Validation of Susceptibility-Based Models with Field Map Measurements in the Breast

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Introduction: MRI is important for detecting, diagnosing and staging breast cancer. The shape of the breast may result in subject-specific susceptibility-induced field changes, which can generate image artifacts such as blurring, signal loss, or unreliable fat suppression, and may limit the use of pulse sequences that are sensitive to field inhomogeneities, such as EPI, bSSFP, and spiral imaging. Shimming can be used to reduce field inhomogeneities; however, it is difficult to accurately shim the breast to obtain a uniform field within the breast tissue, and improper shimming has clinical implications, requiring repeat scans. In this work, we investigated a model-based B_0 field map in the breast based on magnetic susceptibility. The goal of this work is to determine quantitatively how well the field inhomogeneities can be modeled based on the breast shape in 3D datasets [1].

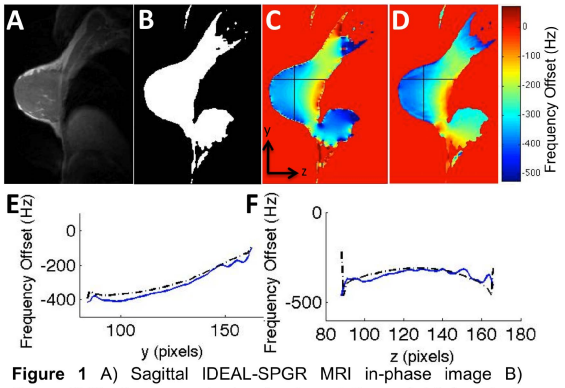

Figure 1 A) Sagittal IDEAL-SPGR MRI in-phase image B) Susceptibility tissue mask showing breast tissue (white) and air (black) C) Computed field map D) Experimental measured field map E) Horizontal (y) field trace F) Vertical (z) field trace showing good agreement between the modeled and measured field maps.

Methods: ***Acquisition*:** We obtained water, fat, in-phase, out-of-phase, and field map images using an investigational version of IDEAL-SPGR [2]. The shims that compensate for the background magnetic field were initially measured in a spherical ball phantom. For the *in vivo* scans, the additional compensatory linear shims were subtracted from the measured field map, while leaving the background shims on. Therefore, any variations were only due to the subject and not the background main B_0 magnetic field. ***Modeling*:** We filtered and then thresholded the in-phase image (Fig. 1A) in order to calculate the tissue susceptibility mask $\chi(r)$ (Fig. 1B). We then used MATLAB74 (Natick, MA) to calculate the distribution of the magnetic field perturbation $\Delta B_0(r)$,

$$\Delta B_0(r) = FFT^{-1}\left[B_0\left(\frac{1}{3}-\frac{k_z^2}{|k|^2}\right)FFT[\chi(r)]\right],$$

where B_0 is the main static magnetic field, k is the Fourier space coordinate, and $\chi(r)$ is the 3D magnetic susceptibility distribution [3,4]. A global B_0 frequency offset was experimentally determined and subtracted from each computed field map (Fig. 1C). ***Experiments***: We scanned 10 healthy female volunteers on a GE 3T MR750 scanner to obtain the IDEAL-SPGR measured field map (Fig. 1D), and then modeled the B_0 field for each dataset based on the in-phase image. ***Analysis***: The area of analysis was restricted to breast tissue by manual segmentation [5]. The computed field map, $\Delta B_0(r)$, was quantitatively compared with the IDEAL-SPGR measured field map by subtraction to create a difference field map and a corresponding histogram for the entire 3D volume for each of the field maps of each volunteer (Fig. 2). We calculated the absolute mean error and the full width at half maximum (FWHM) values of the histograms of the frequency values inside the tissue of the computed, measured and difference field maps for all 10 cases (Fig. 3).

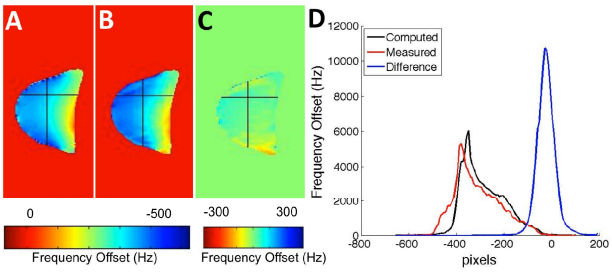

Figure 2 A) Computed field map B) Experimental measured field map C) Difference field map D) Histogram of ΔB_0 values within the restricted tissue mask region of the measured, computed and difference field maps. The absolute error is 36 Hz, and the FWHM of the computed, measured and difference field maps are 181 Hz, 194 Hz and 68 Hz.

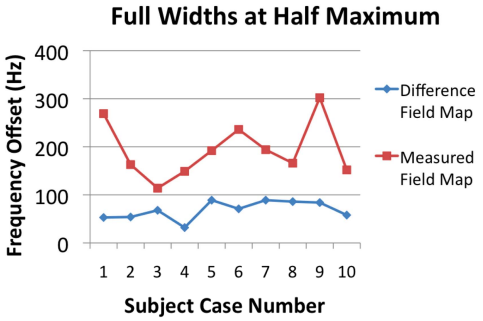

Figure 3 Full widths at half maximum of the histograms of the 10 *in vivo* difference and measured field maps.

Results and Discussion: An example *in vivo* subject is shown in Figures 1 and 2, and the FWHMs of the histograms for all 10 subjects are shown in Figure 3. The absolute error averaged across the 10 subjects was 36 Hz. The average FWHM of the histograms of the 10 computed, measured, and difference field maps were 181 Hz, 194 Hz, and 68 Hz, respectively. **Conclusion:** We have shown that we can model and predict B_0 inhomogeneities in the breast based on the subject shape, using a susceptibility model. The 10 *in vivo* volunteer scans show visual agreement in the field maps, and quantitative agreement of the average FWHM values and the average absolute error. These subject-specific estimates of field inhomogeneity provide an accurate predictive model for field maps in the breast and may be used as an initial field map estimate to improve shimming.

Acknowledgements: NIH R01 EB009055-01, NIH P41RR009784, Lucas Foundation, NSF GRFP, GE Healthcare. **References: 1.** Jordan et al. ISMRM 2009 p. 2125. **2.** SB Reeder et al, Magn. Reson. Med, 51, 35–45 (2004). **3.** Koch et al. ISMRM 2008 p. 1180. **4.** KM Koch et al, Phys. Med. Biol, 52, 6381-6402 (2006). **5.** Maril N et al. Magn Reson Med 2005;54:1139–1145.