Published in final edited form as:

Magn Reson Med. 2009 December ; 62(6): 1368–1378. doi:10.1002/mrm.22140.

Hybrid Adiabatic-Rectangular Pulse Train for Effective Saturation of Magnetization within the Whole Heart at 3T

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Abstract

Uniform T_1 -weighting is a major challenge for first-pass cardiac perfusion MRI at 3T. Previously proposed adiabatic B_1 -insensitive rotation (BIR-4) pulse and standard and tailored pulse trains of three non-selective pulses have been important developments but each pulse has limitations at 3T. As an extension of the tailored pulse train, we developed a hybrid pulse train by synergistically combining two non-selective rectangular RF pulses and an adiabatic half-passage pulse, in order to achieve effective saturation of magnetization within the heart, while remaining within clinically acceptable specific absorption rate (SAR) limits. The standard pulse train, tailored pulse train, hybrid pulse train, and BIR-4 pulse train were evaluated through numerical, phantom, and in vivo experiments. Among the four saturation pulses, only the hybrid pulse train yielded residual magnetization $< 2\%$ of equilibrium magnetization in the heart, while remaining within clinically acceptable SAR limits for multi-slice first-pass cardiac perfusion MRI at 3T.

Keywords

MRI; heart; cardiac perfusion; B_1^+ inhomogeneity; B_0 inhomogeneity; T_1 ; saturation; adiabatic pulse; SAR

Introduction

A recent study showed that the diagnostic performance of first-pass cardiac perfusion MRI is superior at 3T than at 1.5T, for the identification of both single and multiple vessel coronary artery disease (1). However, transmit radio-frequency (RF) field (B_1^+) variations and static magnetic field (B_0) variations are comparatively higher at 3T than at 1.5T (2–6). Specifically, the peak-to-peak variation of B_0 within the heart at 3T has been reported to be on the order of 130–260 Hz (3,5), and B_1^+ variation within the heart at 3T has been reported to be on the order of 50–80% (4,6). Increased B_0 and B_1 inhomogeneities make it more difficult to perform uniform T_1 -weighting at 3T using a conventional non-selective 90° pulse, which is known to be sensitive to both B_0 and B_1^+ inhomogeneities (7–9).

Non-uniform T_1 -weighting can confound both the interpretation and quantitative analysis of first-pass cardiac perfusion MR images. Given the limitations of commercially available automated B_0 shimming and RF calibration procedures for the heart, it is important to design

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robust saturation pulses that are insensitive to clinically relevant B_0 and B_1^+ variations within the heart at 3T. Previously proposed adiabatic B_1 -insensitive rotation (BIR-4)(9,10) pulse, RF pulse train (11) of three non-selective 90° pulses (9,12), and RF pulse train of three non-selective pulses with different flip angles (e.g., 96° , 228° , 141°)(13) can be used to improve the efficacy compared with a conventional non-selective 90° pulse, but at the expense of relatively longer pulse duration and higher specific absorption rate (SAR). For convenience, the pulse train of three non-selective 90° pulses will be referred to as the standard pulse train, and the pulse train of three non-selective pulses with different flip angles will be referred to as the tailored pulse train. The longer pulse duration (10–12 ms) is inconsequential because a typical perfusion image acquisition time is on the order of 120–150 ms. While the higher SAR is not a problem for both the standard and tailored pulse trains, it is a major limitation for a BIR-4 pulse (9) that consequently reduces the slice coverage in first-pass cardiac perfusion MRI. These saturation pulses have been important developments but each has limitations for performing uniform T_1 -weighting in multi-slice first-pass cardiac perfusion MRI at 3T.

We have implemented these previously proposed saturation pulses on a 3T whole-body MRI system, evaluated their performances in the human heart, and discovered that these pulses do not consistently yield complete saturation of magnetization within the heart while remaining within clinically acceptable SAR limits. As an extension of the tailored pulse train (11,13), we propose to synergistically combine two non-selective rectangular RF pulses with relatively low RF energy and a non-selective adiabatic half-passage (AHP) pulse (14,15) with relatively high RF energy, in order to achieve effective saturation of magnetization within the heart, while remaining within clinically acceptable SAR limits for multi-slice (≥ 3), first-pass cardiac perfusion MRI at 3T. Therefore, the purposes of this study were to develop a hybrid adiabatic-rectangular pulse train that can achieve both of the aforementioned objectives at 3T and evaluate its efficacy and predicted SAR against those of the standard pulse train, tailored pulse train, and BIR-4 pulses.

Methods

Saturation Pulses

A hybrid adiabatic-rectangular RF pulse train was designed with the following three subpulses: non-selective rectangular 140° pulse, non-selective rectangular 90° pulse, and non-selective AHP pulse (Fig. 1). The two non-selective rectangular pulse angles were numerically optimized by minimizing the root-mean-square (RMS) M_{ZR} for B_0 ranging from ± 130 Hz (1 Hz steps) (5) and B_1^+ scale ranging from 0.5 – 1.1 (0.01 steps)(6), with each non-selective rectangular pulse angle ranging from 80 – 200° (5° steps). The frequency sweep of the AHP pulse was ± 16.7 kHz, and its nominal B_1^+ calibrated by the transmit body coil was 667 Hz (14)(i.e., 49% larger than the 447 Hz frequency difference between the fat and water signal at 3T). The net effect of these subpulses is relatively strong immunity to clinically relevant B_0 and B_1^+ variations within the heart at 3T (see Figure 2 and Figure 3 for the numerical and phantom M_{ZR} maps).

For comparison, we implemented previously proposed standard and tailored pulse trains on a 3T whole-body MR scanner (Tim Trio, Siemens Healthcare, Erlangen, Germany), equipped with a gradient system capable of achieving a maximum gradient strength of 45 mT/m and a slew rate of 200 T/m/s. For reference, we implemented a train of two BIR-4 pulses (BIR-4 pulse train), as suggested by Cunningham et al. (16), in order to evaluate the efficacy of other saturation pulses. The frequency sweep of the BIR-4 pulse was ± 12.5 kHz, and its nominal B_1^+ calibrated by the transmit body coil was 500 Hz (14)(i.e., 12% larger than the 447 Hz frequency difference between the fat and water signal at 3T).

Using the pulse sequence simulator (IDEA, Siemens Healthcare, Erlangen, Germany), we calculated the transmit RF energy for each pulse, assuming that the RF voltage needed for B_1^+ of 500 Hz is 600 V. This RF calibration assumption is based on our extensive experience with cardiac MRI at 3T. The theoretically calculated transmit RF energy of the standard pulse train, tailored pulse train, hybrid pulse train, and BIR-4 pulse train was 10.8, 17.8, 25.3, and 105.8 J, respectively. The RF excitation was performed using the transmit body coil, and 6-element anterior and 6-element posterior phased arrays were used for signal reception. The RF energy deposition depends on several variables, including: linearity of the RF power amplifier, efficacy of the transmit body coil, and patient body. Therefore, we also calculated the relative RF energy of each pulse as its RF energy divided by the RF energy of the corresponding standard RF pulse train, which is then by definition a "1." The relative RF energy of the tailored pulse train, hybrid pulse train, and BIR-4 pulse train was 1.65, 2.34, and 9.80, respectively. Based on our previous experience with the standard pulse train and adiabatic pulses (9), we have empirically determined that the proposed hybrid pulse train, consisting of two non-selective rectangular pulses and one AHP pulse, does not exceed clinically acceptable SAR limits for multi-slice (≥ 6), first-pass cardiac perfusion MRI at 3T (see Table 3).

All three pulse trains were comprised of the same following gradient pulses in succession: 1 ms long spoiler gradients (magnitude of net zeroth gradient moment = 15 mT/m·ms), 3 ms long crusher gradients (magnitude of net zeroth gradient moment = 75 mT/m·ms), 2 ms long crusher gradients (magnitude of net zeroth gradient moment = 45 mT/m·ms), and 3 ms long spoiler gradients (magnitude of net zeroth gradient moment = 75 mT/m·ms). The BIR-4 pulse train used the same spoiler and crusher gradients, except that it used only the first set of crusher gradients (magnitude of net zeroth gradient moment = 75 mT/m·ms). The spoiler gradients are applied before and after each saturation pulse to dephase the transverse magnetization, and the crusher gradients in between RF pulses are cycled to eliminate stimulated echoes. Table 1 summarizes the RF pulse characteristics of the four different pulses.

Pulse Sequence

The M_{ZR} left behind by a saturation pulse can be measured by performing saturation-recovery (SR) ultra-fast gradient echo (i.e., TurboFLASH) imaging with "zero" recovery and centric k-space reordering, as previously described (9). Relevant imaging parameters include: field of view (FOV) = 340×276 mm, acquisition matrix = 64×52 , slice thickness = 8 mm, TE/TR = 1.1/2.3ms, saturation recovery time delay (TD) = 3 ms, image acquisition time = 78 ms, flip angle = 10° , generalized autocalibrating partially parallel acquisitions (GRAPPA)(17) with an acceleration factor = 1.5, and receiver bandwidth = 1002 Hz/pixel. Proton density (PD) image was acquired using 4° flip angle and without the saturation pulse, in order to perform image normalization (see the **Image Analysis** section for more details).

Numerical Simulation

The M_{ZR} left behind by each saturation pulse was simulated for B_0 ranging from ± 130 Hz (3 Hz steps)(5) and B_1^+ scale ranging from 0.5 – 1.1 (0.5 steps)(6). The numerical simulation was performed by ignoring T_1 relaxation between subpulses (i.e., during crusher gradient pulses), assuming complete dephasing of the transverse magnetization between subpulses, and with M_0 scalar equal to 1. For each non-selective rectangular pulse, its M_{ZR} was calculated using the Bloch equations in the rotating frame. Specifically, the effective field and effective nutation angle were calculated, as previously described (7), using a customized code programmed in MATLAB® R2008a (Mathworks, Natick, MA). Another customized software programmed in visual C++® 6.0 (Microsoft Corporation®, Bellevue, WA) was used to calculate the M_{ZR} left behind by adiabatic pulses. Specifically, the actual RF pulse profiles (i.e., amplitude and phase) used by the MR scanner were used to calculate the generalized solutions of the Bloch equations using the Laplace transformation, as previously described (18). The simulation for the adiabatic

pulses was performed by ignoring T_1 relaxation between subpulses, assuming adiabaticity, and with $T_1 = 1000$ ms, $T_2 = 40$ ms, and M_0 scalar equal to 1. For the standard pulse train, tailored pulse train, hybrid pulse train, and BIR-4 pulse train, their net M_{ZR} was calculated as the product of M_{ZR} of individual subpulses. For each saturation pulse, M_{ZR} was plotted as a function of B_0 and B_1^+ scales, and the corresponding RMS M_{ZR} was calculated.

Phantom Imaging

Phantom experiments were performed to evaluate the performance of the four saturation pulses. A spherical oil phantom ($T_1 = 530$ ms) was imaged to minimize the dielectric effects (19). The T_1 of this phantom was calculated from a separate multi-point SR experiment. The image acquisition was performed in an axial plane at magnet isocenter, in order to achieve relatively homogeneous B_1 excitation within the phantom using the body coil. The B_0 offset was manually adjusted to produce -130 , -65 , 0 , 65 , and 130 Hz off-resonance, in order to mimic clinically relevant B_0 variation within the heart at 3T (3,5). For each B_0 offset, the image acquisition was repeated with B_1^+ scale of each saturation pulse manually adjusted from 0.5–1.1 (0.05 steps) of its nominally calibrated B_1^+ value, in order to determine the sensitivity of each saturation pulse to clinically relevant B_0 and B_1^+ variations within the heart (3–6). For each saturation pulse, 60 M_{ZR} measurements were recorded using aforementioned B_0 and B_1 scales, with a repetition time of 4000 ms to allow for full recovery of magnetization between image acquisitions. For each saturation pulse, M_{ZR} was plotted as a function of B_0 and B_1^+ scales, and the corresponding RMS M_{ZR} was calculated.

Cardiac Imaging

M_{ZR} Measurements—Nine healthy human subjects (6 males; 3 females; mean age = 29.0 ± 4.7 years; mean weight = 68.8 ± 8.9 kg) were imaged in 3 short-axis views (apical, mid-ventricular, and basal) of the LV and three long-axis views (2-chamber view of LV, 2-chamber view of RV, and 4-chamber). For imaging the 2-chamber view of the RV, with all other parameters being equal, the FOV was increased from $340 \text{ mm} \times 276 \text{ mm}$ to $400 \text{ mm} \times 325 \text{ mm}$, in order to avoid aliasing artifacts. Human imaging was performed in accordance with protocols approved by the Human Investigation Committee at our institution; all subjects provided written informed consent.

Electrocardiogram (ECG) gating was used to acquire the images at mid diastole (trigger delay = 500 ms), in order to minimize the effects of cardiac motion. For each cardiac view per saturation pulse, PD (heart beat 1) and SR (heart beat 8) images were acquired with a breath-hold duration of approximately 8 s, with a wait time of 6 heart beats (~ 6 s) to allow for full recovery of magnetization [$> 5 \times T_1$; T_1 of myocardium = 1150 ms at 3T (3)] between the two image acquisitions. This relatively short breath-hold duration ensured reliable registration of PD and SR images. For each subject, 24 separate breath-hold acquisitions were performed (i.e., 6 cardiac views $\times$ 4 saturation pulses) with the magnet isocenter positioned at the mid-ventricular, short-axis plane. To optimally utilize the dynamic range of the 12-bit analog-to-digital converter (ADC), for each subject, the receiver gains were adjusted so that the maximum intensity of the PD image is approximately 80–90% of the maximum value of the ADC (i.e., 4095). This ensured that the signal of the SR image will be well above the noise. For each subject, the RF calibration, B_0 shimming, and receiver gains were kept unchanged between different SR image acquisitions, in order to permit M_{ZR} comparisons.

SAR Measurements—In the same set of nine subjects, a separate experiment was conducted to measure the predicted SAR for each saturation pulse, using a multi-slice first-pass cardiac perfusion MRI protocol without contrast agent. The saturation pulses were incorporated into a TurboFLASH pulse sequence with the following relevant imaging parameters: FOV = 350×285 mm, acquisition matrix = 144×118 (59 acquired phase-encoding lines), spatial

resolution = 2.4 mm × 2.4 mm, slice thickness = 8 mm, TE/TR = 1.2/2.34ms, TD = 10 ms, time-adaptive sensitivity encoding (TSENSE)(20) with an acceleration factor = 2, image acquisition time = 138 ms, total scan time (including TD, saturation pulse duration) ~ 160 ms, flip angle = 10°, repetitions = 40, and receiver bandwidth = 990 Hz/pixel. This experiment was performed with a simulated heart rate of 60 beats per minute, in order to measure the predicted SAR and maximum allowable perfusion image acquisitions per cardiac cycle within clinically acceptable SAR limits, for all nine subjects using an identical total acquisition window of 1000 ms. In addition, SAR per slice was calculated by dividing the SAR measurement by the maximum permitted number of slices.

Preliminary Evaluation—A 27 years old female subject was imaged without contrast agent using the aforementioned first-pass cardiac perfusion MRI protocol (except repetitions = 9), in order to evaluate the efficacy of these saturation pulses in vivo. For each saturation pulse, a short-axis plane of the LV was imaged repeatedly for 9 heart beats. A PD image was acquired during the first heartbeat with 4° flip angle, no saturation pulse, and centric k-space trajectory to minimize T₁-weighting. The RF calibration, shimming, and receiver settings were kept unchanged between acquisitions, to permit signal comparisons.

A 35 years old male subject was imaged during first-passage of contrast agent using the aforementioned cardiac perfusion protocol with the hybrid pulse train as the saturation pulse, in order to evaluate the uniformity of the myocardial T₁-weighted signal in three short-axis planes of the LV (apex, mid, base). A single dose (0.1 mmol/kg) of Gd-DTPA (Magnevist, Bayer Healthcare) was injected at a flow rate of 6 ml/s by a power injector (MEDRAD, Warrendale, PA), followed by a 20 ml flush of saline. For signal normalization, pre-contrast PD image was acquired during the first heartbeat with 4° flip angle, no saturation pulse, and centric k-space trajectory to minimize T₁-weighting.

Influence of the Saturation Pulse on the Accuracy of T₁-Weighted Imaging

A numerical simulation was performed to assess the influence of each saturation pulse on the accuracy of dynamic T₁-weighted imaging with the aforementioned TurboFLASH protocol. Specifically, for each pulse, the simulation was performed with three LV M_{ZR} measurements (Table 2): mean – standard deviation (SD), mean, and mean M_{ZR} + SD. For the M_Z calculation, the initial M_Z(0) after the saturation pulse and immediately prior to the TurboFLASH readout was defined as $M_z(0) = M_0 + (M_{ZR} - M_0)e^{-TD/T_1}$, where M₀ scalar = 1. This initial M_Z was then inputted to solve an iterative multiplication of rotation matrices in the rotating frame as a function of phase-encoding line. The simulation was performed with 10° excitation of a single isochromat at on-resonance and assumed complete dephasing of the transverse magnetization before the next RF excitation.

The T₁-weighted signal (S) was defined as M_Z at the center of k-space (21)(i.e., phase-encoding line = 30; effective recovery time = 80.2 ms), and this calculation was repeated for T₁ ranging from 50 – 1150 ms (100 ms steps). The true T₁-weighted signal (S₀) was calculated with M_{ZR} = 0, and the corresponding normalized signal was calculated as S/S₀ and plotted as a function of T₁.

Image Analysis

For signal normalization, the SR image (nominally zero signal) was divided by the corresponding PD (nominally maximum signal) image on a pixel-by-pixel basis, in order to correct for signal variations due to the receiver coil sensitivity (i.e., B₁⁺), T₂^{*} decay, B₁⁺ variation of the excitation pulse, imperfect slice profile, and underlying M₀, as previously described (8,9). The quotient image was multiplied by the factor sin(4°)/sin(10°) to account

for the difference in the two excitation angles. The resulting normalized image intensity conveniently has a range of 0 (complete saturation) to 1 (no saturation).

For phantom image analysis, a region of interest (ROI) was manually drawn to cover the whole phantom, and the mean M_{ZR} was calculated for each ROI. For cardiac image analysis, an ROI was manually drawn for the whole LV and/or RV per cardiac view, and the mean M_{ZR} was calculated for each ROI. For statistical analysis, each ROI was considered to be an independent sample, and the reported values represent the mean $\pm$ SD of averaged M_{ZR} over ROIs. For in vivo SAR analysis, the reported values represent the mean $\pm$ SD of SAR measurement. A single-factor analysis of variance was used to compare the mean values between the four pulse groups ($p < 0.05$ was considered to be significant), and the Tukey's honestly significant difference test was used to compare the mean values between each pair of two groups ($p < 0.05$ was considered to be significant). The statistical analysis was performed using the SPSS 13.0 software (SPSS[®] Inc, Chicago, IL).

For evaluation of the homogeneity of T_1 -weighted signal, an ROI was manually drawn for the whole myocardium. Care was taken to avoid partial volume effects. As described above, the T_1 -weighted images were divided by the PD image on a pixel-by-pixel basis. The three short-axis planes were subdivided into 16 segments according to the American Heart Association standardized model (22). For each segment, the normalized T_1 -weighted signal was plotted as a function of cardiac cycle.

Results

Figure 2 shows the magnitude of numerical M_{ZR} map as a function B_0 and B_1^+ scales for each of four saturation pulses. Compared with the standard and tailored pulse trains, both the hybrid pulse train and the BIR-4 pulse train yielded considerably lower M_{ZR} maps. The resulting RMS M_{ZR} measurements were 0.136, 0.055, 0.013, and 0.012 for the standard pulse train, tailored pulse train, hybrid pulse train, and BIR-4 pulse train, respectively. Consistent with the simulated M_{ZR} maps, the corresponding phantom M_{ZR} maps were in good agreement, and both the hybrid pulse train and BIR-4 pulse train produced considerably lower M_{ZR} maps than the standard and tailored pulse trains. The resulting RMS M_{ZR} measurements were 0.164, 0.053, 0.035, and 0.016 for the standard pulse train, tailored pulse train, hybrid pulse train, and BIR-4 pulse train, respectively. Figure 3 shows the corresponding profiles of M_{ZR} as a function of B_1^+ scale, for -65 Hz, 0 Hz, and 65 Hz off-resonance. Note that the standard pulse train produced lower M_{ZR} than the tailored pulse train for B_1^+ scale > 0.72 and vice versa for B_1^+ scale < 0.72 . In contrast, both the hybrid pulse train and BIR-4 pulse train yielded considerably lower M_{ZR} throughout the B_1^+ scale.

Figure 4 shows representative normalized SR images of one volunteer in all six cardiac views. The bright areas in the normalized images represent the (undesirable) non-zero M_{ZR} left behind from the effective flip angle variation due to B_0 and B_1^+ inhomogeneities. Consistent with the previous studies (9,13), the standard pulse train performed well within the LV (dark regions) but poorly within the RV (bright regions). The tailored pulse train performed similarly within the whole heart, but at the expense of producing medium level of non-uniform M_{ZR} values (gray regions and dark bands) within both ventricles. In contrast, both the hybrid RF pulse train and BIR-4 pulse train yielded uniformly minimal M_{ZR} within both ventricles (dark regions). Statistically, the mean M_{ZR} within the LV (0.018 ± 0.008 vs. 0.048 ± 0.007 vs. 0.015 ± 0.005 vs. 0.016 ± 0.004 ; standard pulse train, tailored pulse train, hybrid pulse train, and BIR-4 pulse train, respectively; $n = 45$; $p < 0.001$) and RV (0.069 ± 0.043 vs. 0.057 ± 0.012 vs. 0.018 ± 0.006 vs. 0.015 ± 0.005 ; standard pulse train, tailored pulse train, hybrid pulse train, BIR-4 pulse train, respectively; $n = 45$; $p < 0.001$) was different between all pulse groups (see Table 2). Within each pair of groups, the mean M_{ZR} within the LV was significantly higher for the

tailored pulse train than for the other three pulses ($p < 0.001$), and the mean M_{ZR} was not different for any pair between the standard pulse train, hybrid pulse train, and BIR-4 pulse train. The corresponding range (min-max) of M_{ZR} was 0.009–0.054, 0.035–0.069, 0.008–0.028, and 0.010–0.027 for the standard pulse train, tailored pulse train, hybrid pulse train, and BIR-4 pulse train, respectively. Within the RV, the mean M_{ZR} was significantly higher for both the standard and tailored pulse trains than for either the hybrid pulse train or BIR-4 pulse train ($p < 0.001$), and not different between the standard and tailored pulse trains and between the hybrid pulse train and BIR-4 pulse train. The corresponding range (min-max) of M_{ZR} was 0.013–0.195, 0.026–0.079, 0.009–0.042, and 0.007–0.032 for the standard pulse train, tailored pulse train, hybrid pulse train, and BIR-4 pulse train, respectively. These in vivo cardiac M_{ZR} results demonstrate the superior efficacy of the hybrid pulse train and BIR-4 pulse train over the standard and tailored pulse trains.

In nine subjects studied, the standard pulse train, tailored pulse train, and hybrid pulse train were capable of acquiring six slices using the TurboFLASH protocol (i.e., total acquisition time = 960 ms; total acquisition window = 1000 ms), whereas the BIR-4 pulse train was capable of acquiring only 1.4 ± 0.5 slices within clinically acceptable SAR limits (see Table 3). The mean SAR per slice was 7.5 ± 0.5 , 10.7 ± 0.7 , 14.2 ± 1.3 , and 50.5 ± 4.0 for the standard pulse train, tailored pulse train, hybrid pulse train, and BIR-4 pulse train, respectively. Statistically, the mean SAR per slice was significantly different between all pulse groups ($p < 0.001$) and between each pair of groups ($p < 0.02$). These in vivo results strongly suggest that the hybrid pulse train can be safely used for multi-slice (≥ 6), first-pass cardiac perfusion MRI at 3T.

Figure 5 shows repeated acquisitions of non-contrast T_1 -weighted images over multiple heart beats. Both the standard and tailored pulse trains yielded residual signal compared with the hybrid and BIR-4 pulse trains. As can be seen in Figure 5, as well as in Figure 4, both the hybrid and BIR-4 pulse trains are also capable of performing effective saturation of magnetization beyond the heart (e.g., the liver) because of their insensitivity to B_0 and B_1 inhomogeneities. Figure 6 shows representative first-pass cardiac perfusion images at pre-contrast, peak blood enhancement, and peak myocardial wall enhancement, using the hybrid pulse train as the saturation pulse. Figure 6 also shows the corresponding plots of normalized T_1 -weighted signal for each of 16 segments as a function of cardiac cycle. These normalized signal-time curves show good agreement between segments, suggesting that the hybrid pulse train had produced uniform saturation of magnetization.

Figure 7 shows plots of S/S_0 as a function of T_1 . The standard pulse train yielded relatively accurate S/S_0 , whereas the tailored pulse train yielded less accurate S/S_0 . Compared with the standard and tailored pulse trains, both the hybrid pulse train and BIR-4 pulse train produced more accurate S/S_0 . It should be noted that for $M_{ZR} = 0.02$, the normalized $S \geq 1.2$ (i.e., 20% error) for $T_1 \geq 1050$ ms (data not shown). These numerical results highlight the importance of uniform T_1 -weighting in first-pass cardiac perfusion MRI, and that the examined first-pass perfusion protocol (acquisition time ~ 160 ms; TD = 10 ms, flip angle = 10°) is sensitive to M_{ZR} left behind by the nominal saturation pulse for large T_1 values.

Discussion

This study has compared the efficacy of four saturation pulses: standard pulse train, tailored pulse train, hybrid pulse train, and BIR-4 pulse train. Compared with the standard and tailored pulse trains, both the hybrid pulse train and BIR-4 pulse train yielded more uniform saturation of magnetization ($M_{ZR} \leq 0.02$) within the whole heart (Table 2). These findings are significant because M_{ZR} as small as 5–7% of M_0 can cause significant error in the dynamic T_1 -weighted signal (Fig. 7). The influence of the saturation pulse on the accuracy the T_1 -weighted signal was determined to be relatively high, particularly for large T_1 values (≥ 1000 ms). These

findings confirm that it is critically important to perform uniform T_1 -weighting for accurate quantification of T_1 from first-pass cardiac perfusion MR images.

The major advantage of the hybrid pulse train is that it is relatively insensitive to both B_0 and B_1^+ variations within the whole heart at 3T, and, therefore, can be used in conjunction with commercially available automated B_0 shimming and RF calibration procedures. Its efficacy is comparable to that of the BIR-4 pulse train, whereas its RF energy is 76% less than that of the BIR-4 pulse train. Our results show that the extra RF energy (134% more than that of the standard pulse train) needed to employ the hybrid pulse train does not exceed the clinically acceptable SAR limits for multi-slice (≥ 6) first-pass cardiac perfusion MRI (Table 3).

In our study, the standard pulse train produced lower M_{ZR} in the LV than the tailored pulse train, whereas vice versa was true in the study by Sung et al. (13). The conflicting comparative results between the standard and tailored pulse trains by our study and by Sung et al. (13) deserves an explanation. As shown in Figure 2 and Figure 3 and Figure 4 in (13), both the standard and tailored pulse trains are relatively insensitive to clinically relevant B_0 variation, and the standard pulse train produces lower M_{ZR} than the tailored pulse train for B_1^+ scale > 0.72 and vice versa B_1^+ scale < 0.72 . It should be noted that the commercially available automated RF calibration procedures and B_1 homogeneity of the transmit body coil are likely to differ between different vendors (e.g., Siemens vs. GE). Therefore, the conflicting comparative results between the standard and tailored pulse trains by our study and by Sung et al. (13) are highly likely due to the differences in RF calibration and/or B_1 homogeneity of the transmit body coil. Our in vivo results (Figure 4; Table 2) suggest that the mean B_1^+ scale in the heart examined by our study was greater than 0.72. In contrast, in the study by Sung et al. (13), the tailored pulse train yielded lower M_{ZR} than the standard pulse train (see Figure 4 and Table 2 in reference 13), suggesting that the mean B_1^+ scale in the heart was less than 0.72. Furthermore, it should be noted that Sung et al. (13) measured relative M_{ZR} , because they acquired both SR and PD images with two “shots” segmented over two heart beats, and with only one heartbeat for magnetization recovery between the SR and PD acquisitions. Therefore, this key difference needs to be considered when comparing their relative M_{ZR} measurements with our absolute M_{ZR} measurements.

This study had three limitations that warrant discussion. First, the tailored pulse train used in this study was optimized for a GE whole-body 3T system and may not represent the optimal tailored pulse train for our Siemens whole-body 3T system. To fully optimize the tailored pulse train for our Siemens 3T system, it is necessary to perform prior measurements of B_0 and B_1 variations within the heart. The conflicting results of the tailored pulse train by our study and by Sung et al. (13) suggest that a tailored pulse train needs to be calibrated for each scanner, using rapid cardiac B_0 and B_1 mapping pulse sequences. An optimized tailored pulse train, however, is likely to yield lower efficacy than a hybrid pulse train, because the latter utilizes an AHP which is more effective than a non-selective rectangular pulse. Second, the proposed hybrid pulse train design was based on B_0 variation of 260 Hz, as reported by Schar et al. (5) using a Philips whole-body 3T system, and B_1^+ scale of 0.5 – 1.1, as reported by Sung et al. (6) using a GE whole-body 3T system. Despite the lack of direct cardiac B_1^+ scale and B_0 offset measurements from our Siemens 3T system, the proposed hybrid pulse train performed remarkably well within the whole heart in all nine subjects, because the hybrid pulse train was designed to be highly insensitive to B_0 and B_1 inhomogeneities (Figure 2 and Figure 3). As such, our results suggest that the hybrid pulse train may perform effectively without calibration for different whole-body 3T systems. Third, the proposed hybrid pulse train was designed without SAR optimization. While it may be possible to perform joint optimization of effective saturation of magnetization and SAR, the optimization will require a complex theoretical SAR model that is beyond the scope of this study. We have shown the feasibility to acquire up to six slices using a typical first-pass cardiac perfusion MRI protocol with the hybrid pulse train

as the saturation pulse. As such, the hybrid pulse train can be used safely for typical multi-slice, first-pass cardiac perfusion MRI.

The hybrid pulse train can also be used for other applications. In particular, the hybrid pulse can train be used for T_1 mapping in the presence of B_0 and B_1 inhomogeneities. In particular, it can be used in an Look-Locker approach to measure myocardial T_1 (23), where the benefit of a saturation pulse over an inversion pulse would be that the former is insensitive to heart rate variability (24). The hybrid pulse train can also be used in a SR double angle method for rapid B_1 mapping (16).

Acknowledgments

The authors would like to thank Professor Michael Garwood of the University of Minnesota for providing us with the AHP and BIR-4 pulse-generating software.

Grant Sponsor: American Heart Association: 0730143N

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Figure 1.

Pulse sequence diagrams: (a) standard pulse train, (b) tailored pulse train, (c) hybrid pulse train, and (d) BIR-4 pulse train. The spoiler gradients are applied before and after each saturation pulse to dephase the transverse magnetization. The crusher gradients in between RF pulses are cycled to eliminate stimulated echoes. These diagrams are drawn to approximate proportions but not to exact scale. G_{ss} : slice-select gradient; G_{PE} : phase-encoding gradient; G_{FE} : frequency-encoding gradient. The RF pulse characteristics are summarized in Table 1.

Figure 2.

(Top row) Simulated M_{ZR} maps as a function of B_0 and B_1^+ scales. Compared with the (a) standard pulse train and (b) tailored pulse train, both the (c) hybrid pulse train and the (d) BIR-4 pulse train yielded considerably lower M_{ZR} maps. The resulting RMS M_{ZR} measurements were 0.136, 0.055, 0.013, and 0.012 for the standard train, tailored train, hybrid train, and BIR-4 pulse train, respectively. (Bottom row) M_{ZR} maps of the phantom as a function of B_0 and B_1^+ scales. Consistent with the simulation results, the corresponding phantom M_{ZR} maps were in good agreement, and both the (g) hybrid pulse train and (h) BIR-4 pulse train produced considerably lower M_{ZR} maps than the (d) standard and (e) tailored pulse trains. The resulting RMS M_{ZR} measurements were 0.164, 0.053, 0.035, and 0.016 for the standard train, tailored train, hybrid train, and BIR-4 pulse train, respectively. The M_{ZR} maps are displayed with identically narrow grayscale (0 – 0.4 in dimensionless units) to bring out regional variations.

Figure 3.

(Top row) Numerically simulated profiles of M_{ZR} as a function of B_1^+ scale: (a) -65 Hz, (b) 0 Hz, and (c) 65 Hz off-resonance. (Bottom row) Corresponding profiles of M_{ZR} in the phantom as a function of B_1^+ scale: (d) -65 Hz, (e) 0 Hz, and (f) 65 Hz off-resonance. Note that the standard pulse train yielded lower M_{ZR} than the tailored pulse train for B_1^+ scales > 0.72 , and vice versa for B_1^+ scales < 0.72 . In contrast, the hybrid pulse train and BIR-4 pulse train yielded considerably lower M_{ZR} throughout the B_1^+ scale.

Figure 4.

Representative set of normalized SR images from one volunteer that compares the performance of four saturation pulses. The standard pulse train performed well within the LV (dark regions) but poorly within the RV (bright regions). The tailored pulse train performed similarly within the whole heart, but at the expense of producing medium level of M_{ZR} values (gray regions and dark bands) within both ventricles. In contrast, both the hybrid RF pulse train and BIR-4 pulse train yielded uniformly minimal M_{ZR} within both ventricles (dark regions). The normalized images are displayed with identically narrow grayscale (0 – 0.2 in dimensionless units) to bring out regional variations.

Figure 5.

Representative repeated acquisitions of non-contrast T_1 -weighted images over multiple heart beats: (first row) standard train, (second row) tailored train, (third row) hybrid train, and (fourth row) BIR-4 train. Both the standard and tailored pulse trains yielded residual signal compared with the hybrid and BIR-4 pulse trains. The T_1 -weighted images are displayed with identically grayscale (200 – 1000 in arbitrary units) to bring out regional variations.

Figure 6.

Representative first-pass cardiac perfusion images at (first column) pre-contrast, (second column) peak blood enhancement, and (third column) peak myocardial wall enhancement, using the hybrid pulse train as the saturation pulse: (first row) apex, (second row) mid-ventricular, and (third row) base. (Fourth column) The corresponding plots of normalized T_1 -weighted signal for each of 16 segments as a function of cardiac cycle. These normalized signal-time curves show good agreement between segments, suggesting that the hybrid pulse train had produced uniform T_1 -weighted signal. The T_1 -weighted images are displayed with identically grayscale (200 – 1500 in arbitrary units).

Figure 7.

Simulated plots of S/S_0 as a function of T_1 for the LV: (a) standard pulse train, (b) tailored pulse train, (c) hybrid pulse train, and (d) BIR-4 pulse train. The standard pulse train yielded relatively accurate S/S_0 , whereas the tailored pulse train yielded less accurate S/S_0 . Compared with the standard and tailored pulse trains, both the hybrid pulse train and BIR-4 pulse train produced more accurate S/S_0 . Solid line represents S/S_0 calculated with mean in vivo M_{ZR} . Upper and lower dotted lines represent S/S_0 calculated with mean+SD and mean-SD in vivo M_{ZR} , respectively.

Summary of RF pulse characteristics of the standard pulse train, tailored pulse train, hybrid pulse train, and BIR-4 pulse train. Each pulse is preceded by 1 ms long spoiler gradient and followed by 3 ms long spoiler gradient. For each pulse train, the first and second crusher gradient durations are 3 and 2 ms, respectively. For the hybrid train, the last subpulse is an AHP pulse. For the BIR-4 pulse train, only the first set of crusher gradients is employed. Note that the B_1^+ represents the peak nominal value. The transmit RF energy was calculated assuming that the RF voltage needed for B_1^+ of 500 HZ is 600 V.

Table 1

	Angle (°)	Duration (ms)	B_1^+ (Hz)	RF Energy (J)	Relative RF Energy	Duration (ms)
Standard Train	90	0.5	500	10.8	1	10.5
	90	0.5	500			
	90	0.5	500			
Tailored Train	96	0.6	444.44	17.8	1.65	11.7
	228	1.3	487.18			
	141	0.8	489.58			
Hybrid Train	140	0.8	486.11	25.3	2.34	11.8
	90	0.5	500			
	90 (AHP)	1.54	666.67			
BIR-4 pulse train	90	8.1	500	105.8	9.80	23.2
	90	8.1	500			

Mean, maximum, and minimum M_{ZR} left behind after nominal saturation. Each ROI was considered to be an independent sample ($n = 45$). Statistically, the mean M_{ZR} within the LV and RV was different between all pulse groups ($p < 0.001$). Within each pair of groups, the mean M_{ZR} within the LV was significantly higher for the tailored pulse train than for the other three pulses ($p < 0.001$), and the mean M_{ZR} was not different for any pair between the standard pulse train, hybrid pulse train, and BIR-4 pulse train. Within the RV, the mean M_{ZR} was significantly higher for both the standard and tailored pulse trains than for either the hybrid pulse train or BIR-4 pulse train ($p < 0.001$), and not different between the standard and tailored pulse trains and between the hybrid pulse train and BIR-4 pulse train. The reported mean values represent the mean $\pm$ SD of averaged M_{ZR} over ROIs.

Table 2

Ventricle		Standard Train	Tailored Train	Hybrid Train	BIR-4 pulse train
LV	Mean	0.018 $\pm$ 0.008	0.048 $\pm$ 0.007	0.015 $\pm$ 0.005	0.016 $\pm$ 0.004
	Max	0.054	0.069	0.028	0.027
	Min	0.009	0.035	0.008	0.010
RV	Mean	0.069 $\pm$ 0.043	0.057 $\pm$ 0.012	0.018 $\pm$ 0.006	0.015 $\pm$ 0.005
	Max	0.195	0.079	0.042	0.032
	Min	0.013	0.026	0.009	0.007

Table 3

The mean predicted SAR and maximum allowable perfusion image acquisitions per cardiac cycle. For each saturation pulse, the SAR per slice was calculated by dividing the SAR measurements by the maximum permitted number of slices. The acquisition time per slice was 160 ms, and these measurements were recorded with a total acquisition window of 1000 ms (i.e., heart rate of 60 beats per minute). The reported values represent the mean $\pm$ SD.

	Standard Train	Tailored Train	Hybrid Train	BIR-4 pulse train
Total SAR (%)	45.0 $\pm$ 3.2	64.4 $\pm$ 4.2	85.4 $\pm$ 8.0	72.0 $\pm$ 23.8
Slices per cardiac cycle	6 $\pm$ 0	6 $\pm$ 0	6 $\pm$ 0	1.4 $\pm$ 0.5
Mean SAR per slice (%)	7.5 $\pm$ 0.5	10.7 $\pm$ 0.7	14.2 $\pm$ 1.3	50.5 $\pm$ 4.0