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Rapid B_1^+ Mapping Using a Pre-Conditioning RF Pulse with TurboFLASH readout

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Abstract

In magnetic resonance imaging (MRI), radiofrequency (RF) field (B_1^+) inhomogeneity can lead to signal intensity variations and quantitative measurement errors. By independently mapping the local B_1^+ variation, the RF-related signal variations can be corrected for. In this study, we present a new fast B_1^+ mapping method using a slice-selective pre-conditioning (SS-Pre) RF pulse. Immediately after applying a SS-Pre pulse, a turbo fast low angle shot (TurboFLASH) imaging sequence with centric k-space reordering is performed to capture the residual longitudinal magnetization by the SS-Pre pulse due to B_1^+ variation. Compared to the reference double-angle method (DAM), this method is considerably faster. Specifically, the total scan time for the DAM is equal to the product of 2 (number of images), the number of phase-encoding lines and approximately $5T_1$, whereas the SS-Pre method takes approximately $5T_1$. This method was validated in vitro and in vivo at a 3T whole-body MRI system. The combined brain and pelvis B_1^+ measurements showed excellent agreement and strong correlation with those by the DAM (mean difference=0.025; upper and lower 95% limits of agreement were -0.07 and 0.12; $R=0.93$, $p<0.001$). This fast B_1^+ mapping method can be used for a variety of applications, including body imaging where fast imaging is desirable.

Keywords

Fast B_1^+ mapping; slice-selective pre-conditioning RF pulse; B_0 insensitivity

INTRODUCTION

In magnetic resonance imaging (MRI), the radiofrequency (RF) field (B_1^+) inhomogeneity can produce flip angle variations, which can ultimately lead to signal intensity variations and quantitative measurement errors. Typically, B_1^+ variation within the human body at high magnetic field strengths ($\geq 3T$) can vary on the order of 20-63% (1-3). The B_1^+ variation is in part due to the spatial variation of the field strength produced by the RF coil and in part due to the dielectric properties of the human body. Local variation of B_1^+ can lead to spatially varying image signals and contrasts (4-7), and produce errors for quantitative imaging (5,8), signal intensity correction (9-10), or parallel transmission (11). Thus, these applications may require B_1^+ mapping to correct for the errors.

Several B_1^+ mapping methods have been developed for in vivo applications. These methods are mainly based on measuring the signal ratio from two flip angles (12-16), stimulated echoes (17-18), or two identical RF pulses in the steady state (19). Alternative approaches

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have also been proposed, such as using the 180° signal null (20), monitoring the signal phase (21-23), or applying preparation pulses (24). Among these methods, the most widely used B_1^+ mapping method is the double angle method (DAM) (12), which uses the ratio of two images acquired at two different nominal flip angles α and 2α . While DAM is straightforward to implement, it is inherently inefficient, due to a need for long repetition time (TR) $\geq 5T_1$, in order to allow for a full recovery of the magnetization before each excitation. To improve time efficiency of the DAM, several modifications have been proposed for reducing TR by adding compensating pulses (13), resetting the longitudinal magnetization using a saturation pulse (3,15), or using catalyzed pulses (16). However, these accelerated DAM methods are still inefficient for breath-held body imaging, using conventional single-echo imaging pulse sequence.

In this study, we describe a novel and efficient method for rapid B_1^+ mapping. This method is similar to the previous method proposed by Klose (24) as using a pre-conditioning RF pulse, but there are two different features. First, a slice-selective sinc pulse is used as a pre-conditioning RF pulse because of its relative insensitivity to off-resonance instead of using a nonselective pre-conditioning RF pulse. Second, in the previous method, a series of twenty images was acquired to find the transmitter amplitude corresponding to the local signal maximum resulting from a 180° flip angle. In this study, only two data acquisitions are performed, where the two images differ from each other primarily in their dependence on the flip angle: a slice-selective pre-conditioning RF pulse image (here called a SS-Pre image) and a proton density (PD) image for signal normalization. A turbo fast low angle shot (TurboFLASH) readout pulse sequence with centric k-space reordering was incorporated, immediately after a SS-Pre pulse excitation with associated spoiler gradients. To further accelerate imaging, an effective saturation-recovery (SR) module can be optionally added to generate a consistent magnetization state (Fig. 1) with $TR < 5T_1$. Phantom and in vivo volunteer studies were performed to evaluate the accuracy of this method against the DAM at $3T$.

THEORY

The mathematical equation describing the effects of a SS-Pre pulse with the nominal flip angle, α^{nom} , is derived using the Bloch equation (25) in the off-resonance rotating frame. If we assume that the partial saturation effects in the PD and SS-Pre images can be neglected after the normalization and that the off-resonance effects can be neglected, then the equation simplifies to

$$\frac{\text{SSPre}(r)}{\text{PD}(r)} = \frac{M_z(r)}{M_z^0(r)} = \cos(\kappa(r) \cdot \alpha^{\text{nom}}), \quad (1)$$

where $M_z(r)$ is the longitudinal magnetization, $M_z^0(r)$ is the fully recovered longitudinal magnetization, $\kappa(r)$ is a B_1^+ scale factor at position r , which by definition is actual B_1^+ divided by nominal B_1^+ . Eq. 1 can be rewritten as

$$\kappa(r) = \cos^{-1} \left(\frac{\text{SSPre}(r)}{\text{PD}(r)} \right) / \alpha^{\text{nom}}. \quad (2)$$

METHODS

Pulse Sequence

Figure 1 shows the pulse sequence diagram, which consists of three modules. First, the SS-Pre pulse module consists of a slice-selective pre-conditioning RF pulse excitation, with associated spoiler gradients for dephasing any residual transversal magnetization. Second, the imaging module consists of a TurboFLASH readout pulse sequence with centric k-space reordering, to image the residual longitudinal magnetization. Third, the SR module consists of an effective SR pulse with a SR time delay (TD_{SR}). It can be optionally used for generating a consistent magnetization state before a SS-Pre pulse imaging sequence. In this study, a hybrid adiabatic-rectangular pulse train (26) was used to achieve effective saturation, followed by a TD_{SR} . A $TD_{SR} = 950$ ms was used for all experiments, except for the T_1 -doped water phantom, where a $TD_{SR} = 300$ ms was used. The relevant imaging parameters of the SS-Pre pulse (27) used for phantoms and volunteers were: $\alpha^{nom} = 60^\circ$, $\tau = 2.8$ ms, time-bandwidth product = 6, transmitter bandwidth = 2.1 kHz, slice thickness = 6 times that of the imaging slice thickness, and 2.6-ms-long spoiler gradients with the magnitude of net zeroth gradient moment of 135m/Tm ms. The relevant imaging parameters were: acquisition matrix = 64×48 , TR = 2.56 ms, echo time (TE) = 1.29 ms, flip angle = 10° , receiver bandwidth = 1500 Hz/pixel and slice thickness = 8 mm. A PD image was acquired with the same imaging parameters without a SS-Pre pulse. Without using the SR module, a long relaxation-recovery time ($\geq 5T_1$) is needed to allow for a full recovery of magnetization before the next excitation. For the purpose of validation, the DAM was performed. Relevant imaging parameters for DAM included: excitation flip angle = $60^\circ/120^\circ$, time-bandwidth product = 6.0, RF pulse duration = 3.84 ms and bandwidth = 1500 Hz/pixel. The pulse sequences were implemented on a whole-body 3T MR scanner (Tim Trio; Siemens Medical Solutions, Erlangen, Germany) and tested on two phantoms, four healthy volunteers for brain imaging, and seven healthy volunteers for pelvic imaging. All experiments were performed using the body coil for RF transmission. For signal reception, a standard phased-array coil was used for imaging phantoms and in vivo volunteers. The manufacturer's standard automated B_0 shimming was performed prior to the image acquisitions. Protocols of the human studies were approved by the Human Investigation Committee at our institution and informed consents were obtained from all volunteers.

Dynamic range and nominal flip angle

The dynamic range of κ depending on the choice of α^{nom} was tested on a phantom (oil phantom, 20-cm diameter sphere, $T_1 = 530$ ms). The α^{nom} of 40° , 60° and 80° was used to image the phantom in an axial plane at magnet isocenter. The amplitude of the nominal RF transmitter was manually set to acquire the nominal κ ranging from 0.2-1.8 (0.2 steps). The region-of-interest (ROI) was defined as where measured κ falls within 0.98-1.02 at nominal $\kappa = 1.0$. The mean and coefficient of variation (CV) of κ were calculated within the ROI. The CV was calculated as the ratio of the standard deviation and mean of κ .

Numerical simulation of the partial saturation effects

One of the main assumptions underlying the above analysis (Eq. 1) is that the saturation effects after signal normalization are negligible. However, although the image intensity is weighted by the center line of k-space, the evolution of longitudinal magnetization during the acquisition of the whole k-space is not identical in both the SS-Pre and PD images, due to the difference in the non-steady states of magnetization (28). In order to estimate the magnitude of this partial saturation effects, a brain PD image was used as a template (Fig. 3a). The saturation effect of the centric k-space trajectory was simulated in the k-space using the Bloch equations (25) with the following parameters: matrix size = 64×48 , $T_1 = 10$ -3000 ms, TR = 2.56 ms, flip angle = 10° and $\alpha^{nom} = 60^\circ$. The brain template was multiplied by

the simulated signal weighting in the k-space and then the SS-Pre and PD images were finally calculated as the inverse Fourier transform of the resulting k-space images. The κ map was obtained from the simulated SS-Pre and PD images and the root-mean-squared error (RMSE) of κ (Fig. 3c) was calculated pixel-by-pixel within the brain region (shown in Fig. 3a) against those without the saturation effect.

Off-resonance Effects

A second assumption underlying the above analysis is that the SS-Pre pulse is insensitive to B_0 . To evaluate the sensitivity to off-resonance, phantom imaging (oil phantom) was performed on an axial plane at magnet isocenter with resonance offsets ranging from 0-500 Hz (100 Hz steps). The RMSE of the κ map was calculated on a pixel-by-pixel basis within the whole phantom, with the DAM at on-resonance as reference.

Dynamic range and noise analyses

The dynamic range of κ depending on the noise level was tested with the oil phantom in an axial plane at magnet isocenter. In order to see the effect of changing κ on the SS-Pre pulse and TurboFLASH excitation pulse, the nominal κ was manually set from 0.2 to 1.8 (0.2 steps) in three different ways: (1) only the amplitude of the SS-Pre pulse was changed, (2) only the amplitude of the TurboFLASH excitation pulse was changed, and (3) the amplitude of the RF transmitter was changed (i.e., all RF pulses were affected). The CV was calculated within an ROI at $\kappa = 1.0$, where measured κ falls within 0.98-1.02 at nominal $\kappa = 1.0$.

To evaluate the effects of random noise on the method, Gaussian noise was added to the phantom SS-Pre and PD images, prior to the κ calculation. To emulate clinically relevant SNR, we measured the SNR of representative brain and pelvis PD images acquired in a volunteer with nominal TurboFLASH flip angle of 10° . These preliminary experiments showed that the SNR in the brain and pelvis PD images was approximately 300 and 100, respectively. We then empirically determined the level of Gaussian noise needed to decrease the SNR (e.g., 300 and 100 for the brain and pelvis, respectively) of the phantom PD image acquired with nominal $\kappa = 1.0$, and applied this noise value to all three different sets of in vitro SS-Pre and PD images. The CV of κ was calculated from the simulated images. The resulting analysis produced two additional sets of dynamic range and noise results, with clinically relevant SNR values.

Phantom Imaging

Two phantoms (T_1 -doped water, 20-cm diameter sphere, $T_1 = 320$ ms; agar, 7-cm diameter bottle, $T_1 = 1680$ ms) were imaged in a coronal plane at magnet isocenter (T_1 -doped water, agar; FOV = 300×225 mm², 200×100 mm²; acquisition matrix = 64×48 , 64×32 ; in-plane resolution = 4.69×4.69 mm², 3.12×3.12 mm²; respectively). The total image acquisition times for acquiring the SS-Pre and PD images were approximately 2s (water) and 9s (agar), including the relaxation delay time ($\geq 5T_1$) between two image acquisitions, and were approximately 0.6s (water) and 2s (agar) with the SR module. The reference DAM measurements were also acquired, with total image acquisition times of approximately 153.6s (water) and 806.4s (agar). The RMSE of the κ map compared against the reference κ map was calculated on a pixel-by-pixel basis within the whole region of the phantom. Pearson correlation coefficient (R) was calculated to compare the κ measurements within the ROI, using MATLAB® R2008b software (The Mathworks, Inc., Natick, MA).

Brain Imaging

Four healthy volunteers (31.4 ± 2.7 years old) were imaged in a sagittal plane (isocentered at the eyebrow level) with of the brain with FOV = 360×270 mm² and in-plane resolution =

$5.62 \times 5.62 \text{ mm}^2$. The observed T_1 value within the brain is approximately 2s (29), except in the CSF (about 4s). Since the majority of quantitative applications are focused on the brain parenchyma, we assume a T_1 of approximately 2s for the brain in this study. The total image acquisition times for the SS-Pre and PD images, including the relaxation delay time, were approximately 10s and approximately 2.3s using the SR module. For the reference DAM measurement, total image acquisition time was approximately 960s. The brain was manually contoured and the RMSE and R of κ against the reference κ map was calculated pixel-by-pixel within this ROI.

Pelvic Imaging

Seven healthy volunteers (4 females, 31.5 ± 4.4 years old; 3 males, 32 ± 2.6 years old) were imaged in an axial plane of the pelvic area with $\text{FOV} = 380 \times 190 \text{ mm}^2$, acquisition matrix = 64×32 and in-plane resolution = $5.93 \times 5.93 \text{ mm}^2$. The axial imaging plane was located in the middle of the uterus for females and in the middle of the prostate for males. The total image acquisition times for the SS-Pre and PD images, including the relaxation delay time, were approximately 10s and approximately 2.2s with the SR module (assuming $T_1 \sim 2\text{s}$ for pelvic area at 3T (30)). For the reference DAM measurement, total image acquisition time was approximately 640s. Because of the long acquisition time for the DAM, an ROI was manually selected on both sides of pelvic region (shown as green rectangular regions in Fig. 8a) to exclude the central regions which might suffer from motion artifacts, such as respiratory or intestinal motions. The RMSE and R of κ against the reference κ map was calculated pixel-by-pixel within the ROI.

RESULTS

Dynamic range and nominal flip angle

Figure 2 shows the mean and corresponding CV of κ with nominal κ ranging from 0.2 to 1.8 for $\alpha^{\text{nom}} = 40^\circ, 60^\circ$ and 80° . In Fig. 2a, the mean of κ shows an excellent agreement with the nominal value until the signal is rectified at $\kappa \cdot \alpha^{\text{nom}} \approx 90^\circ$ ($\kappa \approx 2.25, 1.5$ and 1.12 for α^{nom} of $40^\circ, 60^\circ$ and 80° , respectively). Figure 2b shows that the lower bound of κ is limited at $\kappa = 0.39, 0.33$ and 0.26 with a CV of less than 5%, while the upper bound is limited near $\kappa \approx 2.25, 1.5$ and 1.12 for α^{nom} of $40^\circ, 60^\circ$ and 80° , respectively. Since the κ is likely to vary within $0.6 < \kappa < 1.5$ in the body at 3T (3,18-19), we used the α^{nom} of 60° to minimize noise while avoiding signal rectification in the SS-Pre image.

Numerical simulation of the partial saturation effects

Figure 3b shows the representative simulated magnetizations of the SS-Pre and PD with $T_1 = 500 \text{ ms}$ and 2000 ms along the phase-encoding lines. The centric k-space ordering scheme acts like a low-pass filter for both images. Figure 3c shows that the RMSE of κ is approximately less than 0.5% at $T_1 \geq 500\text{ms}$. Since the clinically observed T_1 range within the body is typically greater than 800ms at 3T (29), the saturation effect on the κ map is likely to be negligible for most applications at 3T, as shown in Fig. 3c.

Off-resonance Effects

Figure 4 shows that the κ map is relatively insensitive to off-resonance, with less than 1.6% RMSE in κ up to 500Hz off-resonance. Since the clinically observed B_0 offset within the body at 3T is on the order of $\pm 150 \text{ Hz}$ (32), this method can provide a considerably B_0 -insensitive result for most applications at 3T.

Dynamic range and noise analyses

Figure 5a shows the CV of κ within the ROI when only the SS-Pre pulse, only TurboFLASH pulse or all RF pulses were affected by the changes of nominal κ ranging from 0.2-1.8 (0.2 steps). The CV observed when only the SS-Pre pulse is affected is relatively significant at lower nominal κ compared to the case when only the TurboFLASH pulse is affected. This is because the signal in the SS-Pre image is more sensitive to the noise for lower SS-Pre pulse excitation, due to the cosine behavior of the signal as shown in Eq. 1. The SNR of the PD image at nominal $\kappa = 1$ was measured as approximately 980.

Figures 5b-5c show the CV and corresponding mean of κ within the ROI after adding Gaussian noise to the SS-Pre and PD images to decrease the SNR of the PD image (at nominal $\kappa = 1$) to approximately 300 and 100, as observed in the brain and pelvis, respectively. In Fig. 5b, the lower bound of κ is limited at $\kappa = 0.33$, 0.50 and 0.68 with a CV of less than 5% for SNR = 980, 300 and 100, respectively. In Fig. 5c, the mean of κ within the ROI shows an excellent agreement with the nominal value until the upper bound of κ is limited by the signal rectification near $\kappa \cdot \alpha^{\text{nom}} \approx 90^\circ$.

Phantom Imaging

Figure 6 shows the reference DAM κ maps of the water (Fig. 6a) and agar phantoms (Fig. 6e), and the corresponding κ maps (Fig. 6b, Fig. 6f) from the SS-Pre method. The differences of κ maps of water and agar phantoms against the reference are shown in Fig. 6c and Fig. 6g, respectively. The profiles of the water (Fig. 6d) and agar (Fig. 6h) phantoms were also plotted along the vertical lines shown in Fig. 6a and Fig. 6e. Since water has a high dielectric constant, the water phantom has a larger variation of κ than the agar phantom (31). The κ maps of the SS-Pre method and DAM showed excellent agreement and strong correlation within the phantoms (water, agar; mean difference = 0.0005, 0.0049; 95% limits of agreement were -0.015 and 0.016, -0.004 and 0.014; $R = 0.99$, 0.99; $p < 0.001$; respectively). The RMSE of the κ maps for the water and agar phantoms were 0.008 and 0.007 for the SS-Pre method, and slightly increased to 0.010 and 0.016 for the SS-Pre method with the SR module, respectively (Table 1).

Brain Imaging

Figures 7b-7c show a representative reference κ map in the brain and the corresponding κ map from the SS-Pre method. The difference κ map is also shown in Fig. 7d. The profiles were plotted along the vertical line (Fig. 7e) and the horizontal line (Fig. 7f) drawn in Fig. 7a. The brain κ maps measured by the SS-Pre method and DAM in the volunteers showed excellent agreement (mean difference = 0.02; 95% limits of agreement were -0.03 and 0.07) and strong correlation ($R = 0.98$; $p < 0.001$). The range of RMSE of the κ maps for all volunteers was 0.015-0.033 for the SS-Pre method and 0.02-0.036 for the SS-Pre method with the SR module (Table 2).

Pelvic Imaging

Figures 8b-8c show a representative reference κ map and the corresponding κ map from the SS-Pre method of a female pelvis. The difference κ map is also shown in Fig. 8d. The profiles were plotted along the vertical line (Fig. 8e) and the horizontal line (Fig. 8f) drawn in Fig. 8a. Although the κ maps contained some motion artifacts, particularly in the DAM images, they showed good agreement within the ROI (mean difference = 0.026; 95% limits of agreement were -0.082 and 0.135) and a good correlation ($R = 0.9$; $p < 0.001$). The RMSE of the κ maps for all volunteers was 0.041-0.075 for the SS-Pre method and 0.041-0.077 for the SS-Pre method with the SR module (Table 2).

DISCUSSION

This study demonstrates a new fast B_1^+ mapping method using a pre-conditioning RF pulse. This method depends on the amplitude of signal change in a SS-Pre image with respect to the flip angle. Therefore, the pre-conditioning RF pulse has to be chosen carefully. In this method, a slice-selective sinc pulse was used as a pre-conditioning RF pulse because of its relative insensitivity to off-resonance. From the phantom experiment with the resonance offset ranging from 0-500 Hz (100 Hz steps), the RMSE of the κ measurement was less than 1.6% up to 500Hz off-resonance. Since the clinically observed B_0 offset within the body at 3T is on the order of ± 150 Hz (32), this method can provide a considerably B_0 -insensitive result for most applications at 3T. In addition, the slice profile of the SS-Pre pulse should be as close as possible to an ideal slice profile, in order to limit errors in the B_1^+ measurement due to the flip angle transition at the edges of the slice (33). To minimize the effect of the slice profile, we increased the excited slice thickness of the SS-Pre pulse to be six times that of the imaging excitation pulse. This could be improved further by increasing the time-bandwidth product of the SS-Pre sinc pulse. After a SS-Pre pulse excitation, complete spoiling of transverse magnetization is also desirable, in order to eliminate any residual transversal magnetization. In this study, 2.6-ms-long spoiler gradients were applied in all three directions with the magnitude of net zeroth gradient moment of 135mT/m ms. Immediately after a SS-Pre pulse excitation followed by spoiler gradients, a TurboFLASH imaging sequence was performed without any recovery time, to image the residual longitudinal magnetization. Centric k-space reordering was used to minimize the effect on the image intensity of T_1 recovery during the whole k-space acquisition. From the simulation shown in Fig. 3, we observed that the RMSE of the κ map due to filtering effects was approximately less than 0.5% at $T_1 \geq 500$ ms. Since the clinically observed T_1 range within the body is typically greater than 800ms at 3T (29), the saturation effect on the κ map can be negligible for most applications at 3T.

The main advantage of the SS-Pre method is the considerable scan time reduction compared to DAM or DAM-related methods. While the total scan time for DAM is dependent on both a long TR ($\sim 5T_1$) and the number of phase encoding lines for each of the two acquired images, the total scan time for the SS-Pre method requires only one long TR ($\sim 5T_1$) for a full recovery of the longitudinal magnetization between the PD and SS-Pre image acquisitions. The relaxation delay time can be further reduced to less than or on the order of T_1 with the use of the SR module in order to generate a consistent magnetization state before the next acquisition. In this study, the total image acquisition times of the SS-Pre method were approximately 2s/0.6s for the water phantom, 9s/2s for the agar phantom and 10s/2.3s for the brain and pelvis without/with using the SR module, while the DAM took approximately 153.6s for the T_1 -doped water phantom, 806.4s for the agar phantom, 960s for the brain and 640s for the pelvis.

There are several limitations in the presented study. First, this method allows imaging over a limited range of B_1^+ variation as shown in Fig. 5. Due to the cosine behavior of the signal with respect to $\kappa \cdot \alpha^{\text{nom}}$ in the SS-Pre image, the lower bound of κ is limited approximately at $\kappa \cdot \alpha^{\text{nom}} \approx 20^\circ$ where the signal becomes more sensitive to the noise for lower SS-Pre pulse excitation. In this study, the lower bound of κ is limited at $\kappa = 0.33$ with a CV of less than 5% for α^{nom} of 60° (Fig. 5a), while the upper bound of κ is limited by the signal rectification near $\kappa \cdot \alpha^{\text{nom}} \approx 90^\circ$ ($\kappa \approx 1.45$). By adding Gaussian noise (Figs. 5b-5c), the lower bound of $\kappa \cdot \alpha^{\text{nom}}$ was changed to approximately 30° and 41° ($\kappa = 0.50$ and 0.68) with a CV of less than 5% for SNR = 300 and 100, respectively. Second, the slice-selective sinc SS-Pre pulse is relatively insensitive to large off-resonance at 3T up to 500 Hz with less than 1.6% RMSE in κ (Fig. 4), but larger degrees of off-resonance can be encountered at higher fields (≥ 7 T). For example, the magnitude of the B_0 variation can reach up to 3 ppm

(approximately 383 Hz at 3T and 894 Hz at 7T) in the vicinity of air/tissue interfaces in the brain (7). Therefore, the SS-Pre method is applicable for most applications at 3T and 7T, but the user has to be aware of potentially large off-resonance effects in specific regions at 7T. Third, chemical shift artifacts can produce lower-intensity signals at regions of water-fat transitions. In this study, we considered two types of chemical shift artifact (27): one during readout and the other from the differential spatial shifts of fat between the pre-conditioning pulse and the imaging pulse. For the first one, we increased the readout bandwidth to 1500 Hz/pixel in order to minimize the chemical shift artifact to 0.3 pixel in a 64×48 image matrix, but at the expense of a lower SNR in the images (34). For the second one, we increased the slice thickness of the SS-Pre pulse to be six times that of the imaging pulse in order to compensate for the differential spatial shifts of fat between the pre-conditioning and imaging pulses (27). Fourth, validation of the SS-Pre method in the body is rendered difficult because the long scan time of DAM is not compatible with a single breath-hold unless a multi-echo pulse sequence is used (3,13,15-16). In this paper, the SS-Pre method has been validated against DAM in phantoms and in vivo brains. In addition, we performed pelvic imaging to show that in a body region with minimal respiratory motion, a good agreement was found between the SS-Pre method and DAM. Therefore, the SS-Pre method can be used for body imaging within a very short breath-hold, which is of particular interest for applications such as cardiac imaging.

In conclusion, the proposed SS-Pre method provides a rapid and efficient B_1^+ mapping technique. With incorporation of the SR module, it can overcome the long relaxation delay time otherwise required between two image acquisitions, so that the total scan time is significantly reduced, e.g., to less than approximately 2s for most subjects at 3T. Therefore, it can be used for a variety of applications, including body imaging applications where fast imaging is desirable.

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Figure 1.

Sequence diagram of the SS-Pre method. The SS-Pre sequence is divided into three parts: (optional) a saturation-recovery (SR) pulse, a SS-Pre pulse with associated spoiler gradients and TurboFLASH readout sequence. TD_{SR} : SR time delay between SR and SS-Pre pulses. α^{nom} : excitation pulse of SS-Pre pulse. α^0 : excitation pulse of TurboFLASH.

Figure 2. Experiment of dynamic range of κ depending on the α^{nom} of 40°, 60° and 80°. (a) Mean and (b) corresponding coefficient of variation of the measured κ within the ROI, plotted against the nominal κ ranging from 0.2-1.8 (0.2 steps), for α^{nom} .

Figure 3.

Simulation of the saturation effect during the acquisition of the whole k-space lines depending on T_1 . (a) A brain template image indicating the ROI (white line). (b) Plots of the simulated normalized magnetizations (M_z/M_0) using a centric k-space reordering for a SS-Pre (solid line) and a PD (dashed line) images with $T_1 = 0.5s$ (gray line) and $2s$ (black line). (c) The RMSE of the κ maps were plotted against the T_1 values. The RMSE was approximately 0.52%, 0.27%, 0.18%, 0.13%, 0.1% and 0.08% for the T_1 values ranging from 500-3000ms (500ms steps), respectively.

Figure 4. RMSE of κ compared to reference DAM measured on resonance within the whole phantom, plotted against off-resonance conditions ranging from 0-500Hz (100Hz steps).

Figure 5.

(a) Coefficients of variation of κ within the ROI, plotted against the nominal κ ranging from 0.2-1.8 (0.2 steps) affecting only the SS-Pre pulse, only the TurboFLASH pulse, or all RF pulses. (b) CV and (c) corresponding mean of κ within the ROI, plotted against the nominal κ affecting all RF pulses when the SNR of the PD image is decreased to approximately 300 and 100 by adding Gaussian noise. The same level of Gaussian noise was added to both the PD and SS-Pre images for each specific SNR.

Figure 6.

T₁-doped water phantom (T₁ = 320 ms): (a) Reference DAM κ map and (b) corresponding κ map from the SS-Pre method. (c) Corresponding difference κ map and (d) vertical profiles of κ obtained along the white line in (a). Agar phantom (T₁ = 1680 ms): (e) Reference DAM κ map and (f) corresponding κ map from the SS-Pre method. (g) Corresponding difference κ map and (h) vertical profiles of κ obtained along the white line in (e).

Figure 7.

(a) Representative in vivo sagittal brain PD image. (b) Reference DAM κ map and (c) corresponding κ map from the SS-Pre method. (d) Corresponding difference κ map. Profiles of κ are plotted along (e) the vertical and (f) the horizontal lines shown in (a).

Figure 8. Pelvic imaging: (a) Representative pelvic image indicating the ROI (green) and profile lines. (b) Reference DAM κ map and (c) corresponding κ map from the SS-Pre method. (d) Corresponding difference κ map. Profiles of κ are plotted along (e) the vertical and (f) the horizontal lines shown in (a).

Table 1

RMSE of the κ maps of the SS-Pre method and the SS-Pre method with the SR module (SR SS-Pre) for the water and agar phantoms.

Phantom	Water	Agar
SS-Pre	0.008	0.007
SR SS-Pre	0.010	0.016

Table 2
RMSE of the κ maps of the SS-Pre method and the SS-Pre method with the SR module (SR SS-Pre) for the brain and pelvic ROIs.

Brain*	S1	S2	S3	S4			
SS-Pre	0.023	0.033	0.028	0.015			
SR SS-Pre	0.028	0.036	0.030	0.020			
Pelvis■	W1	W2	W3	W4	M5	M6	M7
SS-Pre	0.045	0.061	0.041	0.051	0.075	0.046	0.059
SR SS-Pre	0.046	0.065	0.041	0.056	0.077	0.044	0.065

* S1-S4 indicate the four individual volunteers for brain imaging.
■ W1-W4 indicate female volunteers and M5-M7 indicate male volunteers for pelvic imaging.