Enhanced constrast of signal from distant dipolar field on relaxation times and B_0 in tissues

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INTRODUCTION

In the past few years distant dipolar field (DDF) and iMQC due to intermolecular spin interactions has been explored as a possible new source of contrast for MR imaging [1-3]. In this abstract, the Bloch equation including the dipolar field and relaxation effects for a typical CRAZED pulse sequence [4] is solved analytically for arbitrary flip angles θ_1 and θ_2. We analyze the dependence of the signals on the field strength B_0, echo time τ_2, and relaxation times T_1 and T_2. Experimental data are compared with the theory. The results show that signals due to DDF ($|n| \neq 1$) rise faster at large B_0. We also estimate the signal changes in tumor and normal muscle tissue in mice. The results show the signals due to DDF are significantly more sensitive to the change of T_1 and especially T_2 than the conventional signal ($|n| = 1$). These analyses have potential applications to tumor studies.

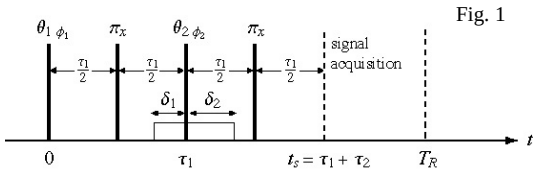

Fig. 1

THEORY

We consider the CRAZED sequence in Fig. 1 acting on a homogeneous spin-1/2 system. Two rf pulses are applied at $t = 0$ and $t = \tau_1$, and two linear gradient fields, $\boldsymbol{G}_1$ and $\boldsymbol{G}_2$, are applied around the second rf pulse with durations δ_1 and δ_2 ($\boldsymbol{G}_2\delta_2 = n\boldsymbol{G}_1\delta_1$). The π pulses are used to reduce inhomogeneous effects. Here we ignore both radiation damping and molecular diffusion. The acquired transverse magnetization in the steady state is $M_+(\boldsymbol{r}, t_s) = \sum_{(m=0 \text{ to } \infty)} M_+^{(m)}(\boldsymbol{r}, t_s)$, with each $M_+^{(m)}(\boldsymbol{r}, t_s)$ consisting of a phase $\exp[i(\boldsymbol{G}_2\delta_2 - m\boldsymbol{G}_1\delta_1)\cdot\boldsymbol{r}]$. Since $M_+^{(m)}(\boldsymbol{r}, t_s)$ is averaged out when $\boldsymbol{G}_2\delta_2 \neq m\boldsymbol{G}_1\delta_1$, only the particular term $M_+^{(m=n)}(\boldsymbol{r}, t_s)$ contributes to the signal. The magnitude of the n^{th} order transverse magnetization $|M_+^{(n)}(\boldsymbol{r}, t_s)|$ and the steady state longitudinal magnetization M_{z0} obtained from the Bloch equation are [5,6]

$$\frac{\left|M_+^{(n)}(\boldsymbol{r},t_s)\right|}{M_0} = \frac{1}{2}\left|\frac{M_{z0}}{M_0}\sin\theta_1\left[1+\cos\theta_2\right]\, e^{-(\tau_1+\tau_2)/T_2} J_{n+1}(x) + \frac{M_{z0}}{M_0}\sin\theta_1\left[1-\cos\theta_2\right]\, e^{-(\tau_1+\tau_2)/T_2} J_{n-1}(x) + 2i\sin\theta_2\left[1-\left(1-\frac{M_{z0}}{M_0}\cos\theta_1\right)e^{-\tau_1/T_1}\right] e^{-\tau_2/T_2} J_n(x)\right|,$$

$$M_{z0} \approx \frac{M_0[1-e^{-T_R/T_1}]}{1-\cos\theta_2\cos\theta_1\, e^{-T_R/T_1} + \sin\theta_2\sin\theta_1\cos(\phi_2-\phi_1+\Delta\omega\,\tau_1+\gamma G_1\cdot r\,\delta_1)\, e^{-\tau_1/T_2}\, e^{-T_R/T_1}}. \quad (1)$$

Here $J_n(x)$ is the n^{th} order Bessel function of the first kind with $x = \gamma\mu_0 M_{z0} T_1 \sin\theta_2 \sin\theta_1 \exp(-\tau_1/T_2)\,[1-\exp(-\tau_2/T_1)]$, M_0 is the equilibrium z-magnetization, and $\{\theta, \phi\}$ are respectively the flip angle and phase of the rf pulses. Note that Eq. (1) differs from the conventional form [7] with the additional last term. It corresponds to residual magnetization when $\theta_1 \neq 90^\circ$ and has significant contribution under certain conditions, which will be discussed elsewhere.

RESULTS AND DISCUSSIONS

To obtain optimal signal $|M_+^{(n)}(\tau_2)|$ in the experiment, we used $\theta_1 = 90^\circ$ and $\theta_2 = 0^\circ$ and 60° respectively for $n = -1$ and -2 [8]. We measured the relaxation times of swine muscle at $B_0 = 1.5$, 9.4 and 14T. They are respectively $T_{1sw} = 0.914$, 1.17 and 1.87s, and $T_{2sw} = 42.1$, 21.0 and 17.5ms. The conventional MR signal $|M_+^{(-1)}(\tau_2)|$ measured decays exponentially as described by Eq. (1). Its maximum always occurs immediately after the second gradient pulse.

The second order magnetization $|M_+^{(-2)}(\tau_2)|$ of swine muscle at the three different B_0 is measured (see Fig. 2). Note that the data are normalized by $|M_+^{(-1)}(\tau_{2,\text{peak}})|$ and are fitted with Eq. (1), where $\tau_{2,\text{peak}}$ is the time at which the peak of $|M_+^{(-2)}(\tau_2)|$ occurs. For swine muscle, we find that $\tau_{2,\text{peak}} = T_1 \ln(1+T_2/T_1) \sim T_2$. Since T_2 generally decreases with B_0, the peak of $|M_+^{(-2)}(\tau_2)|$ occurs at an earlier time as B_0 increases (see data set (a) in the inset). This shows the sensitivity of the peak position of $|M_+^{(-2)}(\tau_2)|$ to B_0. On the other hand, even though T_2 is smaller, the relative signal rises faster with large B_0 (see data set (b) in the inset). This suggests the advantage of high magnetic field.

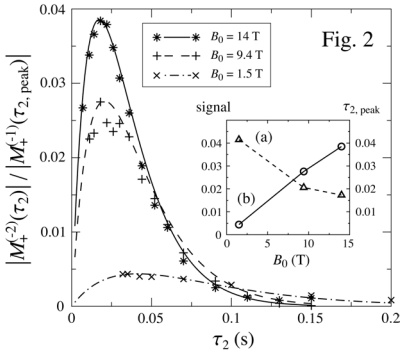

Fig. 2

We also studied the possible signal change in tumor at $B_0 = 9.4$T. Because of the necrosis in the tumor, T_1 and T_2 in tumor are usually longer than that in normal muscle. We have measured the T_2 values for normal cells ($T_{2,\text{normal}}$) and tumor (MCa-4 mammary carcinoma murine) cells ($T_{2,\text{tumor}}$) in mouse legs to be $T_{2,\text{normal}} = 27$ms and $T_{2,\text{tumor}} = 37$ms. Note that the signal does not depend much on T_1 providing that $\tau_{2,\text{peak}}$ is much smaller than T_1 and T_R is sufficiently longer than T_1. We take $T_R = 10$s and estimated T_1 for normal cells ($T_{1,\text{normal}}$) and tumor cells ($T_{1,\text{tumor}}$) to be $T_{1,\text{normal}} = 1.2$s and $T_{1,\text{tumor}} = 1.7$s. The relative signal changes between tumor and normal cells $(|M_+^{(n)}{}_{\text{tumor}}| - |M_+^{(n)}{}_{\text{normal}}|)/|M_+^{(n)}{}_{\text{normal}}|$ for $n = -1$ and $n = -2$ with the above T_1 and T_2 are plotted in Fig. 3. From the figure, we see that $|M_+^{(-2)}|$ has a higher relative change than $|M_+^{(-1)}|$. This indicates that $|M_+^{(-2)}|$ has a higher sensitivity to the change of T_1 and T_2, and may have enhanced tumor detection capability.

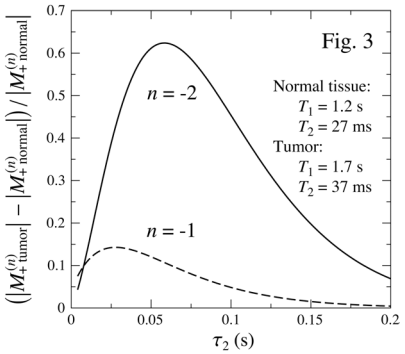

Fig. 3

In conclusion, even though signals from DDF ($|n| \neq 1$) are smaller than the conventional signals ($|n| = 1$), our results show that the relative signals due to DDF are significantly more sensitive to the change in magnetization and relaxation times than the conventional signal. The high sensitivity on the relaxation times provides a good contrast to distinguish tumor cells from normal cells, which has potential applications to tumor studies by providing a new possible source of high contrast MR imaging.

REFERENCES

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