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Probing of susceptibility structures through the distant dipolar field effect

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ABSTRACT

In this paper, the utilization of the distant dipolar field (DDF) signal to extract the properties of susceptibility structures over a subvoxel length scale is investigated. Numerical simulations are performed to study a system of randomly distributed blood vessels with a susceptibility offset inside a voxel. It is shown that the DDF signal of the system as a function of the strength of the correlation gradient field manifests a peak that depends on the volume ratio, size, and susceptibility offset of the blood vessels. In particular, the location of the signal peak is found to vary as powers of these parameters. As a result, by varying the strength of the correlation gradient field, the characteristic properties of the blood vessels can be extracted from the peak position of the DDF signal. It is also found that, for a given volume ratio of the blood vessels, a smaller size of the blood vessels can be probed when the susceptibility offset is increased. Nevertheless, it is demonstrated that, owing to the broad width of the signal peak, the DDF effect generally cannot be used for the preferential selection of the signal arising from the blood vessels on the length scale determined by the correlation length.
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1. Introduction

Over the past two decades, there has been an increasing interest in studying the magnetic resonance (MR) signals formed under the influence of distant dipolar field (DDF) [1–4]. DDF is originated from the intermolecular dipolar interactions between distant spins. The DDF effect is also known as the intermolecular multiple-quantum coherences (iMQCs) when interpreted using the quantum coherence picture [5]. It has been shown that the DDF signal exhibits unique signal features in liquid nuclear magnetic resonance that lead to important applications in magnetic resonance spectroscopy (MRS) and imaging (MRI). For example, the DDF signal is found to yield sharp spectral lines in inhomogeneous magnetic field [3,6], to manifest image contrast enhancement based on the changes in transverse relaxation [4,7–11], and to have a reduced sensitivity to macroscopic motion in diffusion weighted imaging [12,13]. It also enables the determination of equilibrium magnetization [14], simultaneous measurements of diffusion coefficient and transverse relaxation time [15], and accurate imaging of temperature in vivo [16].

A prominent feature of the DDF signal is that it is contributed predominantly from the surrounding spins within a correlation length [2,17], which is defined as half of the spatial modulation period induced by the applied field gradient. The specific sensitivity of the DDF signal to the correlation gradient field thus can potentially be utilized as a new contrast mechanism to assess sample structures on a subvoxel length scale. Studies were carried out to examine the DDF signal of samples with non-uniform spin density distribution [17–23] and electronic susceptibility-induced field distribution [24–28] for different strengths and directions of the applied correlation gradient field. For the studies with local susceptibility-induced field variation, the DDF signal is found to provide information about structural anisotropy by varying the direction of the gradient field [24,27]. The signal is also demonstrated to reflect the characteristic length scale of the local susceptibility-induced field distribution when the correlation length is varied [28].

For applications in functional MRI (fMRI), the DDF is found to provide an enhanced blood oxygen level-dependent (BOLD) signal contrast based on the change in transverse relaxation [7–9,29]. Because of the specific sensitivity of the DDF signal to the correlation field gradient, it had been thought that it is possible to select the signal that is contributed from the local structures on a length scale characterized by the correlation length. Such a preferential signal selection would be valuable in fMRI as it allows an enhanced signal contrast originated from the hemodynamic changes in the capillaries and hence localizes the site of neuronal activity when the correlation length is adjusted to the size of the capillaries. However, although the signal was shown to reflect the characteristic length scale of the local field variation, no affirmative result upon the preferential selection of the signal has been obtained thus far [26,28,30]. The underlying reason for the negative result is still unclear in view of the

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signal dependence on the correlation length. This motivates a more elaborate study of the DDF signal as a function of the correlation length to evaluate the possibility of the preferential signal selection and to examine how the properties of the susceptibility structures can be extracted from the signal features.

In this paper, the properties of the DDF signal in the presence of a local magnetic field induced by subvoxel structures are examined in an extensive range of the correlation length. Parameters that simulate the magnetic field induced by blood vessels in brain are considered. Specifically, the spin density distribution is taken as uniform and the blood vessels are modeled by infinitely long cylinders. The possibility of the preferential signal selection will be addressed.

2. Theory

The evolution of the magnetization $\left(\vec{M}\left(\vec{r},t\right)\right)$ of a spin-1/2 system with free diffusion is governed by the Bloch–Torrey equation. In the rotating frame at the Larmor frequency, the Bloch–Torrey equation during the free evolution period of the magnetization is written as

$$\frac{\partial \vec{M}\left(\vec{r},t\right)}{\partial t} = \vec{M}\left(\vec{r},t\right) \times \Delta\omega\left(\vec{r}\right)\hat{z} \tag{1a}$$

$$-\frac{M_x\left(\vec{r},t\right)\hat{x} + M_y\left(\vec{r},t\right)\hat{y}}{T_2} - \frac{M_0\left(\vec{r}\right) - M_z\left(\vec{r},t\right)}{T_1}\hat{z}$$

$$+\gamma\, \vec{M}\left(\vec{r},t\right) \times \vec{B}_d\left(\vec{r},t\right) + D\nabla^2\, \vec{M}\left(\vec{r},t\right). \tag{1b}$$

Here γ is the gyromagnetic ratio, T_1 and T_2 are respectively the longitudinal and transverse relaxation times, M_0 is the equilibrium magnetization, $\Delta\omega$ is the susceptibility-induced frequency offset, $\vec{B}_d$ is the distant dipolar field, and D is the free diffusion coefficient. The dipolar field, which is non-local in r-space, is written as

$$\vec{B}_d\left(\vec{r},t\right) = \frac{\mu_0}{4\pi}\int d^3\vec{r}'\, T\left(\vec{r}-\vec{r}'\right)\left[3\left(\vec{M}\left(\vec{r}',t\right)\cdot\hat{z}\right)\hat{z} - \vec{M}\left(\vec{r}',t\right)\right], \tag{2}$$

where

$$T\left(\vec{u}\right) = \frac{3(\hat{u}\cdot\hat{z})^2 - 1}{2u^3} \tag{3}$$

is the dipolar field factor. The calculation of the non-local spatial integral of the dipolar field is very computational intensive. A more efficient approach to obtain the DDF signal is to solve Eq. (1) numerically with the dipolar field and the diffusion term in Eq. (1b) evaluated in k-space [31].

In this paper, the DDF signal of an image voxel with a local magnetic field induced by subvoxel susceptibility structures is studied. The voxel is modeled by a lattice with dimensions $l_x \times l_y \times l_z$, which constitutes the entire simulation space. With the k-space approach, periodic boundary condition is imposed implicitly to the system. Under such condition, identical copies of the voxel compose an infinite sample medium, which in practice can be viewed as a huge cubic sample with the voxel located at the center. Then the $\vec{k}=\vec{0}$ component of the DDF vanishes, and the Fourier transform of the dipolar field is given by (see Appendix A for details)

$$\tilde{\vec{B}}_d\left(\vec{k},t\right) = \begin{cases} -\frac{\mu_0}{3}\frac{3\left(\hat{k}\cdot\hat{z}\right)^2 - 1}{2}\left[3\tilde{M}_z\left(\vec{k},t\right)\hat{z} - \tilde{\vec{M}}\left(\vec{k},t\right)\right], & \text{for } \vec{k} \neq \vec{0}, \\ 0, & \text{for } \vec{k} = \vec{0}, \end{cases} \tag{4}$$

where μ_0 is the permeability of vacuum, and

$$\tilde{\vec{M}}\left(\vec{k},t\right) = \frac{1}{(2\pi)^3}\int d^3\vec{r}\, e^{-i\vec{k}\cdot\vec{r}}\, \vec{M}\left(\vec{r},t\right) \tag{5}$$

is the Fourier transform of the magnetization. In fact, Eq. (4) is valid for an image voxel that is located at the center of a cubic sample, or inside a spherical sample, when its size is much smaller than the sample size. The DDF signal with the dipolar field evaluated in k-space will be verified numerically against the signal with the DDF calculated in r-space in Section 3.1.

The correlation spectroscopy revamped by asymmetric z-gradient echo detection (CRAZED) sequence (Fig. 1) is the basic pulse sequence that forms the DDF signal and is used in this paper. In the sequence, the strength and duration of the field gradients are G_i and δ_i, where $i = 1,2$. The rf pulses and the field gradients are assumed to be instantaneously short (hard pulse approximation). The modulation wave vectors induced by the field gradients are taken along the z-direction as $k_{G_1}\hat{z}$ and $k_{G_2}\hat{z}$, where $k_{G_i} = \gamma G_i \delta_i$.

Local magnetic field induced by blood vessels in brain is considered in this paper, with the spin density distribution taken as uniform. A common way to model the susceptibility-induced field of a blood vessel is to use an infinitely long cylinder with a finite susceptibility offset from the background [32,33]. In order to control the characteristic lengths of the induced field variation with respect to the direction of the applied gradient field, the blood vessels are first taken to be lying along the same direction and are randomly placed perpendicular to the applied field gradient on the x-y plane. This configuration maximizes the induced field variation and helps to increase the computational efficiency when a wide range of the correlation length is studied. For simplicity, the blood vessels are taken to be lying along the x-direction. At the end of Section 3, a general three-dimensional system with randomly distributed vessels will be studied to compare with the results of the system with parallel vessels.

The susceptibility-induced frequency $\left(\Delta\omega\left(\vec{r}\right)\right)$ for a given susceptibility offset distribution $\left(\Delta\chi\left(\vec{r}\right)\right)$ can be found by [34]

$$\Delta\omega\left(\vec{r}\right) = -\frac{\gamma B_0}{3}\int d^3\vec{k}\, e^{i\vec{k}\cdot\vec{r}}\left[3\left(\hat{k}\cdot\hat{z}\right)^2 - 1\right]\Delta\tilde{\chi}\left(\vec{k}\right), \tag{6}$$

where B_0 is the static field strength, $\Delta\tilde{\chi}$ is the Fourier transform of $\Delta\chi$, and the $\vec{k}=\vec{0}$ component of Eq. (6) is taken as zero (c.f. the derivation of $\tilde{\vec{B}}\left(\vec{k}=\vec{0}\right)$ in Appendix A). For a cylinder (labeled with index i) lying along the x-direction with radius a_{0i} and axis located at (y_{0i}, z_{0i}), the susceptibility-induced frequency distribution is given by [32]

$$\Delta\omega_i(y_i, z_i) = \begin{cases} -\frac{\Delta\chi_0}{6}\gamma B_0, & \text{for } y_i^2 + z_i^2 \le a_{0i}^2, \\ \frac{\Delta\chi_0}{2}\frac{a_{0i}^2\cos 2\psi_i}{y_i^2 + z_i^2}\gamma B_0, & \text{for } y_i^2 + z_i^2 > a_{0i}^2, \end{cases} \tag{7}$$

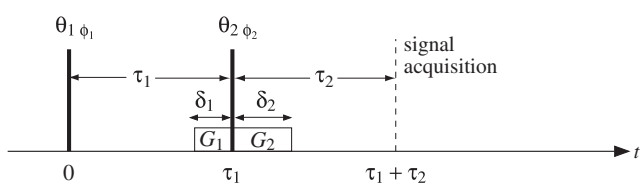

Fig. 1. The correlation spectroscopy by asymmetric z-gradient echo detection (CRAZED) sequence. Two rf pulses with flip angles $\{\theta_1, \theta_2\}$ and phases $\{\phi_1, \phi_2\}$ are applied respectively at times $\{0, \tau_1\}$. The ratio of the modulation wave numbers is taken to be $k_{G_2}/k_{G_1} = G_2\delta_2/G_1\delta_1 = 2$.

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