핵자기 뇌기능 영상에서 SSFPI 기법을 이용한 자화율효과의 관찰

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Observation of Susceptibility Change in fMRI Using SSFP Interferometry (SSFPI) Technique

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ABSTRACT

We have developed a fast steady state free precession interferometry (SSFPI) technique which is useful for the fMRI (functional Magnetic Resonance Imaging). As is known, SSFP sequence with a suitable adjustment of gradient (readout) allows us to measure precession angle θ which is in turn related to the field inhomogeneity [1-3]. When the method is applied to the susceptibility effect based functional magnetic resonance imaging (fMRI), it was found that the direct susceptibility effect measurement was possible without perturbations such as the backgrounds and inflow effect. In this paper, simulation results and experimental results obtained with 2.0 Tesla MRI system are also presented.

I. INTRODUCTION

The field inhomogeneities encountered in MRI can be largely categorized by two; one caused by the magnet geometry and built-in inhomogeneity of the magnet and the other is inhomogeneity induced by the chemical shift and local susceptibility effect due to the paramagnetic substances inside the object. The former is measurable and can be corrected. The latter, however, is object dependent and is difficult to be measured.

Recent developments of the dynamic functional studies based on the susceptibility effect appears an ideal application example of the SSFPI techniques since the method offers direct measurements capability of the local susceptibility or map.

In this paper, a modified SSFP sequence namely SSFPI (SSFP Interferometry) is applied to dynamic functional studies in an attempt to quantitatively observe susceptibility effect without interference from such as the inflow effect, and other background signals.

II. THEORY AND METHOD

Steady State Free Precessing Interferometry (SSFPI)

When the time duration of reading gradient (shaded region) is adjusted and made equal, the FID and echo signals will be merged together and form a single nuclear signal which is believed to be an interference signal or an interferogram representing the field inhomogeneity as follows

$$S(g_x,t_y) = S^+(g_x,t_y) + S^-(g_x,t_y) = \iint\left[\left(M_x^+(x,y)+M_x^-(x,y)\right)+j\left(M_y^+(x,y)+M_y^-(x,y)\right)\right] \cdot \exp\left[-j\gamma\left(xg_xT_x + yG_yt_y\right)\right]dxdy \quad [1]$$

where S, S^+, and S^- are illustrated in Fig. 1.

Fourier transformation of Eq. [1] results in an image signal which is given by,

$$M(x,y) = \frac{M_0 \sin\alpha\left(1-E_1\right)\left[4E_2^2\sin^2\theta+\left(1-E_2^2\right)^2\right]^{1/2}}{\left(1-E_1\cos\alpha\right)\left(1-E_2\cos\theta\right)-\left(E_1-\cos\alpha\right)\left(E_2-\cos\theta\right)E_2} \quad [2]$$

where M_0, α, θ, E_1, and E_2 are the equilibrium spin magnetization, flip angle, free precession angle of the transverse spin magnetization during the pulse repetition interval TR, E_1 = exp (- TR/T_1), and E_2 = exp (- TR/T_2), respectively. It is interesting to observe that Eq. [2] is a function of the free precession angle θ, and in turn free precession angle θ is related to the field inhomogeneity provided that suitable values of the flip angle α, E_1 and E_2 are chosen. Since the free precession angle θ is the time integration of the magnetic fields existing in the given imaging situation, including the field inhomogeneity ΔB(x,y), θ can be written as,

$$\theta = \int_0^{TR}\gamma\left[\Delta B(x,y) + xG_x(t) + yG_y(t)\right]dt \quad [3]$$

where TR is the repetition time, $G_x(t)$ and $G_y(t)$ are the x- and y-directional gradient fields, respectively. For the fast SSFP gradient echo sequence shown in Fig. 1, $\int_0^{TR}G_x(t)dt$ and $\int_0^{TR}G_y(t)dt$ terms will cancel out, consequently Eq. [3] can simply be written as,

$$\theta = \int_0^{TR}\gamma\Delta B(x,y)dt = \gamma\Delta B(x,y)TR. \quad [4]$$

Equation [2] and [4] suggest a possible measurement of the field inhomogeneity ΔB(x,y), i.e., Eq. [2] implies that the voxel signal intensity M(x,y) represents an interference pattern of the FID and echo signals hence the precession angle θ, i.e., signal intensity appears as a periodic function of field inhomogeneity.

Susceptibility Effect in Blood

The volume susceptibility of whole deoxygenated blood was known to be 0.08ppm at 2 Tesla [4]. Using oxygenation of the hemoglobin, the magnetic susceptibility

difference in blood can be written as;

$$\Delta\chi_{v,blood}(Y) = (1-Y)\,\Delta\chi^{o}_{v,blood} \quad [5]$$

where Y is the fraction of oxygenated hemoglobin and $\Delta\chi^{o}_{v,blood}(Y)$ is $\Delta\chi^{o}_{v,blood}$ (Y=0). For human, the fraction of oxygenated hemoglobin Y can be calculated from the blood oxygenated curve which is given as [5]

$$\frac{Y}{1-Y} = \left(\frac{pO_2}{P_{50}}\right)^{2.8} \quad [6]$$

where pO_2 is the partial pressure of oxygen in the blood and P_{50} is the partial pressure of oxygen at which half of the hemoglobin sites are bound (P_{50} = 26 torr). Assuming partial oxygen pressure in capillaries varies between 20 and 40 torr where 20 torr corresponds to the normal stage while 40 torr corresponds to the stimulated or oxygen rich state, respectively [6]. Then the susceptibility differences are calculated assuming that the normal susceptibility, χ_{nor}, is about 0.05 ppm (pO_2 = 20 torr) at normal stage while χ_{sti} is about 0.01 ppm (pO_2 = 40 torr) at stimulation stage using Eq. [5] and [6]. Assuming main magnetic field inhomogeneity (ΔB_0) is in the order of 0.04ppm in our imaging of interested region (visual cortex) and the susceptibility χ was calculated varying the flip angle α for the χ between the normal to the stimulation. The normalized magnetization versus flip angle α is calculated using Eq. [2] and plotted in Fig. 2. Since the ΔB_0 is assumed 0.04ppm, the range of total field inhomogeneity variations become roughly between 0.05ppm ($\Delta B_0 + \chi_{sti}$) and 0.09ppm ($\Delta B_0 + \chi_{nor}$) and this range resides well within the linear region for the case of TR/TE=40/20(msec) and T_1/T_2=1000/100(msec) shown in Fig. 3. As seen, in Fig. 2 the sensitivity of susceptibility difference ($\Delta\chi = \chi_{sti} - \chi_{nor}$) is maximum value at near the α=15° ~ 30°.

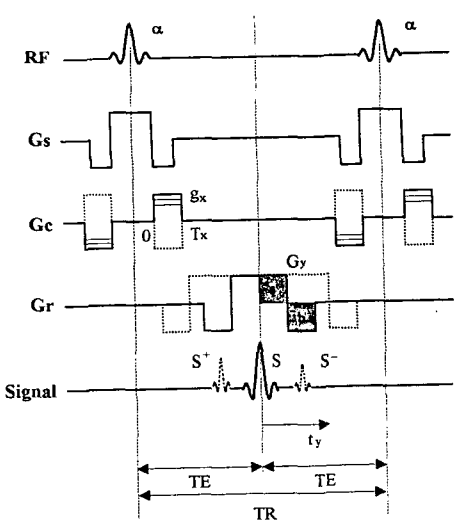


Fig. 1

Fig. 1 Basic SSFPI (Steady State Free Precession Interferometry) sequence. By suitable adjustments of the reading gradient, an interference pattern of the FID and echo signals can be obtained, i.e., adjusting the shaded areas, a and b, to an equal size. Note also that the sum of selection, coding and reading gradients is all becomes zero within a given TR.

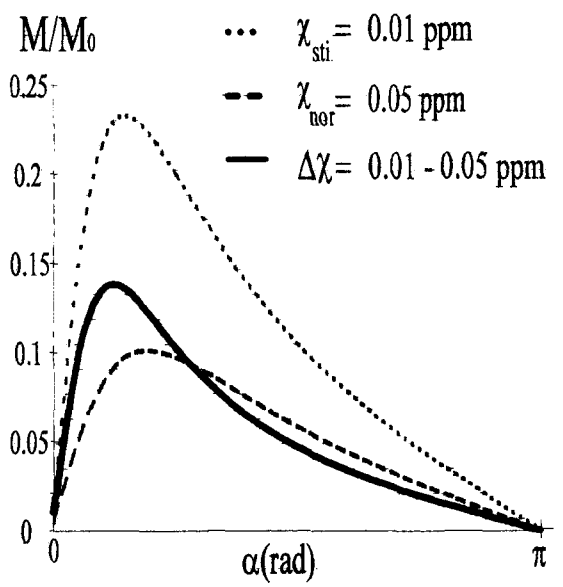


Fig. 2

Fig. 2 Assuming that stimulation stage and normal stage produce susceptibility χ_{sti} =0.01ppm and χ_{nor}=0.05ppm, respectively, normalized voxel signal intensities of each as a function of the flip angle α are plotted at SSFPI sequence together with the differences ($\Delta\chi = \chi_{sti} - \chi_{nor}$).

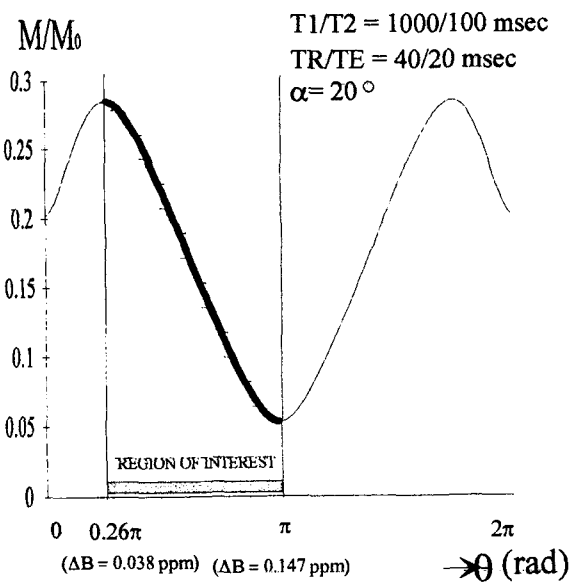


Fig. 3

Fig. 3 A same plot for a fixed α, i.e., α=20° with varying θ (0 to 2π). By adjusting TR, range of the phase θ should be placed in the linear region shown.

Computer Simulation

For 2-D image simulation with SSFPI sequence, we consider a simple model, shown in Fig. 4, in which a 5mm diameter tube filled with agent (such as a group of capillaries) situated in perpendicular to B_0 direction. In Fig. 5(a) SSFPI gradient echo image (coronal view) and its

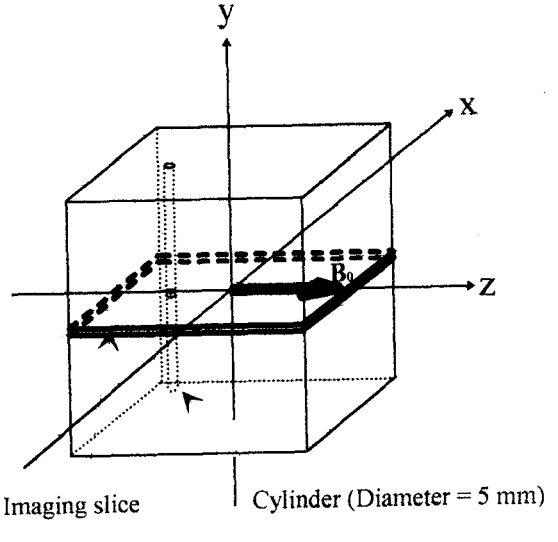
cutview are obtained when χ is varied between 0.01ppm and 0.05ppm (assuming χ_{sti} = 0.01ppm, χ_{nor} = 0.05ppm) inside the tube. It is also assumed that the total main magnetic field inhomogeneity in z-direction and x-direction varied total of 0.147ppm, i.e., θ varied about between 0 and π (1/2 periods) at each direction. The tube is located in the linear region, i.e., the tube is located between θ=0.26π and π. In Fig. 5(a), flip angle α is fixed to 20° . The distortion of field pattern appears in the area that where the tube is located, i.e., where susceptibility change occurred, signal intensity change is observed. By subtraction of χ_{nor} from χ_{sti} tissue oxygenation differences in the capillary can be inferred as shown in Fig. 5(b). For the imaging simulation, first the value of main magnetic field is calculated at each pixel and secondly a new main magnetic field value at each pixel affected by the susceptibility effect is calculated. The new interior magnetic field B_i and the value of external magnetic field $(B_e)_z$ and $(B_e)_x$ in the directions to z and x axes are calculated using the Eq. [7] and [8], respectively, and is given by [7]

$$B_i = B_{oi}\left(1 - \frac{\Delta\chi}{2}\right) \qquad [7]$$

$$(B_e)_z = B_{oi}\left(1 - \frac{\Delta\chi}{2} R^2 \frac{z^2 - x^2}{r^4}\right) \qquad [8a]$$

$$(B_e)_x = -B_{oi} \frac{\Delta\chi}{2} R^2 \frac{2xz}{r^4} \qquad [8b]$$

where B_{oi} is the value of the main magnetic field which we have assumed as total of 0.147ppm in both z- and x-directions as a function of r ($r = \sqrt{x^2 + z^2}$) and $\Delta\chi = \chi_e - \chi_i$. Next, the new field inhomogeneity values (B_i, $(B_e)_z$, and $(B_e)_x$) are calculated using the B_{oi} at each pixel and relative signal intensity (M/M_0) is calculated and then plotted. (See Fig. 5(a) and (b)).



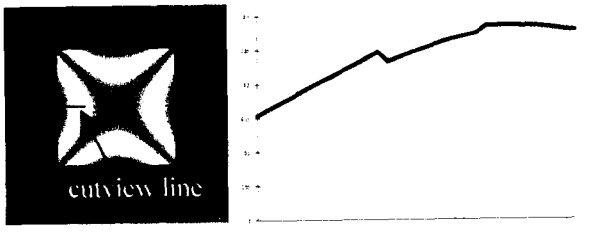
Interior susceptibility of the cylinder : χ_i

Exterior susceptibility of the cylinder : χ_e

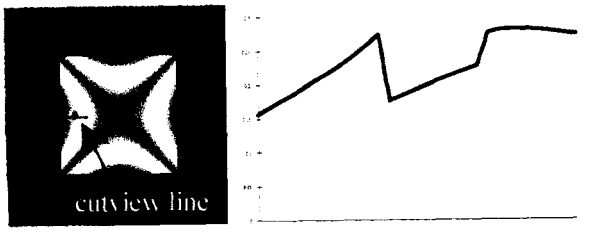
Susceptibility difference : $\Delta\chi = \chi_e - \chi_i$

Fig. 4

Fig. 4 A simple model used for the computer simulation. In this model, a cylindrical tube (representing the veins or a group of capillaries) of diameter D=5(mm) is placed in perpendicular to the B_0 field direction.



0.01ppm



0.05ppm

Fig. 5(a)

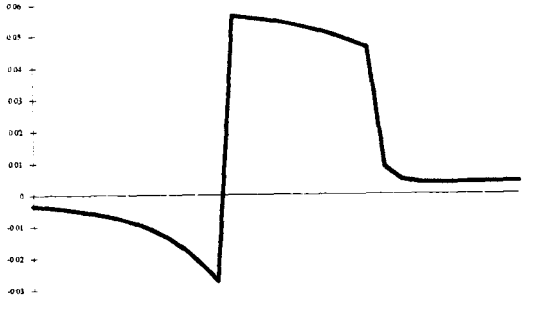


Fig. 5(b)

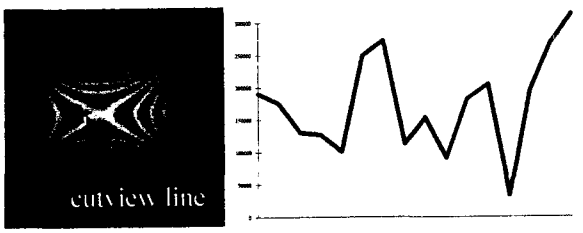
Fig. 5 (a) SSFPI gradient echo simulation results with flip angle α = 20° with TR/TE=40/20(msec) and T_1/T_2=1000/100(msec). (b) differencal cut view data.

III. EXPERIMENTAL RESULTS AND DISCUSSIONS.

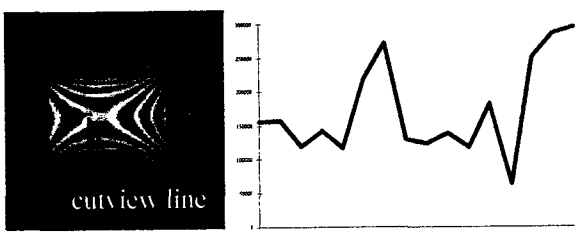
To experimentally demonstrate the validity of the proposed field inhomogenity mapping technique, we have obtained phantom experimental results and displayed in Fig. 6. An experimental phantom consisting of a tube with a different susceptibility value 0.11ppm $CuSO_4$ solution and 0.15ppm $CuSO_4$ solution immersed in water (0.1ppm CuSO4 solution) as shown in Fig. 4, so that susceptibility difference is 0.01ppm and 0.05ppm, respectively. Field inhomogeneity maps of a phantom is obtained using imaging parameters of TR/TE=40/20 msec with α=20° and difference data are also plotted in Fig. 6(b).

In conclusions, we have presented a new method of performing fMRI using a modified SSFP technique which provides field map or field interferogram, i.e., susceptibility field map which is dependent on the status of the

oxygenation in the veins and capillaries. Proposed SSFPI technique, therefore, provides a simple means to susceptibility only fMRI free from the inflow effect which often obscure the real susceptibility measurement in fMRI.



0.01ppm



0.05ppm

Fig. 6(a)

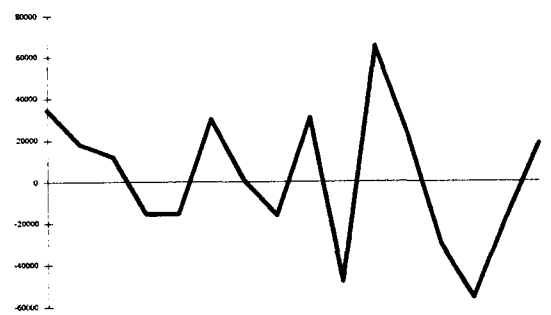


Fig. 6(b)

Fig. 6 An experimental phantom consisting of a tube with a different susceptibility value 0.11ppm $CuSO_4$ solution and 0.15ppm $CuSO_4$ solution immersed in water (0.1ppm CuSO4 solution) as shown in Fig. 4, so that susceptibility difference is 0.01ppm and 0.05ppm, respectively. (a) SSFPI gradient echo experimental results with flip angle α = 20° with TR/TE=40/20(msec). (b) differental cutview data.

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