged and tagged regions is

$$\Delta M_z(n) = \Delta M_z(n-1) \cos \alpha e^{-TR/T_1} \quad [2]$$

Using this relation recursively, the tag contrast at the n th RF pulse can be expressed as a function of the tag contrast at the first time frame, $\Delta M_z(1)$, which is

$$\Delta M_z(n) = \Delta M_z(1) \cos^{n-1} \alpha e^{-(n-1)TR/T_1} \quad [3]$$

Because magnetization in the tagged regions is inverted after each ECG trigger, magnetization in the tagged regions takes longer to reach a steady state than in nontagged regions. The approach to steady state is very dependent on the imaging flip angle used. As the spins in both regions are driven to equilibrium in an SPGR sequence, the available M_z prior to the next tagging pulse is much smaller. Subsequently, the zero crossing of the inverted magnetization is advanced, leading to less persistence of the tags. In our experimental conditions, the difference of magnetization in both regions in the beginning of R wave is very small because of the multiple excitation pulses in the same physical location and the relatively long recovery time after the last RF pulse. This is demonstrated by the low tag contrast in the end systolic images. If we can assume that magnetization in both tagged and nontagged regions has reached a steady state at the end of data acquisition in each cardiac cycle, the tag contrast at the first time frame is

$$\Delta M_z(1) = 2 \left(1 - \frac{1 - \cos \alpha}{1 - \cos \alpha e^{-TR/T_1}} e^{-(T-(N-2)TR)/T_1} \right) \quad [4]$$

where N is the total number of RF pulses applied in each cardiac cycle and T is the length of a cardiac cycle. Therefore, the tagging contrast of SPGR images after the n th RF pulse is

$$C_{\text{spgr}}(n) = 2 \left(1 - \frac{1 - \cos \alpha}{1 - \cos \alpha e^{-TR/T_1}} e^{-(T-(N-2)TR)/T_1} \right) \cdot \cos^{n-1} \alpha e^{-(n-1)TR/T_1} \sin \alpha e^{-TE/T_2^*} \quad [5]$$

where TE is the echo time, and T_2^* is the effective spin-spin relaxation time. The error in estimating the initial tag contrast will only shift the tag contrast curve up and down, but will not change the shape of the curve. In the results shown later, the theoretical curve was normalized to the first experimental data point.

For multi-slice and multi-phase EPI cardiac imaging, magnetization is not affected by the tagging pulse in the nontagged regions. The longitudinal magnetization in the nontagged regions of myocardium is independent of time delay after the ECG trigger; it is only dependent

on flip angle and T , which is equal to the length of a cardiac cycle. If a 90° pulse is used for imaging, the longitudinal magnetization of nontagged regions at any time, t , after the tagging pulse and just before the imaging RF pulse is

$$M_z(t) = 1 - e^{-t/T_1} \quad [6]$$

In the tagged regions, the spin magnetization is inverted by the tagging pulse at each QRS. Because the time from the tagging pulse to the imaging pulse is different for each phase, the signal intensity in the tagged regions is different for each phase, and the tags fade with time. The longitudinal magnetization in tagged regions at time t after the tagging pulse is

$$M_z(t) = 1 - 2e^{-t/T_1} + e^{-t/T_1} \quad [7]$$

The tag contrast between the tagged and nontagged regions is

$$C_{\text{epi}}(t) = 2(e^{-t/T_1} - e^{-t/T_1})e^{-TE/T_2^*} \quad [8]$$

Experiments

All the data were acquired on a GE 1.5 Tesla Signa system (Milwaukee, WI) equipped with a prototype GE single axis SR-230 echo-planar system. This EPI system can switch its fastest gradient from 0 to 2.3 Gauss/cm in $100 \mu\text{s}$. Because only a single fast gradient was available, all EPI images were acquired in the axial plane. The SPGR images were also acquired in the axial plane to ensure an equal comparison.

A stationary phantom made of copper sulfate ($T_1 = 800 \text{ ms}$, $T_2 = 500 \text{ ms}$) was used to emulate the spin-lattice relaxation rate of the myocardium. This phantom study was designed to validate the tag contrast predicted by Eqs. [5] and [8]. The head coil was used for both EPI and SPGR imaging. The phase encoding steps were chosen so that the spatial resolution was the same in both SPGR and EPI images. The pulse was triggered at every 1000 ms. In SPGR, 60 phases, which were used to demonstrate the signal evolution of tagged and nontagged regions, and eight slices were acquired. The FOV was 30 cm, voxel size was $1.2 \times 3.3 \times 10 \text{ mm}$. The repetition time, TR , was 6.5 ms, the echo time, TE , was 2.3 ms, and one average was used. An flip angle of 11° was chosen to optimize the tag contrast (10). Fractional readout was used to shorten the echo time. In the readout direction, 160 points were acquired and half Fourier reconstruction was used to obtain 256 point image resolution. The readout bandwidth was 64 kHz. In the phase encoding direction, the number of phase encoding steps were 90 and zero padding was applied for the remaining 166 views. One view was acquired for each phase in each cardiac cycle, so that the temporal resolution was 6.5 ms. In EPI, eight slices and eight phases were acquired with a temporal resolution of 37 ms. Voxel size was $1.2 \times 3.3 \times 10 \text{ mm}$. The repetition time, TR , was 1000 ms, the echo time, TE , was 13 ms, and flip angle was 90° . Rectangular FOV was used. Ten views were acquired for each phase in each cardiac cycle. All the data for each phase were obtained in four cardiac cycles. The temporal resolution was 37 ms. In the readout direction, 256 points of data were acquired and FOV was 30 cm. The readout bandwidth was 250 kHz. In the phase encoding direction, FOV was 21 cm, 40 phase encoding steps were used, and half Fourier reconstruction was used to obtain 64 point image resolution. Echo shifting was used in EPI to eliminate artifacts along the phase encoding direction. In both EPI and SPGR, a 24-pixel tag width and an 84-pixel tag separation were used to sample tag contrast accurately. The flip angle of spins in tagged regions was 180° , the duration of the tagging pulses was 18 ms, and the tagging gradient area between two RF pulses was 24 Gauss $\mu\text{s/cm}$. The tag contrast was measured by subtracting the minimum from maximum magnitude value along a line perpendicular to the tag lines after correcting for the sign of the tag for those time frames that preceded the zero crossing of tagged spins.

In a human volunteer study, the parameters were the same as those used in the phantom study unless noted otherwise. The body coil was used in both EPI and SPGR imaging. The tag shape was narrowed to a two-pixel tag width and a seven-pixel tag separation. In SPGR, images of eight slices and twelve phases were acquired in eight breath-holds; each breath-hold lasted 19 heartbeats, where the first heartbeat was used to stabilize the magnetization and no data were acquired. Five views were acquired for each phase in each cardiac cycle, so that the temporal resolution was 32.5 ms. In EPI, images of 8 slices and 8 phases were acquired in a single breath-hold lasting 32 heartbeats. The temporal resolution was 37 ms.

RESULTS

The experimentally measured and theoretically predicted contrast between tagged and nontagged regions in the phantom is plotted in Fig. 2. There is good agreement between the theoretical and experimental data. Fig. 3 shows the theoretical prediction of tag contrast when T_2^* is 20 ms, which is close to the expected T_2^* of myocardium. In Fig. 3, the tag contrast in EPI is higher at all times than in SPGR. In particular, the contrast in EPI is 100% and 260% higher than in SPGR at 60 and 330 ms after the ECG trigger, respectively. The tag contrast-to-noise ratio of EPI images at 600 ms is equal to the contrast of SPGR images at 200 ms; the tag lines of routine SPGR images can be easily visualized at this delay time.

Figure 4 shows representative images acquired from a normal volunteer at 60 and 330 ms after the ECG trigger using SPGR and EPI techniques, respectively. The tag contrast-to-noise in the echo-planar images is clearly better than in SPGR and is plotted in Fig. 5. The tag contrast-to-noise in EPI is 70% and 200% higher than in SPGR at 60 and 330 ms after the ECG trigger, respectively. The results are in good agreement with that in Fig. 3. The difference in Figs. 3 and 5 may be caused by the narrow tag lines and motion in the human study.

DISCUSSION AND CONCLUSIONS

As demonstrated in Figs. 2 and 5, not only is the tag contrast-to-noise ratio in EPI images much higher than in SPGR images but tags persist longer in EPI images. EPI is likely to be used to assess myocardial motion further into diastole since EPI contrast at 600 ms is equal to SPGR contrast at 200 ms. In EPI, slices are interleaved and each slice is excited only once in each cardiac cycle. The decrease of tag contrast in later cardiac phases is caused only by the spin-lattice relaxation as shown in Eq. [8]. In SPGR, a slice is excited many times in each cardiac cycle. During each excitation, a fraction of magnetization in both tagged and nontagged regions is brought into the transverse plane, reducing the tag contrast. The decrease of tag contrast is caused not only by the spin-lattice relaxation but also by the saturation effects of RF pulses as shown in the exponential and cosine terms of Eq. [5]. If a large flip angle is used, the tag contrast will disappear quickly because of the saturation effect. If a small flip angle is used, the saturation effect is minimized and the decrease of tag contrast is dominated by spin-lattice relaxation. To increase the persistence of tag contrast, an optimal flip angle (10), typically around 10° – 20° , or variable flip angles (13), can be used.

The total imaging time for a stack of multi-phasic images with EPI is 32 heartbeats and a single breath-hold. This is much shorter than the imaging time required by SPGR, which takes 152 heartbeats and eight breath-holds. Because all of the EPI images are obtained in a single breath-hold, the respiratory motion and registration artifacts that result from multiple breath-holds are eliminated. If a patient cannot manage a breath-hold for 32 heartbeats, two 16 heartbeat breath-holds can be used, but the advantage of a single breath-hold is lost. SPGR images are very robust to chemical shift artifacts and short T_2^* values. This cannot be said for EPI acquisitions, which require more careful tuning.

The temporal resolution of segmented k -space SPGR depends on the number of phase encode views per movie frame. It is clear that using the fast gradients will improve either the time resolution or the SNR of the segmented k -space images by reducing the time required to play out the “dead-time” pulses such as the phase encode pulse and the slice rewinder. If the fast gradients are used, either the temporal resolution can increase approximately 25% when the readout bandwidth is fixed, or the tag CNR can increase approximately 40% by using the optimal bandwidth when the temporal resolution is fixed (14).

In conclusion, EPI can acquire the same amount of myocardial tagging data in a shorter time than SPGR. The tag contrast in EPI is much higher than in SPGR and permits tag visualization much further into diastole.

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FIG. 1.

The SPGR and EPI pulse sequences are shown in (a) and (b), respectively. The tagging pulse is ECG-triggered and is followed by multiple imaging pulses. The tag lines are placed perpendicular to the readout direction. The SPGR sequence is made as short as possible. A spectral-spatial excitation RF pulse is used in EPI to reduce chemical shift artifacts.

FIG. 2.

The tag contrast in SPGR and EPI images as a function of time after the ECG trigger. The solid and dotted lines represent the theoretical predictions of SPGR and EPI tag contrast, respectively. The crosses and circles represent the tag contrasts of SPGR and EPI images, respectively.

FIG. 3.

The tag contrast in SPGR and EPI images when T_2^* is 20 ms. The solid and dotted lines represent the theoretical predictions of SPGR and EPI tag contrast, respectively. The tag contrast of EPI images is 100% and 260% higher than that in SPGR images in 30 and 330 ms after the ECG trigger, respectively. The tag contrast of EPI image in 600 ms after ECG trigger is approximately equal to the contrast of SPGR image in 200 ms after ECG trigger.

FIG. 4. Comparison of tag contrast with SPGR (a) and EPI (b) images. The left column shows images acquired 30 ms after the ECG trigger. The right column shows the same slice acquired at 330 ms after the ECG trigger.

FIG. 5.

The tag contrast of myocardium measured in the series of SPGR and EPI images shown in Fig. 4. The tag contrast of EPI images is about 70% and 200% higher than SPGR images at 30 and 330 ms after the ECG trigger, respectively.