Rapid Proton Density Weighted Abdominal MRI at 3 Tesla With RF Non-Uniformity Correction

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Introduction – In high-field abdominal imaging, RF non-uniformities due to the use of multi-coil receivers, large imaging FOV, and the dielectric and resistive properties of the human body, collectively cause signal intensity variations across the reconstructed image [1-3]. Corrections of these variations can improve diagnostic image quality, and are essential in applications where quantification based on signal intensity is performed. This work presents a retrospective approach to correct RF transmit and receive inhomogeneity for rapid proton-density-weighted (PDW), low-flip-angle (LFA), spoiled-gradient-echo (SPGR) MRI. We hypothesize that a set of rapid breath-hold scans can be acquired to provide RF transmit and receive maps across the abdomen. This information will then be used to remove RF-induced signal intensity variations in the PDW-LFA-SPGR data. We describe our approach and demonstrate its feasibility, and then discuss the applicability of the method for fat quantification in abdominal MRI.

Methods - For low-flip-angles (θ), the SPGR signal is approximately a linear function of proton density P and θ [4] (Eqn. 1). We assume the imaging scenario where the body coil is used for RF transmission, while a multi-coil array is used for signal reception. Our signal-intensity compensation model is presented by Eqn. 2, where S_{raw} denotes the common "sum-of-squares" combined image, which typically exhibits signal shading from non-uniform body coil transmission and surface coil reception. $B+$ and $B-$ are the transmit and receive patterns of the body coil, while $X-$ is the receive pattern of the multi-coil array. In Eqn. 2a, the terms in parenthesis are the proposed correction factors to be measured. ($B-/X-$) is the relative sensitivity of the body coil with respect to the multi-coil array, and ($1/B+$) is the reciprocal of the body coil transmit map, where $B+ = \theta_{actual} / \theta_{nominal}$. To fully correct S_{raw}, a third ($1/B-$) term is needed to account for the $B-$ term in ($B-/X-$). Using the principle of reciprocity between $B+$ and $B-$, Eqn. 2a simplifies to 2b.

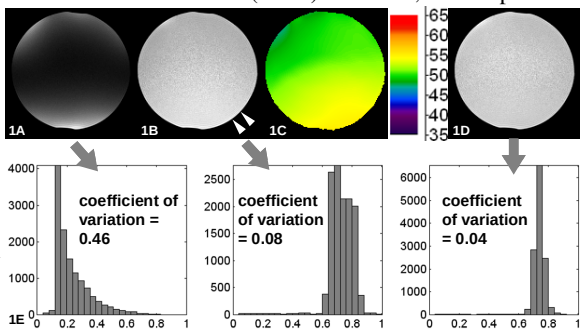
$$[1]\quad S_{SPGR} \approx P \cdot \theta$$

$$[2a]\quad S_{corrected} = S_{raw} \cdot \left(\frac{B-}{X-} \cdot \frac{1}{B+} \cdot \frac{1}{B-} \right)$$

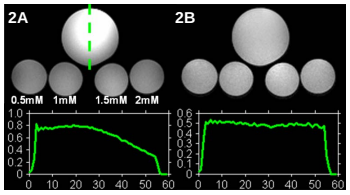
$$[2b]\quad S_{corrected} = S_{raw} \cdot \frac{B-}{X-} \cdot \frac{1}{(B+)^2}$$

Correction terms – ($B-/X-$) is the ratio of two low-resolution scans of the imaging object using separately the body and array receivers. ($B-/X-$) is used in parallel imaging reconstruction to compute coil sensitivities, and as a step to remove signal bias in "sum-of-squares" images. Since rapid abdominal scans often involve parallel imaging, only a few seconds of additional scan time is needed to obtain ($B-/X-$). For $B+$, we adopted the saturated double-angle-method (SDAM) [5-6] with 2DFT readout for its rapid speed.

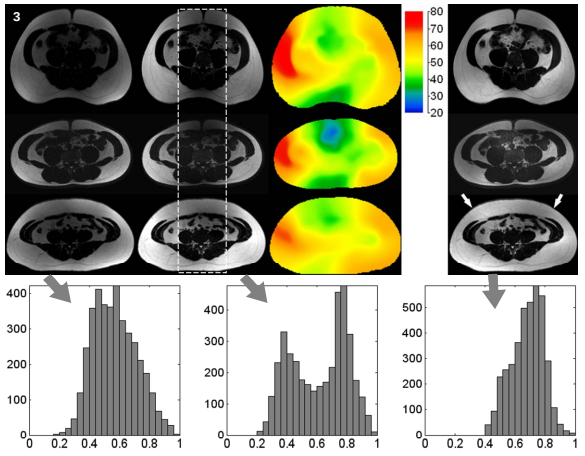
Experiments – A 30-cm sphere and bottles filled with doped water were imaged with an eight element torso-array and a SPGR sequence (TR/TE=3.2/1.4 ms, θ=3°, BW=±62.5 kHz, FOV=34 cm, 160×160 matrix, 10 mm slice). ($B-/X-$) was also obtained with SPGR (TR/TE=2.3/1 ms, θ=5°, BW=±31.25 kHz, 64×64 matrix). SDAM parameters were: T_{SR}=300 ms, $\theta_{nominal}$=60°, BW=±31.25 kHz, 64×32 matrix (see ref. [5-6] for SDAM details). Breath-hold 3D abdominal imaging was performed in three volunteers using IDEAL-SPGR [4] with TR=5.4 ms, TE=2.1, 2.8, 3.5 ms, θ=3°, BW=±62.5 kHz, FOV=40 cm, 192×192×12 matrix, and 10 mm partitions. ($B-/X-$) and SDAM scans took 5 and 20 seconds, respectively. All experiments were performed on a GE 3T scanner.



Results – In Fig. 1A, S_{raw} of the 30-cm sphere exhibits a significant amount of signal non-uniformity, a majority of which is removed after ($B-/X-$) correction (Fig. 1B). However, a small amount of signal variation is still noticeable in the receive-corrected image, where the bottom half of the phantom appears slightly brighter (arrowheads). This is confirmed by a similar pattern in the flip angle map (θ_{actual}) computed by SDAM (Fig. 1C). Further correction with $(1/B+)^2$ yields a more uniform phantom in $S_{corrected}$ (Fig. 1D). Histograms in Fig. 1E corroborate these qualitative observations, where the coefficient of variation (CV=standard deviation/mean) has been reduced from **0.46** in S_{raw} to **0.04** in $S_{corrected}$. S_{raw} and $S_{corrected}$ images from another phantom are shown Fig. 2A and B, respectively. The four bottles at the bottom of the image contain different concentrations of T_1 contrast agent, but exhibit the same signal intensity due to LFA-PDW-SPGR. Line profiles across the top bottle reinforce the improvement in signal uniformity with our proposed model. *In vivo* results are presented in Fig. 3, where the columns show S_{raw}, S_{raw}·($B-/X-$), θ_{actual}, and $S_{corrected}$, respectively. Considering the rim of subcutaneous fat (arrows) in the third row, CV within this region was reduced from **0.25** in S_{raw} to **0.15** in $S_{corrected}$. Note in column two that signal shading near the center of the abdomen (dashed box) has been compensated in $S_{corrected}$ to allow improved visualization of the visceral fat. Note also some similarities in θ_{actual} patterns, despite differences in body habitus.



Discussion – The development of the proposed model is motivated by our interest in abdominal fat quantification for obesity research. In a separate study, we are validating algorithms that can compute **absolute fat mass** based on the proton density of lipids in adipose tissue, which in turn is computed with signal intensity. Therefore, the accuracy of fat mass calculations depends critically on correct fat signal intensities that are free of RF-induced variations, not only in voxels containing pure fat, but also in organs where fat and lean tissues are mixed. We have shown in this work that our approach removed RF effects in phantoms. When applied *in vivo*, the model was able to remove significant signal shading near the abdominal midline. However, some residual signal bias still remained within the fat tissue after correction. Tissue heterogeneity is a potential source of this non-uniformity, although inaccuracies in the RF transmit maps produced by SDAM are also likely culprits. This is an area of ongoing work. Additional *in vivo* studies on the quantitative accuracy of fat mass calculations will reflect the validity of our signal compensation model.



[1] Schmitt M, et al. ISMRM 2004, 197. **[2]** Akisik FM, et al. Radiographics 2007, 27:1433-1444. **[3]** Lee VS, et al. Radiology 2007, 224:692-705. **[4]** Liu CY, et al. MRM 2007, 58:354-364. **[5]** Cunningham CH, et al. MRM 2006, 55:1326-1333. **[6]** Sung K, et al. ISMRM 2007, 355.