Quantitative Characterization of Spatial Variations of Intrinsic Susceptibility by $T_{1\rho}$ Dispersion

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Target Audience: Investigators interested in novel image contrast mechanisms and quantitative characterization of magnetically heterogeneous tissues.

Purpose: A new method is described for characterizing magnetically inhomogeneous media and for quantifying the spatial scales of susceptibility variations within samples. The approach is based on a theoretical analysis of the effects of diffusion through periodic variations of magnetic susceptibility on the behavior of rates of spin-lattice relaxation in the rotating frame, $R_{1\rho}$ ($=1/T_{1\rho}$). Magnetic susceptibility variations induce local field gradients that scale with field strength, and diffusion through such gradients is well known to increase transverse relaxation rates as spins irreversibly dephase [1]. $R_{1\rho}$ is also affected by exchange and diffusion effects to a degree that depends on the magnitude of the locking field used for measurements [2,3]. An appropriate analysis of the behavior of $R_{1\rho}$ with different locking fields may thus be used to spatially characterize susceptibility variations in inhomogeneous tissues.

Methods: We recently derived an expression for quantifying the influence of diffusion on $R_{1\rho}$ in the presence of a sinusoidally varying local field $b(x) = A\sin(qx)$, where q is a spatial frequency, and a mean gradient g, where $g^2 = \frac{A^2q^2}{2}$. This contribution during a spin-locking experiment can be shown to be $R_{1\rho}^{diff} = \frac{\gamma^2 g^2 D}{(q^2D)^2+\omega_1^2}$, where γ is the gyromagnetic ratio, ω_1 is the spin-lock amplitude, and D is the self-diffusion coefficient. The decrease of $R_{1\rho}$ with increasing locking field, or $R_{1\rho}$ dispersion, provides information about the spatial scales and magnitudes of the gradient pattern, and will exhibit an inflection point at a critical locking field $\omega_1 = \frac{Dq^2}{\sqrt{3}}$. Although pure sinusoidal field variations do not occur naturally, the above expression provides useful insights as an approximate theoretical description of more realistic arrays of tissue structures. To demonstrate this, a series of samples consisting of closely packed polystyrene spheres (Polysciences Inc., PA, USA) in water with diameters ranging from 4.5–90 μm were prepared. Measurements of $R_{1\rho}$ were performed at 7 T (Varian Inc. Palo Alto, CA, USA) using a standard spin-locking sequence in which the values of the spin-lock amplitude varied across the clinically relevant range from 0 to ~1,000 Hz. The samples were subsequently imaged (128x128 matrix, 20mm x 20mm FOV) using a spin-lock prepped Fast Spin Echo acquisition with a fixed locking time (150 msec) and three different spin-lock amplitudes (0 Hz, 22 Hz, 1000 Hz) spanning an appropriate range that was derived by inspection of the collected dispersion curves. Three images acquired with different values of ω_1 can be combined on a voxel by voxel basis as $\left[\frac{\log(I_{high})-\log(I_{low})}{\log(I_{high})-\log(I_{mid})} - 1\right] = \left(\frac{\omega_{1mid}}{Dq^2}\right)^2$, where I_{low}, I_{mid}, I_{high} are image intensities at the lowest, middle, and highest spin-lock amplitudes respectively. This combination produces parametric images reflecting a direct measure of the spatial scale of inhomogeneity within the sample.

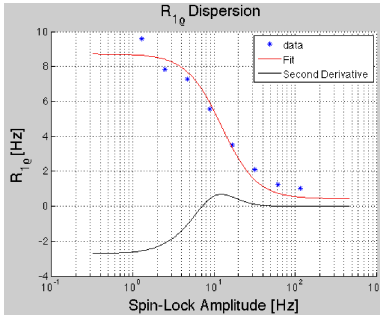

Figure 1: $R_{1\rho}$ dispersion curve (red) for 45 μm polystyrene microspheres in water and the corresponding second derivative (black).

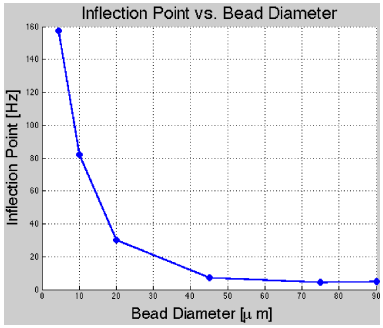

Figure 2: Inflection point of $R_{1\rho}$ dispersion curves as a function of bead diameter for all six samples.

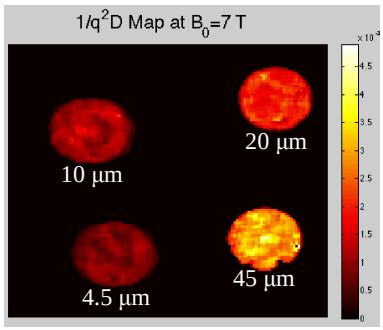

Figure 3: $1/q^2D$ map for 4.5, 10, 20, and 45 μm diameter polystyrene sphere samples. Pixel intensity represents $1/q^2D$ in seconds.

Results: Significant dispersion of $R_{1\rho}$ with increasing locking field was apparent for all samples and figure 1 depicts the dispersion curve for one such sample (red) along with a plot of the second derivative of the fitted curve (black). The resultant curves were analyzed to derive the inflection frequencies for each sample. It was noteworthy that the amplitude of the dispersion increased with increasing bead size due to stronger field gradients. Figure 2 shows how the inflection frequency decreased with increasing bead size, indicating more power was needed to overcome the internal gradients. Figure 3 shows an image that represents the above quantity and depicts the average spatial scale of the inhomogeneities.

Discussion: Spin-echo and gradient-echo measurements have previously been shown to characterize susceptibility variations within tissues, and appropriate comparisons between these have been used to infer spatial scales e.g. vessel size imaging [4]. Here we show an alternative approach using a spin-locking imaging method to derive parametric images characterizing the dimensions of inhomogeneities that, at high field, cause dephasing via diffusion. Note that for scales of practical interest these dispersion effects occur at much lower frequencies than the range in which chemical exchange effects between labile protons and water cause dispersion in $R_{1\rho}$ with locking field, allowing for separate analysis of each relaxation mechanism [5]. Further imaging studies examining the behavior of different media and the influence of microvasculature will be of interest.

Conclusion: Using $R_{1\rho}$ dispersion curves measured with relatively low spin-lock amplitudes and a novel image subtraction technique, parametric images reflecting the spatial scale of inhomogeneities that cause susceptibility gradients can be constructed. This technique has the potential to quantitatively characterize magnetically inhomogenous tissues *in vivo*.

References: [1]Kennan R. et al. Magn Reson Med 22:197-203, 1991. [2]Deverell C. et al. Mol Phys 18:553-559, 1970. [3]Cobb J. et al, Magn Reson Med 66:1563-1571, 2011. [4]Boxerman J. et al. Magn Reson Med 34:555-566, 1995. [5]Cobb J. et al. Magn Reson Med 2012 Jul 12 (in press).