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Suppression of Ghost Artifacts Arising from Long T_1 Species in Segmented Inversion-Recovery Imaging

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Abstract

Purpose: We demonstrate an improved segmented inversion-recovery sequence that suppresses ghost artifacts arising from tissues with long T_1 (> 1.5 s).

Theory and Methods: Long T_1 species such as pericardial fluid can create bright ghost artifacts in segmented, inversion-recovery MRI because of oscillations in longitudinal magnetization between segments. A single dummy acquisition at the beginning of the sequence can reduce oscillations; however, its effectiveness in suppressing long T_1 artifacts is unknown. In this study, we systematically evaluated several test sequences, including a prototype (saturation post-pulse readout to eliminate spurious signal: SPPRESS) in simulations, phantoms, and patients.

Results: SPPRESS reduced artifact signal $90\% \pm 25\%$ and $74\% \pm 28\%$ compared with Control and Single-Dummy methods in phantoms. SPPRESS performed well at 1.5 Tesla (T) and 3T, with steady-state free precession (SSFP) and fast low-angle shot (FLASH) readout, with conventional and phase-sensitive reconstruction, and over a range of physiologic heart rates. A review of 100 consecutive clinical cardiac MRI scans revealed large fluid collections (eg, regions with long T_1) in 14% of patients. In a prospectively enrolled cohort of 16 patients with visible long T_1 fluids, SPPRESS appreciably reduced artifacts in all cases compared with Control and Single-Dummy methods.

Conclusion: We developed and validated a new robust method, SPPRESS, for reducing artifacts due to long T_1 species across a wide range of imaging and physiologic conditions.

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Dr. Raymond J. Kim is an inventor on a US patent on Delayed Enhancement MRI, which is owned by Northwestern University. Dr. Wolfgang G. Rehwald is an employee of Siemens Healthcare. Drs. Elizabeth R. Jenista, Wolfgang G. Rehwald, Enn-ling Chen, and Raymond J. Kim are inventors on a US patent application on Long- T_1 Artifact Suppression, which is owned by Duke University and Siemens Healthcare.

Keywords

delayed contrast-enhancement MRI; late gadolinium enhanced MRI; artifacts; cardiac MRI

INTRODUCTION

Delayed contrast-enhanced MRI (DE-MRI) with electrocardiogram (ECG)-gated inversion recovery (IR) is considered the reference standard for imaging myocardial infarction (MI) (1-3) and serves as a core component of the cardiac MRI exam. In segmented DE-MRI, k-space is acquired in segments distributed over several heartbeats, requiring repeated IR pulses with a relatively short recovery time, TR_{EFF} (effective repetition time) (Fig. 1a). Usually, this does not result in any problems because DE-MRI is performed after the administration of gadolinium-based T_1 shortening contrast agents, and tissue magnetization is at steady-state for each readout segment (Fig. 1b). However, in tissues that do not take up contrast and have long T_1 times similar to or longer than TR_{EFF} (ie, more than 1.5–2 s), the time between IR pulses is insufficient for longitudinal magnetization to recover. As a result, there are oscillations in the magnetization (Fig. 1c), which manifest as ghost artifacts in the image (4).

Regions with long T_1 are commonly encountered in patients undergoing cardiac MRI and include pericardial effusion, pleural effusion, gastric fluid, cysts, cerebrospinal fluid (CSF), and saline breast implants. The ghost artifacts that arise from these tissues may overlay cardiac structures, reduce image quality, and result in erroneous interpretation. One method to reduce the impact of ghost artifacts is to acquire a second image with the same parameters, except for a change in the phase-encode direction. This will move the location of the ghosts but at the cost of additional imaging. Moreover, if the region with long T_1 is large or surrounds the heart (eg, circumferential pericardial effusion), some portion of the heart may be obscured by artifact, regardless of the phase-encode direction. A simple method to directly suppress ghost artifacts is to place a saturation band over the region with long T_1 immediately prior to readout. However, if the region is large or surrounds the heart, multiple saturation bands may be necessary; even then, these bands may be insufficient to eliminate all regions with long T_1 without overlap of the saturation pulses with vital structures of interest.

A straightforward method to reduce transient signals in the approach to steady-state is to use dummy acquisitions to discard signal acquired during the transient response. Although this increases imaging time, this is a general method that will reduce ghosts caused by a variety of mechanisms. Some vendors incorporate a single dummy acquisition as part of the standard sequence for delayed-enhancement imaging. Unfortunately, the efficacy of this approach in reducing ghost artifacts due to long T_1 species is unknown.

Kellman et al. (6) describe a method to reduce long T_1 ghost artifacts caused by CSF by employing phase-sensitive inversion recovery (PSIR) reconstruction with phased-array coils. This method relies on B_1 -weighted phased-array combining, which scales down signal with increasing distance from the coil. Thus, for small localized fluids adjacent to only one or a limited number of coils, such as CSF, the result is a reduction of ghosting signal far from the

coil, while also preserving the true signal close to the coil. The efficacy of this method in the situation when the long T_1 species is central in the image (ie, near the heart) and produces signal on multiple coil elements is unclear.

In this study, we test several methods to reduce long T_1 ghosts in the context of segmented inversion-recovery imaging. These methods were systematically compared in simulations, phantoms, and patients across a wide range of imaging conditions.

THEORY

We examined strategies to eliminate oscillations in the magnetization of long T_1 species between readout segments. Diagrams of each suppression method, along with control, are provided in Figure 2. For each scheme described below, the behavior of the longitudinal magnetization was calculated using the Bloch equations, assuming that the effects of readout pulses, diffusion, and T_2 relaxation were negligible.

Standard Vendor Implementation with Single Dummy Acquisition (Single-Dummy)

The normalized magnetization after n IR pulses is:

$$\frac{M_z(n * TR_{EFF})}{M_0} = 1 + \sum_{k=1}^{k=n} \left[(-1)^k 2e^{-\frac{kTR_{EFF}}{T_1}} \right] \quad [1]$$

where n is the integer number of IR pulses; TR_{EFF} is the time between IR pulses; $M_z(n*TR_{EFF})$ is the longitudinal magnetization at time $n*TR_{EFF}$, which is immediately prior to the next IR pulse; and M_0 is the unperturbed, initial M_z .

Equation [1] shows that the magnitude of the oscillations between segments decays exponentially with n , indicating that the largest change occurs between the first and second IR pulse (Fig. 2b). Although additional dummies reduce oscillations further, this is at the cost of two additional heartbeats ($1 TR_{EFF}$) per dummy. This would quickly exceed the breath-holding capacity of patients.

Saturation Pre-Pulse

A different approach would be to prime the magnetization before the conventional segmented IR sequence by applying a single saturation Pre-Pulse (with crusher gradients), followed by an appropriate time delay. The time delay, T_S , is set such that the Pre-Pulse has negligible effects for short T_1 species (due to rapid magnetization recovery), whereas magnetization is rapidly pushed to steady-state for long T_1 species (Fig. 2c) (7). To calculate T_S for a given long T_1 , we set the magnetization after the saturation Pre-Pulse followed by T_S to be equal to the magnetization after the first IR pulse followed by TR_{EFF} , and then we solve for T_S :

$$T_s = -T_1 \ln \left(\frac{2e^{-\frac{TR_{EFF}}{T_1}}}{1 + e^{-\frac{TR_{EFF}}{T_1}}} \right) \quad [2]$$

With the saturation Pre-Pulse (Pre-Sat) method, the magnetization prior to the n^{th} IR pulse becomes:

$$\frac{M_z(n * TR_{EFF})}{M_0} = \frac{2}{e^{\frac{TR_{EFF}}{T_1}} + 1} + 1 \quad [3]$$

With this derivation, there are no oscillations in magnetization between segments (right-hand side of Eq. [3] is independent of n). Unfortunately, this assumes only a single T_1 value and will not be accurate if multiple long T_1 species are present. Moreover, in clinical practice the exact T_1 value of a long T_1 species present in an individual patient is not known a priori. Therefore, to provide a practical method for implementation, we took the first term of the Laurent expansion for Equation [2] to obtain a simple approximation for T_s , which is independent of T_1 : $T_s = 0.5 * TR_{EFF}$. Note, this implementation of T_s also provides a practical way to account for cardiac gating because T_s is equal to one heartbeat (recall that $TR_{EFF} = 2$ heartbeats).

Combining Saturation Pre-Pulse With Dummy Acquisition (Pre-Sat-Plus-Dummy)

The pre-sat module with calculated delay T_s only approximates the steady-state condition, and some residual oscillations are expected. Combining this module with a dummy acquisition could reduce oscillations further (Fig. 2d). In this scenario, the magnetization after n IR pulses:

$$\frac{M_z(n * TR_{EFF})}{M_0} = 1 + 2 * \sum_{k=1}^{k=n} \left[(-1)^k e^{-\frac{kTR_{EFF}}{T_1}} \right] + (-1)^{n+1} e^{-\frac{(2n+1) * TR_{EFF}}{2T_1}} \quad [4]$$

The cost of this method is three heartbeats of additional breath-hold time without data acquisition.

Saturation Post-Pulse Readout to Eliminate Spurious Signal (SPPRESS)

If multiple regions with long T_1 are present (eg, pericardial and pleural effusion and CSF), it would not be expected that the T_1 values are identical. A limitation of the methods described above is that they do not directly account for the separate evolutions of magnetization for multiple T_1 values. Applying a saturation pulse after each readout acquisition would reset the total magnetization after each segment (Fig. 2e). As a result, each T_1 species recovers from $M_Z = 0$ during every TR_{EFF} . After incorporating a leading saturation post-pulse readout to eliminate spurious signal (SPPRESS) module with dummy readout, this consistent evolution should significantly suppress oscillations in magnetization between segments for all T_1 values.

If we define T_{SR} as the time after the postreadout saturation pulse to the next IR pulse, the magnetization after n IR pulses is simply:

$$\frac{M_z(n * TR_{EFF})}{M_0} = 1 - e^{-\frac{T_{SR}}{T_1}} \quad [5]$$

Hence, the magnetization is not dependent on n , and unlike the pre-sat module, steady-state can be achieved for multiple T_1 values simultaneously. The time cost of this method is two heartbeats, the same as the single-dummy method. We note that with a PSIR acquisition there are two readout events every TR_{EFF} : one at TI and a second in the next heartbeat to obtain phase data. In this situation, the post-readout saturation pulse only is applied to the first of the pair to maintain the same magnetization recovery time (to the next IR pulse) as without PSIR acquisition.

METHODS

Simulation Studies

The performance of each suppression method was estimated by calculating the change in longitudinal magnetization (M_Z) between the first and second readout (RO1 and RO2) using Equations [1] through [5] above. Simulations were performed with MATLAB (MathWorks Inc, Natick, MA) using the following parameters: $T_1 = 3,000$ ms (approximate T_1 of distilled water at 1.5 Tesla (T) (8)); $TR_{EFF} = 1,500$ ms (equivalent to a heart rate of 80 beats per minute (bpm) with a trigger pulse of 2); and $TI = 300$ ms (approximate TI used to null normal myocardium in conventional DE-MRI).

A second simulation in a digital phantom (Bloch equations) was performed to estimate the magnitude of artifact signal for each suppression method over a range of T_1 and TR_{EFF} values (eg, simulated heart rates). The digital phantom consisted of two test tubes of fluid: one with short T_1 (150 ms) and the other with variable T_1 (100 to 3,000 ms). Gaussian white noise was added to the images to allow for artifact-signal-to-noise ratio (SNR) calculations, which was measured as mean signal in the ROI adjacent to the test tube with variable T_1 divided by the standard deviation (SD) of the signal in the ROI adjacent to the test tube with short T_1 (Fig. 3a). Noise was measured in the region adjacent to the test tube with short T_1

because it was unlikely to have ghosting artifacts. For these simulations, a FLASH (fast low-angle shot) readout with a segmented, interleaved acquisition was assumed with the following parameters: TR = 5 ms, inversion time (TI) = 300 ms, lines per segment = 15, flip angle = 15 °, matrix = 128 × 128. Three heart rates (60, 80, and 100 bpm, ie, TR_{EFF} = 2,000 ms; 1,500 ms; and 1,200 ms, respectively) were tested.

Phantom Experiments

The phantom consisted of four test tubes with different concentrations of gadopentetate dimeglumine (Gd-DTPA, Magnevist, Bayer HealthCare Pharmaceuticals, Whippany, NJ) embedded in a bath of 1% agarose gel. The T₁ values of the test tubes at 1.5T are listed in Figure 3b. Each suppression method was tested on the phantom using both FLASH and steady state free precession (SSFP) readouts at 1.5T (Avanto, Siemens Healthcare, Erlangen, Germany) and at 3T (Verio, Siemens Healthcare) with the TI set to null the bath (~300 ms). Imaging parameters for FLASH readout at 1.5T were: echo time (TE) = 2.33 ms, flip angle = 5 °, bandwidth = 201 Hz/px, 15 lines per segment, matrix 128 × 90, and field of view (FOV) = 400 mm. For SSFP readout, the parameters were identical except: TE = 1.6 ms, flip angle = 90 °, and bandwidth = 588 Hz/px. Parameters at 3T were identical except for TE (FLASH, 1.53 ms; SSFP, 1.09 ms) and bandwidth (FLASH, 399 Hz/px; SSFP, 1502 Hz/px). Images were reconstructed as either magnitude or phase-sensitive images (9). Again, three heart rates were tested (60, 80, and 100 bpm).

Phantom Image Analysis

For magnitude images, artifact SNR was measured in a similar fashion to that on the simulation studies. Regions of interest were placed outside the agarose gel bath in the phase-encode direction of the test tube with the longest T₁ (2900 ms; red box in Fig. 3b) and the shortest T₁ (150 ms; white box in Fig. 3b). Artifact SNR was calculated as mean signal in the artifact ROI divided by the SD of the signal in the noise ROI (Fig. 3b).

In the PSIR reconstruction algorithm, a baseline offset signal is added to allow for rectification of the polarity of magnetization, and the SD of signal in noise-only regions is no longer valid for estimating background noise. Hence, artifact SNR was calculated differently for phase-sensitive images. Artifact signal was calculated as the mean signal in the artifact ROI subtracted by the mean signal in the noise ROI (to account for the baseline offset). Background noise was estimated by measuring the SD of the signal in the test tube with the shortest T₁ fluid. This compartment had homogeneous signal for all suppression sequences, and although the SD of signal in this compartment overestimates background noise, it provides a useful upper bound to allow for comparisons between suppression techniques.

Two readers also qualitatively evaluated artifact conspicuity in the phantom images by visually ranking the images from the five methods from best (1, least conspicuous ghosts) to worst (5, most conspicuous ghosts) for each of the experimental conditions (eg, heart rate of 60, 80, or 100; FLASH or SSFP readout; magnitude or phase-sensitive reconstruction; 1.5T or 3T).

Patients

Two separate studies were performed. In the first, we retrospectively reviewed 100 consecutive cardiac MRI scans that were performed in our clinical service to estimate the rate of occurrence of long T_1 fluids in patients. Long T_1 fluids were deemed to be present if large pericardial or pleural effusions, saline breast implants, large collections of gastric fluids, or ascites were identified by cine SSFP imaging in the DE-MRI slice locations. For each patient, the size of the fluid collection visible within the imaging plane was measured on the image where its extent was largest.

In the second study, we prospectively enrolled 16 patients with visible long T_1 fluids to undergo segmented DE-MRI with each of the suppression methods (and control) to ascertain clinical performance. DE-MRI images were acquired at 1.5T 10 to 20 minutes after intravenous administration of gadolinium contrast (gadoversetamide, 0.1 mM/kg); TI was manually adjusted to null normal myocardium (10). Typical imaging parameters were: FLASH readout, TE = 3.45 ms, flip angle = 20° , bandwidth = 130Hz/px, 31 lines per segment, matrix = 256×192 , and FOV = 340 mm. Written informed consent was obtained in all patients, and the protocol was approved by the Duke University Health System Institutional Review Board.

Patient Image Analysis

Artifact conspicuity was qualitatively assessed by two independent readers (level-3 trained in cardiovascular MR) by visually ranking images from best (1) to worst (5) in a similar fashion to that performed in the phantom images.

Image SNR and contrast-to-noise ratio (CNR) were measured in SPPRESS and control images to determine if SPPRESS resulted in a measurable change. Regions of interest were placed in the anterior chest wall subcutaneous fat (eg, an area of homogeneous tissue with relatively short T_1); in non-hyperenhanced myocardial tissue; and in the air outside the body to measure fat, normal myocardium, and noise signal, respectively. In patients with MI ($n = 6$), an infarct ROI was also placed in hyperenhanced myocardial tissue. For all ROIs, regions with ghost artifacts were avoided, and the same ROIs drawn on control images were copied exactly onto SPPRESS images at the same location to reduce variability in measurements. Signal-to-noise ratio was defined as mean signal in the tissue ROI divided by the SD of the signal in the noise ROI. Contrast-to-noise ratio was defined as the difference between two tissue ROIs divided by the SD of the signal in the noise ROI.

Specific Absorption Rate

All studies in volunteers were performed under adherence to U.S. Food and Drug Administration guidelines for specific absorption rate (SAR) (< 4 W/kg averaged over the whole body for any 15-min period) and with appropriate radiofrequency (RF) power safety checks on the clinical scanner. The relative SAR of an RF pulse was determined by numerical integration of the square of the envelope of the pulse shape (11). The overall SAR of a sequence was estimated by summation of the SARs of the individual RF pulses within the sequence. The SAR of each suppression module sequence was expressed relative to control, which was defined as 100%.

Statistical Analysis

Continuous data are reported as mean $\pm$ SD. A generalized linear model analysis was performed to determine if artifact SNR was different with SPPRESS compared to other methods. Heart rate, readout type, and field strength were regarded as repeated measures in the generalized linear model. For qualitative measurements, scores from the two readers were averaged because Bland-Altman analysis did not show any significant differences between the readers for image quality rank for all of the suppression methods (all $P > 0.34$). Analysis of variance with repeated measures was used to compare SPPRESS with the other methods. Bonferroni correction was used to adjust multiple pairwise comparisons. Signal-to-noise ratio and CNR in the patient images were compared with a paired t test. All statistical tests were two-tailed; values of $P < 0.05$ were regarded as significant.

RESULTS

Simulations

Figure 2 shows the evolution of longitudinal magnetization over several TR_{EFF} for each suppression method. SPPRESS eliminated signal oscillation between the first and second readout (signal = 0% of M_0), suggesting that ghosting artifacts from long T_1 species would be suppressed. Pre-Sat-Plus-Dummy provided the next best reduction in oscillation (signal = 2.1% of M_0), followed by Pre-Sat (signal = 3.6% of M_0) and single-dummy methods (signal = 56.1% of M_0). Control resulted in a signal of 110.5% of M_0 .

Figure 4 shows the results of the digital phantom simulation. In general, artifact SNR increased with higher T_1 and lower TR_{EFF} values (higher heart rates). SPPRESS resulted in the lowest artifact signal for all tested T_1 and TR_{EFF} values. Pre-Sat-Plus-Dummy also performed well and provided stable artifact suppression over tested T_1 and TR_{EFF} values. For exploratory purposes, we also simulated the effects of using two consecutive dummy acquisitions (dummy $\times 2$). As expected, the inclusion of a second dummy reduced artifact SNR compared with the single-dummy method; however, the improvement was modest and there was an additional cost of two heartbeats of breath-hold time.

Phantom Imaging Experiments

The overall performance of the suppression methods is summarized in Table 1. All suppression methods resulted in a reduction in artifact SNR compared with control. Similar to that observed in the simulation studies, SPPRESS resulted in the lowest artifact SNR, and this was found independent of readout type, heart rate, and field strength (comparisons with other suppression methods, all $P < 0.05$). On average, SPPRESS reduced artifact signal $90\% \pm 25\%$ and $74\% \pm 28\%$ compared with Control and Single-Dummy methods.

Similar to the simulation results, Pre-Sat-Plus-Dummy resulted in the second lowest artifact signal for the phantom studies; however, there was a measurably larger artifact SNR compared with SPPRESS (13.2 ± 12.8 vs. 3.4 ± 1.1 ; $P = 0.03$). Figures 5a and 5b show a comparison between simulated and experimental phantom results for all techniques. Overall, there was a good correspondence.

Findings were similar for PSIR reconstructed images, and again SPPRESS resulted in the lowest artifact signal (Table 1). A comparison between magnitude and PSIR images are shown in Figures 5b and 5c. Ghosting was evident on the PSIR images with all suppression techniques except for SPPRESS. In general, faint ghosts on the magnitude images were more overt on the PSIR images. Table 2 summarizes the qualitative image-ranking scores. There were no significant differences in the rank ordering of suppression techniques between magnitude and PSIR images (all $P > 0.20$).

Patients

Retrospective review of 100 consecutive cardiac MRI scans demonstrated that large collections of long T1 fluids were present in 14 patients: pericardial effusion (n = 1), pleural effusion (n = 6), and gastric fluid (n = 7). The average size of fluid collections was $61 \pm 41 \text{ cm}^2$.

In the cohort of 16 prospectively enrolled patients in which all four suppression methods (plus control) were evaluated, subjects had one or more of the following visible within the imaging plane: pericardial effusion (n = 8), pleural effusion (n = 8), gastric fluid (n = 3), ascites (n = 1), or saline breast implants (n = 1). With control, long T1 ghosts were visible in every patient. With single-dummy, ghosts were eliminated in 25% of patients; and when detected, the ghosts were consistently less severe than with control. Overall, SPPRESS resulted in the best artifact suppression, demonstrating complete elimination of artifacts in 75% of patients. Table 2 summarizes the visual ranking scores in patients. Rank order was similar to that found in phantoms. Figure 6 shows typical images in patients with the various techniques.

SNR of fat, infarct, and normal myocardium were not significantly different between SPPRESS and control or SPPRESS and single-dummy methods ($P > 0.40$ for all pairwise comparisons). Likewise, CNR of infarct-to-normal myocardium was similar between all three methods (Table 3).

Specific Absorption Rate (SAR)

In all volunteers, images with each artifact suppression method were obtained without exceeding any SAR constraints. The SAR values of Single-Dummy, Pre-Sat, Pre-Sat-Plus-Dummy, and SPPRESS sequences relative to Control with FLASH readout were 115.6%, 100.1%, 115.7%, and 116.2%, respectively; relative to Control with SSFP readout, SAR values were 114.9%, 100.1%, 115.0%, and 115.5%, respectively.

DISCUSSION

In this study, we describe and validate a new method to suppress ghost artifacts that arise from long T1 species. In phantoms and patients, the prototype method SPPRESS provided a significant reduction in artifact signal compared with all other methods across a wide range of imaging conditions and physiologic heart rates.

Previous reports have described long T1 ghost artifacts; these provide some insights into its origin (4,6). Although the reports suggest some strategies to reduce the effect of the ghosts

or mitigate ghosts from small, localized sources (ie, CSF), a universal approach to eliminating long T_1 ghosts is not proposed. In addition, there does not appear to be any published data regarding the scale of the problem—whether this is a rare, isolated issue or potentially a common problem during routine clinical imaging. In this study, we found that of 100 consecutive patients referred for cardiac MRI, a nontrivial proportion (14%) had large collections of long T_1 fluid that were visible in the field of view. This, in conjunction with the finding in the prospective substudy that every patient with long T_1 fluid had visible ghosts with control imaging, suggests that long T_1 artifact represents an appreciable problem with routine cardiac MRI.

Of interest, we found that one of the simplest methods—the use of a single-dummy acquisition prepended to the segmented inversion-recovery sequence—was effective in reducing ghosts. In phantoms, artifact signal was typically reduced to less than one-half that of control. Likewise, in patients, artifacts were substantially reduced, although residual ghosts were still visible in many subjects (Fig. 6). Given its simplicity and effectiveness, it is not surprising that some vendors incorporate a single-dummy acquisition as part of the standard sequence for delayed-enhancement imaging.

Considering the improvement with a single-dummy acquisition, it is possible that two or more dummies may represent a simple solution to long T_1 ghosts. However, our simulations indicate that although additional dummies will reduce artifacts further (Fig. 4), the magnitude of artifact reduction decreases exponentially with the number of dummies, indicating the largest benefit comes with the first dummy. Moreover, each extra dummy carries a cost of two heartbeats of additional acquisition time. We estimate that it would take four to five consecutive dummy acquisitions to achieve the level of artifact reduction that is provided by SPPRESS, which has the same acquisition time cost as single-dummy.

Because a saturation pulse is played after each segment readout throughout the acquisition, there may be a concern that SPPRESS reduces SNR in addition to suppressing artifact signal. However, we note that for standard DE-MRI, the time between the IR pulse and readout (inversion time) is usually set to null normal myocardial signal. Hence, an immediate postreadout saturation pulse should have negligible effects on tissues with T_1 values near the T_1 of normal myocardium (~350–400 ms at 1.5T following gadolinium administration (12)) because signal is already nulled. For tissues with T_1 values shorter than normal myocardium, again there should be negligible effects given the rapidity of magnetization recovery. This supposition is consistent with our experimental finding that the SNR of subcutaneous fat (relatively short T_1 compared with normal myocardium) in SPPRESS images was not significantly different than in control images. For tissues with substantially longer T_1 than normal myocardium, these are the tissues that result in long T_1 ghosting. Although the SNR of these tissues may be reduced by SPPRESS, this may not be important for routine delayed-enhancement imaging because generally the focus is on myocardial regions with excess gadolinium accumulation and short T_1 .

We observed that long T_1 fluid results in substantial ghosting on both magnitude and PSIR reconstructed images. Although a previous study has shown that PSIR with phased-array combining can reduce long T_1 ghosts from small, localized sources (ie, CSF), the situation

in the current study was different in that regions with long T_1 were large (relative to the field of view) and often central in the image. Hence, regions with long T_1 were within the sensitive volume of multiple coil elements, and B_1 -weighted phased-array combining did not provide any ghost artifact suppression. Indeed, in many cases ghosts that were faint on the magnitude images were more conspicuous on the PSIR images (Fig. 5c, blue arrows). For conventional PSIR reconstruction, it is customary to perform surface coil intensity normalization, and this is achieved by dividing the phase-sensitive image by the reference image on a pixel-by-pixel basis. As a result, regions with low signal on the reference dataset will appear amplified on the final PSIR image. Regions with long T_1 ghosts are expected to have relatively lower signal on the reference image than on the primary magnitude image because the time from each inversion pulse to readout is greater (by one heartbeat) and oscillations in magnetization between imaging segments should be reduced.

Ghosting artifact arising from tissues with long T_1 is not limited to segmented cardiac DE-MRI but potentially is a problem with any segmented IR 2D or 3D sequence used for imaging various regions of the body with or without ECG-gating (13,14). Even without IR preparation, long T_1 ghosts may occur if the segmented sequence results in oscillations in the transient approach to steady-state. In this study, SPPRESS only was tested for cardiac imaging; however, the method may be more generally applicable and potentially could be used to suppress any artifacts arising from oscillations in magnetization of long T_1 species. Finally, ghosts do not occur with single-shot IR imaging. However, if a multislice stack of 2D images is acquired, regions with long T_1 can have alternating or variable image intensity from slice to slice, which can confound image interpretation. We have observed that in this situation SPPRESS can be useful to provide uniform image intensity throughout the multislice stack (Fig. 6c).

A limitation of this study is that all imaging was performed in sinus rhythm, and arrhythmias could create additional ghosting artifacts. One mechanism by which arrhythmias cause ghosts is by producing a variable TR_{EFF} , which in turn results in variable signal recovery and oscillations in magnetization. We speculate that SPPRESS could reduce these oscillations because the post-readout saturation pulse re-zeros magnetization after each segment. Hence, following an ectopic beat, the next segment will return to the original steady-state, minimizing the overall effect of the ectopic beat on image quality. This conjecture will require investigation. Additionally, although standard arrhythmia rejection approaches could be helpful, and SPPRESS is easily combined with these approaches, this was not investigated in this study.

The prevalence of large fluid collections found during routine cardiac MRI was relatively high (14%) and reflects only one institution's patient population. Although other institutions could have a lower prevalence, our results demonstrate that even in a relatively sick population, for which large pericardial and pleural effusions were not uncommon, SPPRESS could significantly improve DE-MRI image quality. Moreover, it is likely that smaller-sized fluid collections can still result in sufficient ghosting to degrade overall image quality and affect diagnostic accuracy.

CONCLUSION

In this study, we developed and validated a new method, SPPRESS, for reducing artifacts due to long T_1 species across a wide range of imaging and physiologic conditions. SPPRESS reliably reduced artifacts without sacrificing SNR or increasing scan time beyond that used for a single-dummy acquisition.

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FIG. 1.

Timing diagram of the segmented delayed contrast-enhanced MRI sequence, along with evolution of magnetization for short and long T_1 species. **(a)** Segments of image RO are acquired over several heartbeats, requiring repeated IR pulses separated by a recovery period (TR_{EFF}), which generally is 2 electrocardiographic RR interval, time between heart beats (R-waves). **(b)** If the T_1 of the imaged species is sufficiently short to allow for full magnetization recovery between imaging segments, the magnetization is at steady-state for each readout segment. **(c)** If T_1 is long, there is insufficient time between IR pulses to allow for full magnetization recovery, and oscillations occur between imaging segments. These oscillations cause a time-varying weighting of k-space, which manifests as ghosts in the phase-encode direction. IR, inversion recovery; RO, readout; TR_{EFF} , effective repetition time.

FIG. 2.

Comparison of 4 suppression methods with control. The timing diagram of the pulse sequence, along with the simulated longitudinal magnetization trajectories, is shown. The orange dots indicate the timing of acquisition of the center of k-space and highlight the degree of oscillation that is present for each method. Simulations were performed assuming a T_1 of 3,000 ms; a heart rate of 80 bpm; and an inversion time of 300 ms. The cost of each method in terms of additional breath-hold time is shown. See article for further details. IR, inversion-recovery pulse; Pre-Sat, saturation Pre-Pulse; RO, readout; Sat = saturation module (90° radiofrequency pulse with spoiler gradients); SPPRESS, saturation post-pulse readout to eliminate spurious signal; TI, inversion time; TR_{EFF} effective repetition time (between IR pulses); TS, saturation time (between Sat module and first IR pulse); recovery time between postreadout saturation pulse to the next IR pulse.

FIG. 3.

Diagram of the setup and analysis for simulations and phantom experiments. **(a)** The simulation setup (left panel) shows a crosssection of two test tubes in the middle of the image with either short T_1 (150 ms) or variable T_1 (100 to 3,000 ms) fluid. ROIs were chosen adjacent (in the phase-encode direction) to the test tubes to measure the signal of any ghosting artifacts, if present. Artifact-SNR was calculated by taking the mean signal in the ROI adjacent to the variable T_1 species (red box) and dividing by the standard deviation of the signal in the ROI adjacent to the short T_1 species (white box). In the simulated image (right panel), ghosts are present in the phase-encoding direction adjacent to the bottom test tube ($T_1 = 2,900$ ms), but none are visible adjacent to the top test tube ($T_1 = 150$ ms). **(b)** The phantom setup is shown in the left panel. It consisted of a 1% agarose gel bath with 4 test tubes containing water with different T_1 values controlled by the concentration of gadopentetate dimeglumine. A typical acquired image is shown in the right panel. The dashed line indicates the border of the agarose gel bath. As shown in the phantom setup diagram, artifact and noise ROIs were placed outside the agarose gel bath to measure artifact SNR. The T_1 of the tubes at 3T were 160 ms; 455 ms; 895 ms; and 3,120 ms. The T_1 of the bath was 310 ms at 3T and 300 ms at 1.5T. ROI, region of interest; SNR, signal-to-noise ratio; T, tesla.

FIG. 4.

Artifact SNR of the digital phantom simulation. The three panels represent findings for simulated heart rates of 60, 80, and 100 bpm. SPPRESS provided the lowest artifact signal for all T₁ values and heart rates that were tested. See article for details. Pre-Sat, saturation Pre-Pulse; Pre-Sat-Plus-Dummy, saturation Pre-Pulse with dummy acquisition; SNR, signal-to-noise ratio; SPPRESS, saturation post-pulse readout to eliminate spurious signal; TR_{EFF} effective repetition time (between IR pulses).

FIG. 5.

Comparison of simulated (a) and experimental phantom images either magnitude (b) or PSIR- (a) reconstructed. The images acquired in phantoms corresponded well with the simulated results. Ghosting of the long T_1 species (red arrows) was obvious with control, reduced with the various suppression methods, and absent with SPPRESS. Note that test tubes with long T_1 species are black on PSIR images because the TI was set to null the bath (~ 300 ms), and the magnetization vector is below the zero-crossing. Ghosts that appeared faint on the magnitude images were generally more overt on the PSIR images (blue arrows). See article for details. FLASH, fast low-angle shot; HR, heart rate; PSIR, phase-sensitive inversion recovery; Pre-Sat, saturation Pre-Pulse; Pre-Sat-Plus-Dummy, saturation Pre-Pulse with dummy acquisition; SPPRESS, saturation post-pulse readout to eliminate spurious signal.

FIG. 6.

Typical images in patients. **(a)** All 5 methods are compared in a 2-chamber long-axis view of the left ventricle in a patient with saline breast implants. Only SPPRESS resulted in elimination of ghost artifacts (red arrows). **(b)** Additional examples are shown in 2 patients. Severe ghost artifacts (red arrow) are observed with control; ghosts are significantly reduced but to a variable level with single-dummy; and ghosts are consistently eliminated with SPPRESS. **(c)** Example of signal intensity variation of long T_1 fluid in a multislice acquisition of single-shot images in a patient with a pleural effusion. With control, the signal intensity of the pleural effusion varies from slice to slice; however, with SPPRESS, the signal intensity is constant for all slices. See article for details. DE-MRI, delayed contrast-

enhanced MRI; Pre-Sat, saturation Pre-Pulse; Pre-Sat-Plus-Dummy, saturation Pre-Pulse with dummy acquisition; SPPRESS, saturation post-pulse readout to eliminate spurious signal.

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Table 1

Phantom Artifact–Signal-to-Noise Ratio

	Control	Single Dummy	Pre-Sat	Pre-Sat-Plus-Dummy	SPPRESS
Overall ^a	42.5 ± 19.3	19.8 ± 12.6	21.5 ± 15.7	13.2 ± 12.8	3.4 ± 1.1 ^b
Readout type ^c	FLASH	20.6 ± 2.3	5.8 ± 2.5	4.5 ± 0.4	2.9 ± 0.1 ^b
	SSFP	61.0 ± 5.5	14.7 ± 4.0	7.1 ± 3.1	2.3 ± 0.2 ^b
Heart rate ^d	60	11.9 ± 2.2	2.6 ± 0.6	2.6 ± 0.5	2.0 ± 0.4 ^b
	80	13.6 ± 2.1	7.7 ± 1.5	2.9 ± 0.6	1.9 ± 0.5 ^b
	100	14.8 ± 2.7	8.0 ± 1.5	5.7 ± 1.1	1.9 ± 0.5 ^b
	1.5T	20.6 ± 2.2	9.3 ± 4.7	5.8 ± 2.5	2.9 ± 0.1 ^b
Field strength ^e	3T	36.4 ± 18.3	17.0 ± 12.8	23.7 ± 17.7	3.7 ± 1.0 ^b
	PSIR ^f	32.2 ± 3.4	11.9 ± 7.5	10.5 ± 6.2	0.2 ± 0.3 ^b

^aResults are at 1.5T, averaged for the 3 heart rates, and with a FLASH or SSFP readout.^bIndicates statistical significance ($P < 0.05$) compared with all other methods.^cResults are at 1.5T and averaged for the 3 heart rates.^dResults are at 1.5T and with a FLASH readout.^eResults are with a FLASH readout and averaged for the 3 heart rates.^fResults are at 1.5T, with a FLASH readout, and averaged for the 3 heart rates.

FLASH, fast low-angle shot; Pre-Sat, saturation Pre-Pulse; Pre-Sat-plus-dummy, saturation Pre-Pulse with dummy acquisition; PSIR, phase-sensitive inversion recovery; SPPRESS, saturation post-pulse readout to eliminate spurious signal; T, Tesla.

Table 2

Image Rank Visual Scores

Suppression Method	Phantom		Patients Magnitude
	Magnitude	PSIR ^a	
Control	4.9 ± 0.2	5.0 ± 0.2	4.9 ± 0.3
Single-dummy	3.4 ± 0.5	3.3 ± 0.7	3.2 ± 0.7
Pre-Sat	3.4 ± 0.5	3.5 ± 0.6	3.1 ± 0.9
Pre-Sat-Plus-Dummy	2.3 ± 0.7	2.2 ± 0.6	2.3 ± 0.7
SPPRESS	1.0 ± 0.0 ^b	1.0 ± 0.0 ^b	1.5 ± 0.6 ^b

Numbers are reported as mean ± standard deviation.

Lower values represent better image quality (reduced artifact conspicuity).

^aIndicates no statistical difference between magnitude and PSIR images.

^bIndicates statistical significance ($P < 0.05$) compared with all other methods.

Pre-Sat, saturation Pre-Pulse; Pre-Sat-plus-dummy, saturation pre-pulse with dummy acquisition; PSIR, phase-sensitive inversion recovery; SPPRESS, saturation post-pulse readout to eliminate spurious signal.

Table 3

Patient SNR and CNR

Suppression Method	SNR			CNR	
	Infarct (n = 6)	Fat (n = 6)	Normal Myocardium (n = 16)	Infarct: Normal Myocardium (n = 6)	Fat: Normal Myocardium (n = 16)
Control	20.6 ± 13.3	24.7 ± 15.2	4.8 ± 2.3	15.5 ± 9.9	19.8 ± 13.2
Single-dummy	19.8 ± 12.5	24.6 ± 14.1	4.5 ± 3.3	15.8 ± 10.2	19.8 ± 11.6
SPPRESS ^a	20.1 ± 8.1	25.1 ± 15.8	4.7 ± 2.4	15.2 ± 6.2	20.2 ± 13.9

Numbers are reported as mean ± standard deviation.

^a All values for SPPRESS were not significantly different from single-dummy or control ($P > 0.40$ for all pairwise comparisons). CNR, contrast-to-noise ratio; SNR, signal-to-noise ratio; SPPRESS, saturation post-pulse readout to eliminate spurious signal.