Frequency Mapping in the Spinal Cord with WASSR at 3 Tesla

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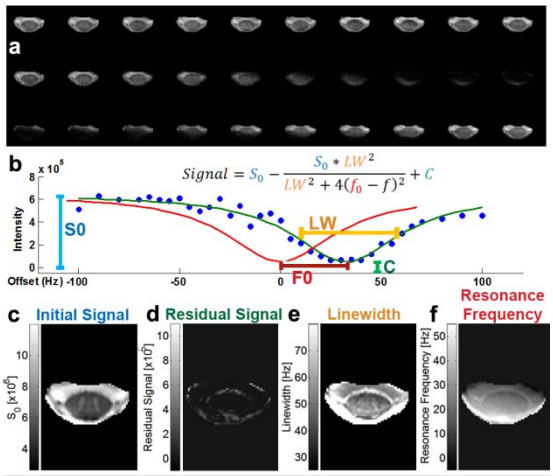

Figure 1: (a) The direct saturation signal in each voxel [●] is (b) fit to a Lorentzian lineshape [\], then shifted [\] to the scanner reference frequency, resulting in maps of the (c) initial signal S_0, (d) residual signal C, (e) linewidth at full width half maximum LW, and (f) resonance frequency F_0.

Introduction: Quantitative Susceptibility Mapping (QSM) methods have shown correlations between magnetic susceptibility and iron or myelin content in the central nervous system.[1,2] The main experimental aspect of quantifying susceptibility is determining the resonance frequency in each voxel, which is affected by the main field, background field gradient inhomogeneities, and the magnetic susceptibility of tissues in and around each voxel. Traditionally, QSM methods generate frequency maps using phase maps from gradient-recalled echo (GRE) imaging.[3-5] Contrast in these phase images evolves with echo time (TE); images at longer TEs result in more phase contrast, but unfortunately also have less signal and more phase wrap discontinuities at interfaces with large susceptibility differences.

Susceptibility would be useful for assessing structural integrity in the spinal cord, which plays a role in many neurological disorders. For example, spinal cord lesions have been shown to correlate with severity of sensorimotor dysfunction in multiple sclerosis.[6] Susceptibility weighted imaging, which qualitatively combines contrast from the magnitude and phase signal[7], has also been used to detect hemorrhage in acute cervical spinal cord injury[8]. However, using GRE and phase for quantitative susceptibility imaging is difficult in the spinal cord, a small structure surrounded by tissues with large susceptibility differences, where imaging is limited by signal drop-off and dephasing effects. Instead, we propose measuring frequency with the WAter Saturation Shift Referencing (WASSR) method, in which the sample is directly saturated with a radiofrequency (RF) pulse across a range of offset frequencies.[9-11] By maintaining a low RF pulse power and short duration, this saturation causes the frequency dependence of the signal in each voxel to follow a Lorentzian absorption lineshape that is not affected by inhomogeneous line broadening.[12,13] The minimum signal occurs at the resonance frequency in each voxel.

Subjects and Methods: For this initial study, one healthy 29-year-old female was studied after IRB approval and written informed consent, using a 3T Philips system with body-coil excitation and 16-channel head-and-neck coil receive. WASSR images were acquired with a 3D gradient-echo multi-shot EPI readout (EPI factor = 25, TR/TE/α = 160ms/29ms/20°, Volume Acquisition Time = 28s, Scan Duration = 17:15min) with a direct saturation sinc-gauss prepulse (B_1 = 0.2μT, t_{sat} = 70ms) applied at 35 offset frequencies between +/-100Hz, as well as two volumes acquired with no applied saturation. Acquired resolution was 0.55x0.60x4mm^3, reconstructed to 0.4x0.4x4mm^3, covering 20 slices of cervical spinal cord. Using MATLAB, the magnitude signal at each offset frequency was fitted per voxel to a Lorentzian lineshape, resulting in maps of the resonance frequency (f_0), the initial signal (S_0), baseline residual signal (C), and linewidth (LW). After masking the spinal cord, the slowly-varying background gradient inhomogeneity in each slice was modeled using a fourth-order polynomial. Subtracting the background gradient inhomogeneity estimation from the raw resonance frequency maps produced images of the residual frequency.

Results and Discussion: Figure 1 illustrates how WASSR images are processed by fitting the signal in each voxel to a Lorentzian equation. Figure 2 shows line profiles through the raw frequency map, the fitted background gradient inhomogeneities, and the residual frequency for selected spinal cord slices. GRE is a convenient and widely available method, but may not be optimal for characterizing resonance frequency in the spinal cord, due to long echo times necessary to provide phase contrast that result in lower SNR, long scan times that are prone to motion artifacts, and phase wraps at interfaces of varying magnetic susceptibility, which complicate data interpretation. The WASSR method uses the magnitude signal (with no phase wraps), does not rely on contrast evolution over echo time (accommodating a shorter TE for more signal), and EPI acquisition (allowing ~20-30sec per volume). Using a high EPI factor may introduce geometric distortions, but allows for fast imaging of multiple volumes. The scan duration for WASSR depends on the number of volumes acquired; generally, more volumes will produce more points to fit along a Lorentzian lineshape, resulting in a better fit. The WASSR volumes should be coregistered; however, if the subject moves while one volume is acquired at a particular RF offset frequency, that volume may be discarded without drastically affecting the Lorentzian fit.

Conclusion: This initial study shows the feasibility of using WASSR to create resonance frequency maps in the spinal cord, which has areas of low signal, large differences in susceptibility, and much motion. WASSR should be especially advantageous for low-field imaging where a long TE is required for GRE phase contrast and SNR is lower, with subjects that are likely to move (e.g., children), or across areas with large susceptibility differences that would produce many phase wraps (e.g., spinal cord). Resonance frequency maps are needed to calculate of quantitative susceptibility maps, which may help to assess structural integrity of the spinal cord in spinal cord injury or demyelinating diseases.

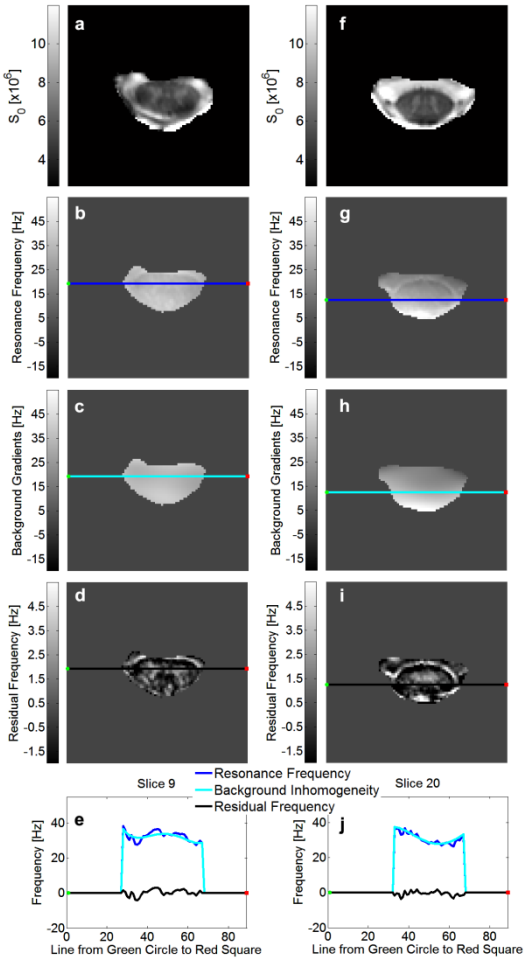

Figure 2: Line profiles (e,j) through lower (a-e) and upper (f-j) slices in the cervical spinal cord showing residual frequency maps (d,i) created by subtracting background inhomogeneities modeled with a 4th order 2D polynomial (c,h) from resonance frequency maps created with WASSR (b,g). S_0 maps (a,f) are shown for comparison.

References: [1] Schweser, et al. NeuroImage 2011, 54:2789. [2] Liu, et al. NeuroImage 2011, 56:930. [3] Shmueli, et al. MRM 2009, 62:1510. [4] Wang, et al. IEEE Eng Med Bio 2009, 2009:53. [5] Wharton, et al. NeuroImage 2010, 53:515. [6] Zackowski, et al. Brain 2009, 132:1200. [7] Haacke, et al. MRI 2004, 52:612. [8] Wang, et al. MRI 2011, 29:365. [9] Kim, et al. MRM 2009, 61:1441. [10] Smith, et al. MRM 2009, 62:384. [11] Lim, et al. ISMRM 2010, #4531. [12] Mulkern et al. Med Phys 1993 20:5. [13] Liu, et al. CMMI 2010 5:162. **Funding**: NIH-P41RR015241