Image-Guided Radio-Frequency Gain Calibration for High-Field MRI

E. Breton[1], K. McGorty[1], G. C. Wiggins[1], L. Axel[1], and D. Kim[1]
[1]Research Radiology - Center for Biomedical Imaging, New York University Langone Medical Center, New York, NY, United States

Introduction: High-field (≥ 3T) MRI provides a means to increase the signal-to-noise ratio compared with 1.5T, as well as to achieve better spectral separation. However, both the static magnetic field (B_0) and the transmit radio-frequency (RF) field (B_1^+) inhomogeneities are comparatively higher than those at 1.5T. In conventional MRI, an automated RF calibration procedure is performed as a pre-scan to calibrate the RF excitation. However, it may not necessarily produce the expected flip angle within a region-of-interest (ROI). An incorrect calibration of the transmit gain will produce the wrong flip angle and consequently altered image contrast, as well as measurement errors for quantitative MRI and MR spectroscopy. In addition, an incorrect transmit gain calibration could lead to excessive specific absorption rate (SAR) for the applied sequence compared to what would be the case for the correct (if lower) RF calibration. One approach to correct for B_1^+ inhomogeneities is to perform B_1^+ mapping. However, B_1^+ mapping pulse sequences are generally time consuming and require off-line image reconstruction and analysis, rendering them impractical for routine clinical MRI. The purpose of this study was to develop a rapid, image-guided RF gain calibration procedure for high-field MRI and evaluate its performance through phantom and *in vivo* experiments at 3T and 7T.

Methods: A series of "saturation-no-recovery" images are acquired immediately following a preconditioning pulse with varying flip angle [1]. The resulting images depict the residual longitudinal magnetization (M_z) left behind by the preconditioning RF pulse, so that a signal null occurs in region where the effective flip angle of the preconditioning pulse is 90º. To locally calibrate the RF gain, a user can thus choose the preconditioning RF amplitude that produced the signal null within a desired ROI and correspondingly scale the RF transmitter gain calibration. The preconditioning RF pulse was designed as a slice-selective sinc pulse with variable flip angle, fixed pulse duration = 2.8 ms, time-bandwidth product = 6, slice-thickness = 6 times that of the imaging slice, and transmitter bandwidth = 2.1 kHz. Right after the preconditioning pulse, 3-ms-long spoiler gradients were applied, which were immediately followed (i.e., "no recovery") by a TurboFLASH sequence with centric k-space ordering (Fig. 1). As suggested by Cunningham et al [2], an effective saturation-recovery (SR) module [3] with recovery time less than $5T_1$ was used to achieve a constant M_z prior to the preconditioning RF pulse module, in order to accelerate data acquisition. Relevant imaging parameters include: FOV = 250-340mm × 81.3% in phase-encoding direction, matrix = 128 × 104, slice thickness = 8 mm, TE/TR =1.2/2.4 ms, flip angle = 10°, GRAPPA reconstruction with effective acceleration factor = 1.6, receiver bandwidth = 1565Hz/pixel, image acquisition time ~ 150ms, total scan time per measurement = 1000 ms, and repetition = 10. This pulse sequence was implemented on our 3T and 7T whole-body MR scanners (MAGNETOM Tim Trio and 7T, Siemens, respectively). For 3T acquisitions, the body coil was used for RF excitation, and phased array coils were used for signal reception. For 7T acquisition, a transmit/receive (transmit birdcage; 24 receive elements) head coil (Nova Medical) was used. By definition, the initial flip angle scale factor of the preconditioning pulse (κ) is the ratio of the nominal flip angle and nominal 90° determined by the standard RF calibration. For quantitative analysis of the image-guided RF calibration data, the mean signal intensity was plotted as a function of κ, in order to accurately identify the signal null in the ROI. For larger ROI with relatively large B_1^+ variation, the mean signal vs. κ curve may not produce a clear minimum, but rather a broad global minimum, because of signal variation within the ROI. In such a case, the standard deviation (SD) of the signal was used to identify the κ value that produces minimum signal dispersion within the ROI, and this κ value that coincides with a minimum of the mean signal vs. κ curve was determined to be the best value for RF calibration (Fig. 2B-C). The performance of the image-guided RF calibration procedure was evaluated in phantoms and in volunteers at 3T and 7T. For phantom imaging, standard B_0 and B_1^+ mapping pulse sequences were also performed to validate the image-guided RF calibration pulse sequence. For initial in vivo application, the influence of the RF calibration on T_2 measurements using an optimal multi-echo fast spin echo (ME-FSE) pulse sequence with reverse centric k-space reordering [4]. For the same subject, the T_2 mapping pulse sequence was performed twice, first using the transmit RF gain calculated by the standard automated calibration procedure, and second using the correct transmit RF gain calculated by the image-guided RF calibration procedure.

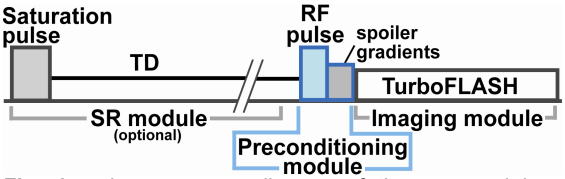

Fig. 1: pulse sequence diagram of the proposed image-guided RF calibration procedure.

Results: In the phantom imaging at 3T and 7T, image-guided RF calibration produced results that were consistent (< 5% error) with standard B_1^+ mapping pulse sequence, up to 300 Hz B_0 off-resonance. In one volunteer at 3T, the standard RF calibration produced approximately 0 and 20% error in RF transmit gain for the right and left kidney, respectively. In a sagittal plane of the brain at 7T (Fig. 2), signal null within the thalamus and whole brain occurred at κ = 0.7 and 0.85, respectively, suggesting 30% and 15% relative error in their respective transmitter gain. The mean T_2 measured in the thalamus, using the RF gain calculated with the standard-automated and the image-guided calibration procedures, were 60.6ms and 48.2ms, respectively. These results suggest that the standard RF calibration resulted in 25% error in T_2.

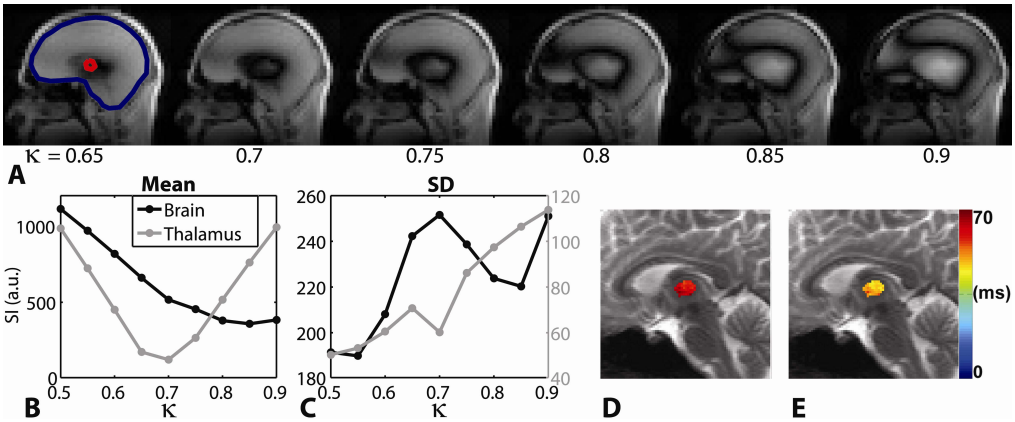

Fig. 2: (A) Series of "saturation-no-recovery" images acquired at 7T for κ = 0.65-0.9, step 0.05. The brain ROI appears in blue, the thalamus ROI in red. Mean signal intensity (SI) (B) and standard deviation (SD) (C) vs. κ curves measured in the brain and thalamus. T_2-map obtained in the thalamus using the standard-automated (D) and the image-guided (E) RF calibration procedures.

Conclusion: The proposed rapid image-guided RF transmitter calibration can be used to optimally calibrate the RF transmit gain for a given ROI within the FOV, avoid excessive RF power deposition, and minimize measurement errors for quantitative MRI and spectroscopic applications. It can easily be added to existing clinical protocols. The quantitative analysis of the RF calibration images can be performed on commercial scanners using readily available simple descriptive statistics tools.

References: [1] Klose U. et al., Med Phys 1992; 19:1099-104. [2] Cunningham C.H. et al., MRM 2006; 55:1326-33. [3] Kim D. et al., MRM 2009; DOI: 10.1002/mrm.22140. [4] Kim D. et al., MRM 2009; 62:300-6.

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