33rd Annual International Conference of the IEEE EMBS
Boston, Massachusetts USA, August 30 - September 3, 2011

An Improved Real-time Cine Late Gadolinium Enhancement (LGE) Imaging Method at 3T

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***Abstract*—A real-time Late Gadolinium Enhancement (LGE) MRI technique (free breathing and non-gated) is presented for detection of myocardial scars. Conventional LGE imaging methods currently in use are applied in conjunction with breath-hold and, thus, are difficult to use in patients with cardiac disease and may lead to motion artifacts. Additionally, conventional techniques involve ECG gating, which is problematic in patients with arrhythmias requiring multiple breath holds and use of arrhythmia rejection techniques. Finally, conventional LGE techniques require accurate estimates for the inversion time in order to null the normal myocardium, revealing the location of the scar with high contrast. Real-time LGE imaging obviates these difficulties and can, in principle, acquire cine images to assess wall motion over several heart phases as part of the same scan. To date, the main limitation of real-time LGE imaging has been long acquisition window and low temporal resolution. These limitations lead to temporal blurring of wall motion and possible overestimation of infarct size. The goal of this study was to increase the temporal resolution of real-time, cine LGE imaging, providing the possibility for better visualization of the wall motion and more accurate assessment of myocardial viability.**

I. Introduction

Late-Gadolinium Enhancement MRI (LGE-MRI) is the gold standard method for differentiation between the injured and normal myocardium [1]-[4]. Development of LGE-MRI techniques and its following emerging applications in clinical cardiology parallels the increasing importance of cardiovascular MRI in routine clinical patient care.

The main approach to LGE-MRI is based on injection of a T_1 shortening contrast agent such as gadolinium and subsequently waiting for 10-30 minutes before acquiring an LGE image using a specific inversion recovery pulse sequence. The timing between contrast injection and data acquisition (inversion time) is very important and can affect the image contrast between the normal myocardium and the infarcted area [3]. In addition, conventional LGE-MRI methods currently in use are applied in conjunction with breath-hold and, thus, are difficult to use in patients with cardiac disease and commonly lead to motion artifacts. Another problem with the conventional techniques is that they involve ECG gating, which is problematic in patients with arrhythmias, requiring multiple breath holds and arrhythmia rejection. Several attempts such as navigator-assisted free-breathing [5] and single-shot techniques [6] were made to circumvent the need for breath-hold imaging. However these methods were still ECG-gated.

Real-time LGE imaging which is a free-breathing, non-gated technique was previously proposed to encounter these difficulties. In the past, real-time LGE was implemented to acquire few cine images (on the order of 5 in the R-R interval) based on steady state free precision (SSFP) in order to assess wall motion in addition to viability as part of the same scan [7],[8]. A clear limitation of this approach however is its long acquisition window and low temporal resolution since it leads to temporal blurring of wall motion and possible overestimation of infarct size. With 5 images acquired on one cardiac cycle, accurate timing parameters such as inversion time (TI) for acquisition of an image with completely nulled-myocardium can be a challenge. To shorten the acquisition window, parallel imaging methods such as SENSE [9], TSENSE [10] and GRAPPA[11] may be utilized. However, parallel imaging is limited by low SNR which could be countered by working at a higher field strength [12],[13].

The goal of the present study was to increase the temporal resolution and image quality of real-time LGE and thus providing the possibility for better visualization of the wall motion and more accurate assessment of myocardial viability. Given sufficient temporal resolution, calculation of inversion time (TI) which is a difficult step with both conventional and previously proposed real-time LGE imaging techniques is no longer necessary with the proposed method.

II. Materials and Methods

A. Pulse sequence

Fig. 1 demonstrates the schematic of proposed sequence. In this sequence, images were acquired continuously after the inversion pulse and accurate calculation of TI was no longer necessary. However, the TI For complete myocardial

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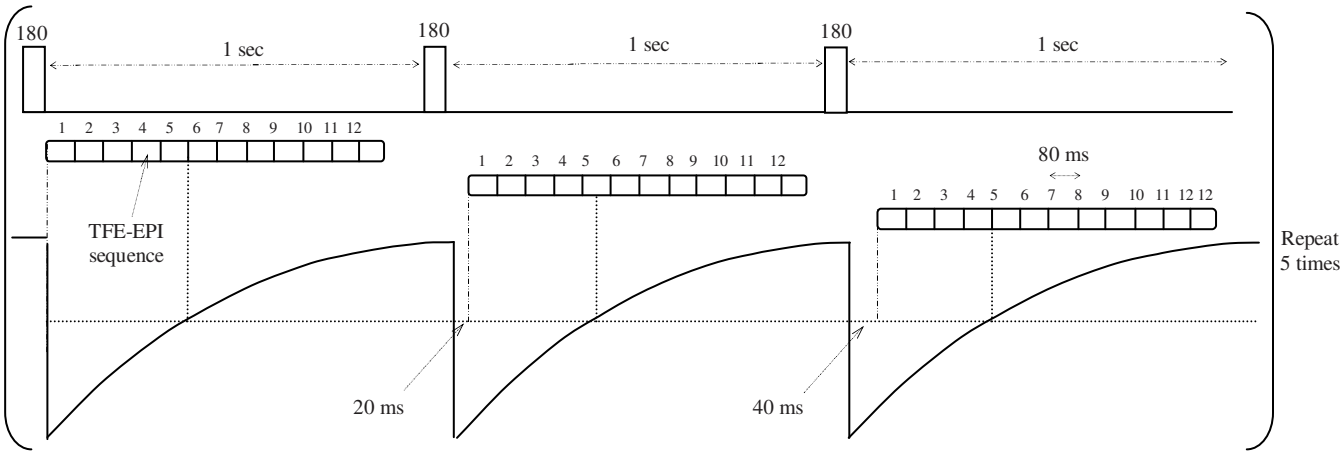

Fig. 1. Schematic of proposed pulse sequence. An inversion (180 deg.) pulse is applied followed by acquisition of 12 images after a trigger delay between the inversion pulse and first image. This trigger delay is changed after each inversion pulse in order to acquire a nulled-myocardium image with highly accurate inversion times. The same three scans are repeated 5 times (with no gating) to ensure acquisition of a nulled-myocardium image with a mid-diastolic cardiac phase.

nulling will depend on the length of acquisition window. To compensate, three dynamic scans were obtained consecutively, and for each dynamic scan, the trigger delay after inversion pulse was increased by approximately 20 ms. Consequently, in one of the three dynamic scans an image with accurate TI is achievable.

Conventional LGE images are collected near mid-diastole using ECG-gated sequences to take advantage of the fact that the heart has less motion in this period of the cardiac cycle. However, ECG-gated sequences are more problematic in the case of patients with arrhythmias and additionally require longer scan times. This is the main motivation for applying non ECG-gated techniques using real-time imaging. Since the real-time LGE sequence is non-gated, there is no guarantee that the nulled-myocardium image is acquired during mid-diastole. In order to ensure that at least one LGE image is collected near a more conventional mid-diastolic cardiac phase, the three dynamic scans, shown in Fig. 1, were each repeated 5 times. These random repetitions, performed manually by the operator at different times, results in various starting points for image acquisition and leads to collection of LGE images at different cardiac phases. In particular, the operator should be able to select one dynamic scan from the data sets collected which has the nulled-myocardial LGE image in the mid-diastolic phase. In the proposed protocol, the resulting imaging time was about 15 sec, resulting from 3 dynamic scans and 5 repetitions for a total of 15 cine LGE data sets.

Imaging was performed using a 16 channel phased array coil (Achieva TX, Philips Healthcare, Best, NL). Our real-time LGE sequence was demonstrated based on a single-shot inversion recovery TFE-EPI sequence. The TFE-EPI was implemented with EPI factor=7 and TFE-factor=14. EPI factor determines the number of *ky* profiles collected per excitation and TFE-factor determines the number of shots needed for acquiring the whole k-space. Therefore, the product of the TFE factor with the EPI factor results in the number of *ky* profiles collected. Generally, TFE-EPI sequences are more prone to B_0 inhomogeneity and susceptibility artifacts. Nonetheless, these artifacts are more dependent on the EPI factor used in the sequence. For small EPI factors, these artifacts are less severe. In addition to applying a relatively small EPI factor, volume shimming in the region of interest could avoid possible B_0 inhomogeneity artifacts appearing in TFE-EPI sequences. The images were acquired similar to previous studies [7],[8] with the following imaging parameters: FOV= 250x340 mm, matrix size= 120x160, TR/TE= 5.7/1.9 ms, flip angle= 25 deg., spatial resolution= 2x2x8 mm, 75% k-space acquisition in the phase encode direction. SENSE parallel imaging (acceleration factor = 1.7) was employed to reduce the length of the acquisition window. The sensitivity of coils were collected using a reference scan, available on the scanner,, before collecting the data. The length of the acquisition window which was achieved was about 80 ms, permitting collection of 12 cine frames in one heartbeat (assuming a heart rate of about 60 bpm). With view-sharing, it should be possible to increase the number of frames to over 20 frames per cardiac cycle.

B. Experimental method

Two types of studies were performed to investigate the proposed sequence. Firstly, a series of T1-weighted images were collected by imaging a phantom containing ten tubes filled with different concentration of Gd-DTPA to create different T1 values. In this phantom with heterogeneous T1, acquired images show the sensitivity of the proposed sequence to T1 variations similar to both normal and infarcted myocardium. Note that this phantom does not mimic the dynamic nature of heart. It only mimics T1 variation of tissues in the heart.

Secondly, a series of single-shot short-axis images were collected in a healthy volunteer to assess visualization of cardiac wall motion and achievable temporal resolution. The sequence parameters for both in-vivo study and phantom study were selected identical and similar to

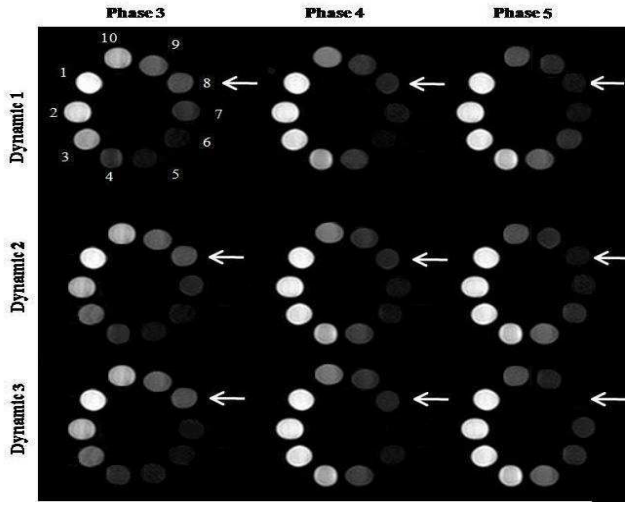

Fig. 2. T1-mapping phantom and the result for phases 3-5 (from a total of 12 frames) and 3 cine dynamic scans corresponding to different trigger delays as shown in Fig. 1. Arrows point to the tube corresponding to T1 value similar to those of normal myocardium.

previous real-time studies. Twelve frames are acquired in each cardiac cycle including nulled-myocardium LGE image and other frames from systole and diastole cardiac phases.

III. RESULT AND DISCUSSION

Fig. 2 demonstrates images of conical tubes with different concentrations of Gd-DTPA. Each row corresponds to one dynamic scan depicted in Fig. 1 and consists of images from 3 different "cardiac phases" from 12 phases that were collected. Note that tubes are numbered 1-10 where each tube has a specific T1 value based on its gadolinium concentration. T1 values for tubes 1-3 are very short and appear bright in all cardiac phases. Other tubes were sorted according to their T1 values in Fig. 3. Tube 8 has T1 value similar to those of normal myocardium. The optimal TI was achieved in dynamic 3, phase 5 when the signal from this tube is completely nulled. The temporal intensity variations of the tubes in different phases show the potential capability of this technique to distinguish between different tissues in the human heart.

Fig. 3 shows the signal intensity curve acquired for nine tubes at a simulated heart rate of 60 bpm. The T1 value calculated using these curves were compared to reference values of T1 for each tube. There was good agreement between estimated T1 values and reference values for each tube. This agreement proves that the proposed sequence achieved an accurate T1 measurement in phantom of heterogeneous T1 values and shows the sensitivity of the proposed sequence to various T1 values.

In the proposed acquisition technique, there is no need to manually calculate TI as the image displaying the highest contrast between the normal and infracted myocardium is immediately recognizable (e.g., phase 4, dynamic scan 3 in Fig. 2).

Fig. 4 displays a free-breathing, non-gated acquisition with the proposed sequence in a normal volunteer clearly depicting both wall motion and T1 dependent signal variation as with the phantom study. We should highlight the fact that the number of acquired images in the in-vivo study is more than twice the number in previously published real-time LGE methods (in [7],[8] the maximum number of acquired phases was 5). With 12 phases, wall motion can now be analyzed with a higher accuracy. In comparison to conventional cine images for assessment of wall motion collected over many cardiac cycles, the real-time method discussed here has acceptable spatial and temporal resolutions and, thus, the sequence may potentially be utilized for real-time Cine imaging (see for example [15].)

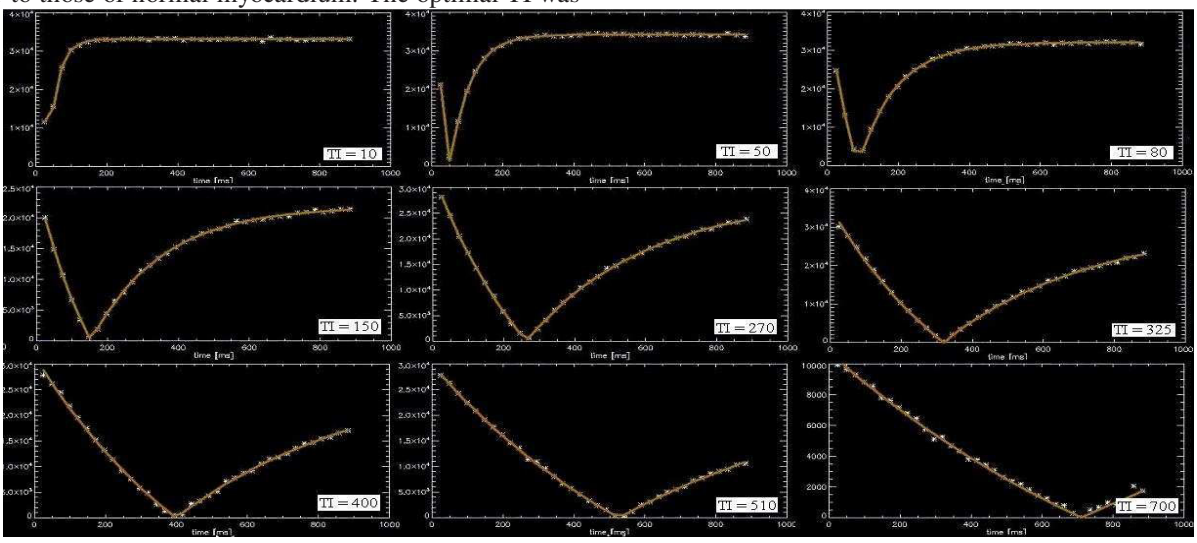

Fig. 3. Signal intensity curve obtained from tubes 2-10 in phantom. Inversion time for each tube has been shown and changes in the range of 10-700 ms. The first row curves correspond to tubes 2-4 which have a short T1 and recover fast appearing bright in fig.2 from the very first phases. Third curve in second row and first curve in third row correspond to the tube with similar T1 to a myocardial scar and normal myocardium respectively.

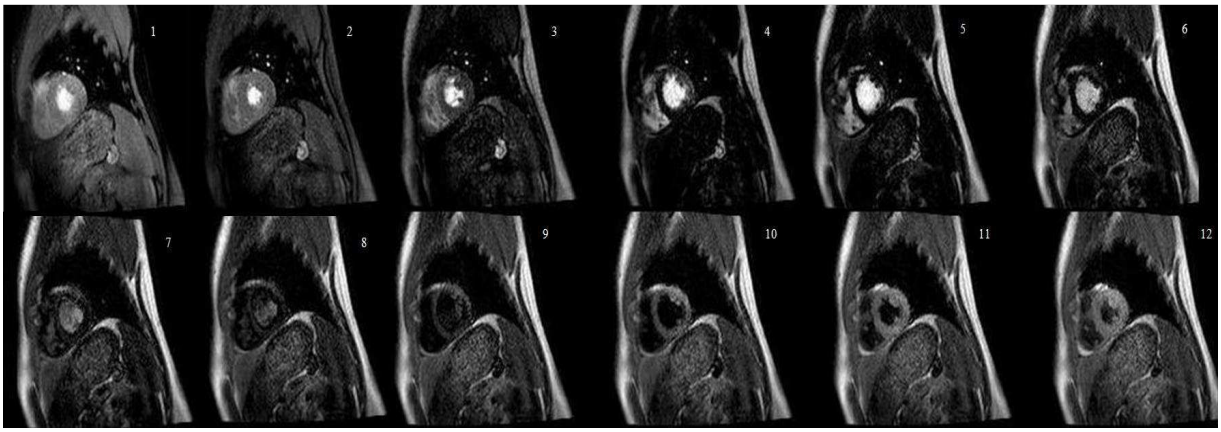

Fig. 4. Real-time LGE imaging in a 29 years old normal volunteer achieving frames in 12 cardiac phases within a single heart beat. Image 5 shows the image with nulled- myocardium.

IV. Discussion and conclusions

The main limitation of previously proposed real-time LGE methods was the large acquisition window and low temporal resolution. The proposed method counters this limitation by reducing the length of the acquisition window, thereby increasing the temporal resolution and number of cardiac phases which can be imaged. The underlying reason for the achievable improvement in real-time imaging is use of parallel imaging in conjunction with TFE-EPI at 3T.

Generally, application of parallel imaging to shorten the acquisition time causes degradation of SNR. Imaging at 3T permits parallel imaging with higher acceleration factors with higher SNR. In addition, at 3T, T1 values are higher leading to greater sensitivity for T1 changes [14]. Finally, imaging at 3T permits shorter TR's – therefore, image acquisition speed is greater and acquisition window is shorter than at 1.5T. Disadvantages of imaging at 3T include banding artifacts in SSFP imaging as well as susceptibility artifact.

Previous real-time LGE techniques are based on SSFP pulse sequence. However, SSFP sequence suffers from insufficient temporal resolution [7],[8]. Additionally, off-resonance or banding artifact in the SSFP cine images as noted above degrades the image quality. TFE-EPI sequence, employed in this paper, can improve the temporal resolution and overall image quality compared to the SSFP sequence with identical parameters [15].

With the proposed sequence, the free-breathing scan time to collect all 12 cine LGE data sets is about 15 seconds which is sufficiently short to be comfortable even for the very sick of patients. To further improve the temporal resolution, view-sharing and parallel imaging may be used concurrently to increase the number of frame to over 20 images in one cardiac cycle.

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