Effects of B_0 and B_1 Inhomogeneity in Ultra-High Field MRI

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Introduction: Ultra-high field (≥ 7 T) MRI offers many advantages over low field MRI, particularly an increased signal-to-noise ratio (SNR), allowing higher spatial resolution, and an enhanced susceptibility contrast [1]. However, it also suffers from more severe static (B_0) [2] and radiofrequency (RF) (B_1) [3] magnetic field inhomogeneity. Severe B_1 inhomogeneity leads to substantial variations of the flip angle and receive sensitivity across the field-of-view (FOV), resulting in variable SNR and image contrast. Overall, both B_0 and B_1 inhomogeneity result in regional signal variation, and it is not intuitive which effects are dominant. The objective of this work is to characterize these effects on representative gradient echo (GE) and spin echo (SE) images of the human brain acquired at ultra-high field strength and relate them to quantitative measures of the B_0 and B_1 inhomogeneity. This knowledge will ultimately help develop better B_0 and B_1 inhomogeneity correction methods.

Methods: The studies were performed on an ultra-high field human whole-body MRI system using a transverse electromagnetic RF head coil with 16 struts and 4 excitation ports individually tuned for each study. We studied 10 healthy volunteers (6 male, 4 female, age 20–53) who gave informed consent and 11 postmortem unembalmed human subjects (5 male, 6 female, age 57–85). Due to the flip angle variability, we first defined a "nominal" flip angle (FA) as the average flip angle in a 1 cm^3 region of interest, determined the transmit power level resulting in a nominal FA of 90° using a voxel-selective stimulated echo sequence, and then used this reference to set the transmit power level corresponding to a chosen nominal FA. High resolution GE images were acquired with: 8 ms sinc RF pulse, TR/TE 600/12 ms, nominal FA 20°, FOV $(18\ cm)^2$, matrix (MTX) 1024x512, and slice thickness (ST) 2 mm. High resolution SE images were typically acquired with TR/TE 1500/70 ms, nominal FA 90°/180°, MTX 512x256, ST 3 mm, and otherwise identical parameters. Axial, coronal, and sagittal images were acquired throughout the brain.

A B_0 map was experimentally measured using a 3D dual-echo GE sequence [4] with: 0.5 ms sinc RF pulse, TR 20 ms, TE 1.2/3.0 ms, nominal FA 10°, FOV $(18\ cm)^3$, and MTX 96^3. For two subjects, the B_0 field was also numerically simulated using a finite difference method and susceptibility distributions obtained from computed tomography images [2]. Maps of the B_0 gradient along x, y, and z were computed to assess the inhomogeneity in the slice direction. To quantify the B_1 inhomogeneity, maps of the local flip angle α and receive sensitivity r were experimentally measured from two series of GE images acquired with $TR >> T_1$ and nominal FA α_0 and $2\alpha_0$. A flip angle map was computed as $\alpha = \cos^{-1}(S_{2\alpha} / 2S_\alpha)$ [5], then a map of ρ r was computed as $S_\alpha / \sin(\alpha)$ and low pass filtered to yield a measure of the receive sensitivity. Typical parameters used were: 8 ms sinc RF pulse, TR/TE 4000/7 ms, nominal FA 60° and 120°, FOV $(18\ cm)^2$, MTX 256x64, and ST 3 mm. The flip angle and receive sensitivity maps were reformatted to yield maps corresponding to the GE and SE images.

Results: Figure 1 shows *in vivo* coronal GE and SE images and the corresponding maps of the B_0 and B_1 inhomogeneity. On the GE image, signal loss in the inferior temporal cortex is due to high B_0 inhomogeneity, as shown by the B_0 gradient maps. This is also seen on axial and sagittal images in this region as well as in the inferior frontal cortex superior to the sphenoid and posterior to the frontal sinuses. Conversely, SE images are largely unaffected by such artifacts. On both GE and SE images, signal loss in the temporal lobes is due to low flip angles and/or receive sensitivity. This is also seen on axial and sagittal images. Signal loss due to B_1 inhomogeneity is more severe on SE than GE images. The *in vivo* images, particularly the SE images, are also affected by severe artifacts due to cerebrospinal fluid flow.

Experimental B_0 mapping is affected by noise and artifacts due to B_1 inhomogeneity, motion, and intravoxel dephasing, which may account for some discrepancies with the simulations. The flip angle mapping method is theoretically accurate for $0° < \alpha < 180°$, but low image SNR introduces errors for flip angles close to 0° or 180°. The measurement may also become inaccurate in regions with high B_0 inhomogeneity. Such errors propagate into the receive sensitivity maps, which are also contaminated by residual proton density weighting, especially in the ventricles. Nevertheless, overall patterns can be assessed and show differences with the flip angle distributions. While different subjects have similar flip angle and receive sensitivity maps, differences arise from variable anatomy and coil tuning.

Discussion: Our studies clearly show that ultra-high field MRI is plagued by substantial signal variation and signal loss. Correlation with B_0 and B_1 mapping shows that these effects can be fully explained by B_0 and/or B_1 inhomogeneity. As expected, signal loss due to B_0 inhomogeneity affects predominantly GE images and is restricted to the vicinity of air/tissue interfaces. B_0 inhomogeneity also affects the effective excitation field $B_{eff} = (B_1^2 + \Delta B_0^2)^{1/2}$ and thus the effective flip angle [6], thereby exaggerating the signal loss. Both GE and SE images show substantial signal variability due to B_1 inhomogeneity. The flip angle and receive sensitivity have non-intuitive complex spatial distributions, which are similar to numerical simulations [7]. Signal loss due to B_1 inhomogeneity affects SE images more severely than GE images because the signal intensity varies as $\sin^3(\alpha)$ and $\sin(\alpha) / [1 - \exp(-TR / T_1) \cos(\alpha)]$ respectively. This predicted signal variability correlates well with measured image SNR.

This discussion demonstrates the utility of B_0 and B_1 mapping for assessment of image artifacts in ultra-high field MRI. Because of the severity of these artifacts, B_0 and B_1 inhomogeneity correction methods are needed. A variety of susceptibility artifact correction methods have been proposed, but their optimization requires knowledge of the B_0 field. While numerical B_0 maps can be used for methods that do not rely on subject specific B_0 maps (*e.g.*, gradient compensation or passive shimming), experimental B_0 maps will be required for other methods (*e.g.*, active shimming or post-processing). Experimental B_1 mapping requiring $TR >> T_1$ is too slow for clinical applications. However, knowledge of the overall characteristics of the B_1 field can help select optimal flip angles. On the other hand, B_1 mapping is crucial for quantitative studies such as T_1 and T_2 measurements and for ultra-high field RF coil design.

Conclusion: Artifacts due to B_0 and B_1 inhomogeneity are severe in ultra-high field MRI and experimental and/or numerical mapping of the B_0 and B_1 inhomogeneity is important in identifying their origin. Detailed characterization of these effects is an important step in the development and assessment of B_0 and B_1 inhomogeneity correction methods.

References: [1] Abduljalil AM. *JMRI* 2003;18:284 [2] Truong T-K. *MRI* 2002;20:759 [3] Ibrahim TS. *Phys Med Biol* 2001;46:2545 [4] Truong T-K. *Proc 11th ISMRM* 2003. p. 1048 [5] Insko EK. *JMR A* 1993;104:78 [6] Haacke EM. Magnetic resonance imaging. New York: Wiley; 1999. p. 47 [7] Ibrahim TS. *MRI* 2001;19:1339

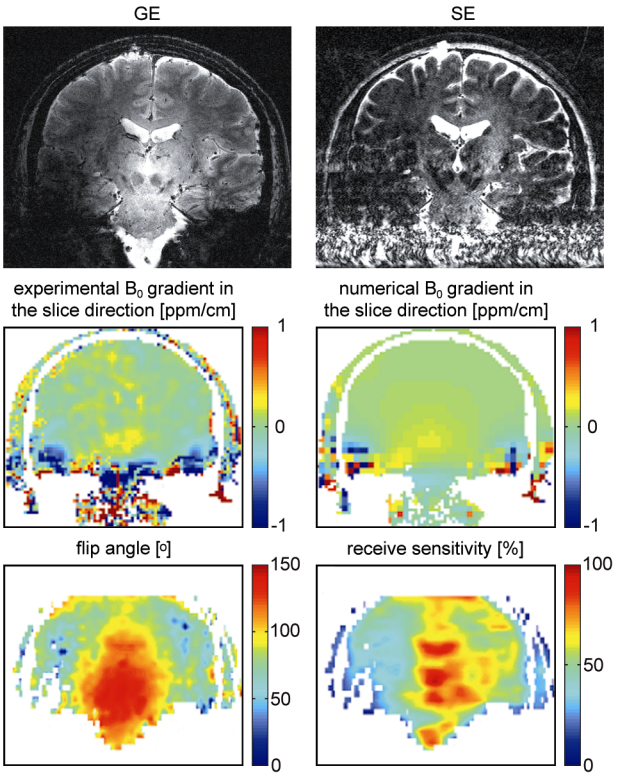

Figure 1: Results of an *in vivo* study (53-year-old male).