Measurement and Characterization of RF Nonuniformity Over the Heart at 3T Using Body Coil Transmission

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Purpose: To measure and characterize variations in the transmitted radio frequency (RF) (B1+) field in cardiac magnetic resonance imaging (MRI) at 3 Tesla. Knowledge of the B1+ field is necessary for the calibration of pulse sequences, image-based quantitation, and signal-to-noise ratio (SNR) and contrast-to-noise ratio (CNR) optimization.

Materials and Methods: A variation of the saturated double-angle method for cardiac B1+ mapping is described. A total of eight healthy volunteers and two cardiac patients were scanned using six parallel short-axis slices spanning the left ventricle (LV). B1+ profiles were analyzed to determine the amount of variation and dominant patterns of variation across the LV. A total of five to 10 measurements were obtained in each volunteer to determine an upper bound of measurement repeatability.

Results: The amount of flip angle variation was found to be 23% to 48% over the LV in mid-short-axis slices and 32% to 63% over the entire LV volume. The standard deviation (SD) of multiple flip angle measurements was <1.4° over the LV in all subjects, indicating excellent repeatability of the proposed measurement method. The pattern of in-plane flip angle variation was found to be primarily unidirectional across the LV, with a residual variation of ≤3% in all subjects.

Conclusion: The in-plane B1+ variation over the LV at 3T with body-coil transmission is on the order of 32% to 63% and is predominantly unidirectional in short-axis slices. Reproducible B1+ measurements over the whole heart can be obtained in a single breathhold of 16 heartbeats.

Key Words: cardiac MRI; RF nonuniformity; saturated double angle method; high-field MRI; prescan calibration

AS CARDIAC MRI moves to higher field strengths such as 3T, imaging protocols require careful consideration of possible nonuniformity of the transmitted radio frequency (RF) (B1+) field. Knowledge of this nonuniformity is crucial for pulse sequence calibration, image-based quantitation (such as in first-pass myocardial perfusion imaging), signal-to-noise ratio (SNR) and contrast-to-noise ratio (CNR) optimization, and the design of new pulse sequences. Estimated variations of 30% to 50% in the B1+ field over the heart at 3T have been reported in the literature (1–3). However, the analysis of in vivo B1+ variations in the chest has been limited by the lack of time-efficient B1+ mapping techniques.

There are several existing methods for B1+ mapping in static body regions (2–8). One of the simplest and the most straightforward methods is the double angle method (DAM) (6,7), which involves acquiring images with two nominal flip angles (operator prescribed values) α and 2α. The method uses the trigonometric double angle formula to determine the true flip angle and requires a long repetition time (TR ≫ T1) to ensure full relaxation before α and 2α pulses.

Cunningham et al (3) recently proposed the saturated double-angle method (SDAM), which permits rapid B1 mapping with TR < T1. A saturation pulse at the end of each data acquisition resets the longitudinal magnetization to a known state. The B1+ field is still derived from the ratio of signal magnitudes after α and 2α pulses. SDAM with TR < T1 was successfully validated against DAM with TR ≫ 5T1 in phantoms and static body regions. The feasibility of rapid cardiac B1 mapping was shown but the methodology was susceptible to irregular heart rates and B0 and B1 variation.

In this work, we implemented a variation of SDAM for B1+ measurement across the entire heart in a single breathhold by considering the B0 and B1 sensitivity of saturation pulses, possible variations in heart rate, and possible cross-talk between slices in a multislice acquisition. We measured and analyzed whole-heart B1+ profiles from 10 subjects using two 3T scanners with body-coil transmission. Repeatability testing was performed to determine practical utility of the measurement methodology for prescan calibration. Profile analysis was performed to determine patterns of variation that may be exploited during B1+ shimming.
**MATERIALS AND METHODS**

**Cardiac SDAM**

Successful multislice cardiac B1+ measurement at 3T using SDAM requires: 1) robust magnetization saturation in the presence of the B0 and B1+ inhomogeneity; 2) identical saturation recovery time (T2*) even during irregular heartbeats (R-R intervals); and 3) minimization of cross-talk between adjacent slices in an interleaved multislice acquisition.

To address the first issue, we examined the effectiveness of various saturation pulses. The performance of saturation pulses as a function of B0 and B1+ inhomogeneity was investigated (3), and we selected an adiabatic composite pulse (BIR-4) (9) to support a B0 bandwidth of ± 130 Hz, typical for 3T cardiac imaging (10). Although the BIR-4 pulse (8 msec) has longer pulse duration and higher RF power than conventional pulses, its effects are minimal in this work because the pulse is applied only once per R-R interval and its duration is relatively small compared to the total image acquisition time (80 msec).

The second issue deals with variations in the R-R interval. When the saturation pulse is timed at the end of data acquisition (3), even a small change in heart rate produces variation in T2*, causing an error in the true flip angle calculation. In this work, we timed the saturation pulse with a prospective triggering signal to make T2* independent of heart rate, producing more robust B1+ maps.

The third issue deals with multislice acquisitions, for which B1+ measurements may be affected by flow-induced cross-talk between slices. If a TR for each slice is too long or an inter-slice gap is too small, cross-talk can occur because the through plane flow can reach to the adjacent slice. If a TR for each slice is too long or an interslice gap is too small, cross-talk can occur. Therefore, we introduced a minimum gap of 15 mm. For TR of 11 msec, a minimum gap is required to avoid cross-talk for through plane flow (3).

The pulse sequence, shown in Fig. 1, consists of three modules: the RESET module, the FATSAT module, and the IMAGING module. The IMAGING module consists of a fat-selective excitation (8 msec) followed by a dephaser. The FATSAT preparation consists of a fat-selective saturation, a short spiral readout (5.9 msec), and a dephaser. Note that the IMAGING excitation is slice selective, while the RESET and FATSAT excitations are not spatially-selective.

**Figure 1.** Cardiac SDAM pulse sequence. Full images are acquired with a nominal 2α and nominal α flip angle. a: Acquisitions are cardiac gated in a single breathhold to prevent motion artifacts, and consist of a magnetization reset, delay (T2*), fat saturation, and multislice acquisition. b: The RESET module consists of an 8-msec BIR-4 saturation pulse followed by a dephaser. c: The FATSAT preparation consists of a fat-selective saturation followed by a dephaser. d: The IMAGING module consists of a slice-selective excitation, a short spiral readout (5.9 msec), and a dephaser. Note that the IMAGING excitation is slice selective, while the RESET and FATSAT excitations are not spatially-selective.

The flip angle calculation in the double angle approach (3,6–8) follows:

\[ \hat{\alpha}(x,y) = \arccos \left( \frac{I_p(x,y)}{2I(x,y)} \right). \]  

(2)

where \( I_p(x,y) \) is the measured flip angle map and \( I(x,y) \) and \( I_p(x,y) \) are the base magnitude images with the nominal flip angle of \( \alpha \) and 2\( \alpha \). Image nonuniformities except for the flip angle-induced variation are identical for both base magnitude images, and therefore will be cancelled out in Eq. [2].

Since \( I_p(x,y) \) and \( I(x,y) \) contain image noise, \( \hat{\alpha}(x,y) \) can be decomposed into the true flip angle map \( \alpha(x,y) \) and an additional noise term \( \bar{n}_p(x,y) \)

\[ \hat{\alpha}(x,y) = \alpha(x,y) + \bar{n}_p(x,y). \]  

(3)

where \( \bar{n}_p(x,y) \) can be analyzed using second order statistics: mean \( \bar{E}(\bar{n}_p(x,y)) \) and standard deviation (SD) \( \sigma_{\bar{n}}(x,y) \). Both \( \bar{E}(\bar{n}_p) \) and \( \sigma_{\bar{n}} \) depend on \( \alpha \) and the base image SNR, \( \sigma_n \) and \( E(n) \) were simulated as a function of \( \alpha \) with different base image SNRs. According to simulations for a base image SNR range of 70–120, \( \hat{\alpha} \) was close to \( \alpha \) with relatively small \( \sigma_n \) and \( E(n) \), when \( \alpha \) was larger than 20°.

**Experimental Methods**

Experiments were performed on two identical GE Signa 3.0T EXCITE systems (General Electric Healthcare, Waukesha, WI, USA) with gradients capable of 40 mT/m amplitude and 150 T/m/second slew rate, and receivers supporting 4-μsec sampling (±125 kHz). Quadrature birdcage body coils (60-cm diameter and 32 rungs) were used for RF transmission and an eight-channel cardiac phased array coil was used for signal reception. Parallel imaging was not used. The transmit gain was calibrated using the vendor-supplied prescan process. The acquisition parameters were as follows:
Cardiac B1+ Measurements at 3 Tesla

Figure 2. Cardiac flip angle maps from one representative volunteer at 3T. All six short-axis slices were acquired in a single 16 R-R breathhold. A mesh plot (top) depicting the flip angle variation in a mid-short-axis slice (slice #3) and flip angle profiles (bottom) in all six slices. Magnitude images with a nominal flip angle of 60° are provided for anatomical reference. Note that the variation appears strong and primarily unidirectional.

sinc RF pulse = 2.6 msec, TE = 2 msec, TR = 11 msec, field of view (FOV) = 30 cm, in-plane resolution = 2.2 mm, and slice thickness = 5 mm. Only the first 1.9 msec of each spiral readout was used during reconstruction by applying an appropriately sized Hamming window to the raw k-space data. This increased the base image SNR while reducing the in-plane spatial resolution to 5 mm.

Cardiac B1+ measurements were acquired in eight healthy volunteers and two cardiac patients (eight males and two females, ages = 24–71 years, weights = 55–86 kg, resting heart rates = 49–81 bpm). The imaging protocols were approved by the Institutional Review Board of the University of South Carolina. Each subject was screened for MRI risk factors and provided informed consent in accordance with institutional policy. Scan plane localization was performed using the GE I-drive real-time system. In each volunteer, six parallel short-axis slices were prescribed spanning the LV from base to apex. The basal slice was denoted as #1 and the apical slice was denoted as #6. Whole-heart B1+ mapping was achieved in a single breathhold of 16 R-R intervals. Synchronization with the cardiac cycle was achieved with prospective triggering based on either an infrared plethysmograph (five subjects) (11) or an ECG (five subjects) signal.

A total of 10 repeated measurements were obtained in each of the healthy volunteers and five repeated measurements were obtained in each of the cardiac patients in separate breathholds with the same scan plane prescription. Subjects were instructed to perform each breathhold in a comfortable exhaled position. The subjects had no prior training for consistent breathhold positioning. Measurements containing at least one missed trigger were excluded from the data analysis, resulting in eight to 10 useful scans for each of the normal volunteers and five useful scans for each of the patients.

Data Analysis

All data analysis was performed in MATLAB 7.0 (The Mathworks, Natick, MA, USA). Circular regions of interest (ROIs) covering left ventricular myocardium and blood pool were manually defined based on magnitude images. Based on \( \hat{\alpha}(x,y) \), the percentage of flip angle variation was defined as

\[
\text{Variation} = \frac{\alpha_{\text{max}} - \alpha_{\text{min}}}{\alpha_{\text{max}}} \times 100\%	ext{.} \tag{4}
\]

We computed the mean and SD across measurements on the same subject and under the same conditions (imaging parameters and scan plane). The repeatability test determined an upper bound of measurement repeatability and indicated the usefulness of the technique for prescan calibration. Factors that likely contributed to the variation in measurements were \( n_a(x,y) \) and inconsistent breathhold positions. Pixel-by-pixel mean \( E(\hat{\alpha}(x,y)) \) and SD \( \sigma_a(x,y) \) of the flip angle maps based on the five to 10 measurements were computed to provide the repeatability. The pixel based \( \sigma_a(x,y) \) was then averaged over the ROI to produce a single number:

\[
A = E(\sigma_a(x,y)), x,y \in \text{ROI}, \tag{5}
\]

as an indicator of overall repeatability.

After observing that the in-plane flip angle variations were primarily unidirectional over the LV, we attempted to model the measured flip angle profile, \( \hat{\alpha}(x,y) \), with a 1D approximation in each short-axis slice. We computed minimum mean squared error (MMSE) 1D approximations, \( \hat{\alpha}(x') \), for in-plane orientation with a 1° increment and defined the primary in-plane axis as the one that minimized the approximation error, \( B \):

\[
B = E\left(\frac{|\hat{\alpha}(x,y) - \hat{\alpha}(x')|}{|\hat{\alpha}(x,y)|}\right) \times 100\%. \quad x,y \in \text{ROI}. \tag{6}
\]

Figure 3. Illustration of repeatability analysis. Pixel-by-pixel (a) mean (gray scale 20°–70°) and (b) SD (0°–10°) of the measured flip angle map in a short-axis slice (slice #3 in subject #1). The metric \( A \) is computed by averaging \( \sigma_a(x,y) \) over the ROI (white circle).
RESULTS

Representative flip angle maps for all six short-axis slices in one healthy volunteer are shown in Fig. 2. For a nominal flip angle of 60°, the observed flip angles across the LV myocardium ranged from 32° to 64°. The percentage variations within 2D short-axis slices ranged from 29% to 48% in this volunteer. One can qualitatively observe that the flip angle variation appears to be significant, smooth, and primarily along one in-plane axis.

Figure 3 illustrates the measurement repeatability testing. The pixel-by-pixel mean $E(\hat{\alpha}(x,y))$ is assumed to be close to true flip angle profile $\alpha(x,y)$ and $\sigma_r(x,y)$ depicts the variation among measurements. The pixel-by-pixel SD within the ROI was less than 1.4° in all subjects, which suggests that the measurement methodology produces highly repeatable results.

Figure 4 illustrates the profile analysis used to determine the dominant direction of variation. The vertical line was selected from the magnitude image and the in-plane angle $\theta$ was defined as a clockwise angle from the vertical line to an axis of variation in degrees. The approximation error as a function of $\theta$ has one clear minima, which indicates a unique primary axis of variation. The approximation errors between the true (2D) profile and “best” unidirectional (1D) approximation were $\leq 3.1\%$ in all subjects, and were $\leq 1.5\%$ in eight of the subjects. The observed profiles were consistent for both 3T scanners.

Figure 5 contains mid-short-axis flip angle maps and illustrates the ROI analysis for all 10 subjects. Magnitude images (background) are shown for anatomical reference. The amount of flip angle variation over the 3D LV volume for the 10 subjects ranges from 31% to 66% while the flip angle variation over the LV in 2D mid-short-axis slices ranges from 23% to 53%. Flip angle maps from all subjects exhibit unidirectional variation along one primary in-plane axis.

DISCUSSION

The observed unidirectional trend at 3T may be used to compensate in-plane B1 inhomogeneity using tailored radiofrequency (TRF) pulses (12,13). The presence of unidirectional in-plane B1+ variation greatly simplifies the design of compensating pulses, and enables the design of exceptionally short pulses (13). Further reductions in the duration of compensating pulses can be accomplished using parallel RF transmission (14).

The IMAGING module can incorporate spin-echo schemes or alternate k-space segmentations such as echo planar imaging (EPI). Spin-echo approaches are able to refocus T2* related inhomogeneities and may provide better noise behavior at lower flip angles (8) but are difficult to combine with interleaved multislice imaging. Although spiral readouts were chosen due to their efficiency and short echo time to reduce T2* effects, EPI readouts may also be used.

The cardiac B1+ measurement may have an important clinical role during prescan calibration at 3T. Kim

Figure 4. Illustration of profile analysis. a: The approximation error, $B(\theta)$, as a function of in-plane angle $\theta$, defined in degrees clockwise from the vertical axis. b: Flip angle variation $\hat{\alpha}(x,y)$ in a circular ROI can be approximated by (c) a 1D function $\hat{\alpha}(x')$ along the primary in-plane axis (dotted line). The primary in-plane axis angle is 108° and the approximation error is 1.2% in this example (slice #3 in subject #1).

Figure 5. Mid-short-axis B1+ maps in all 10 subjects (eight healthy volunteers: 1–8 and two cardiac patients: 9 and 10). Magnitude images with a nominal flip angle of 60° are included for anatomical reference. Within each ROI (white circle), the primary in-plane axis of variation is indicated (dotted line). The flip angle variation in mid-short-axis slices was found to be 23% to 53%.
et al (15) demonstrated that B1+ inhomogeneity can result in regional contrast variations across the heart. Uniform saturation may potentially be achieved if cardiac B1+ measurements are used to guide the choice of B1 amplitude. Knowledge of the B1+ field may also improve the calibration of time–intensity curves in first-pass myocardial perfusion imaging along with several existing methods to correct intensity variations (16,17). Finally, an ideal flip angle in balanced steady-state free precession (SSFP) imaging is often precalculated to achieve the highest contrast between tissues (18) and provides a derived value ranges from 0° to 90°. The calculation assumes a linear relationship between flip angles and B1+ fields and the deviation from the relationship must be taken into account for larger flip angles (> 70°) (7). In addition, the repeatability may become worse for smaller flip angles (< 20° in this work). These limitations should be carefully considered when severe transmitted B1+ variation is expected.

In conclusion, this study has done the following: 1) verified that the proposed SDAM pulse sequence produces reproducible cardiac B1+ maps over the entire heart in a single breathhold; 2) shown the flip angle variation at 3T to be approximately 32% to 63% over the 3D left ventricle, and approximately 23% to 48% within the 2D short-axis slices; and 3) established that the in-plane variation is primarily along one axis in short-axis slices.

The measurement variations found here can be considered upper bounds on the true variation of the measurement methodology. Further improvements in repeatability may be achieved by 3D image registration and wavelet-based denoising (20), which could suppress contributions from breathhold inconsistency and the additive noise term. The difference in overall quality may be subtle due to the low image resolution in this work.

During acquisition, we used a spatial resolution of 2.2 mm with eight-interleaved 5.9-msec readouts, and during data analysis we windowed k-space data, which reduced the effective resolution to 5 mm. In future studies, the same spatial resolution can be achieved with shorter 1.9-msec readouts and eight interleaves, or with 5.9-msec readouts and just three interleaves. The low resolution condition may assist to either shorten the breathhold duration (by a factor of 3) or increase the number of slices with the cross-talk–free through-plane velocities up to 2.14 m/second.

**REFERENCES**