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인터페로메트리 기법을 이용한 뇌기능 영상

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Brain Functional Imaging Using Interferometry Technique

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INTRODUCTION

A fast SSFP interferometry (SSFPI) sequence is applied to fMRI (functional Magnetic Resonance Imaging). As is known, a suitably adjusted SSFP sequence allows us to measure precession angle with which the field inhomogeneity can be deduced [1-3]. For example, combining the two pulses (known as FID and Echo) in the FADE (Fast Acquisition Double Echoes) sequence, one can produce an interference term which is directly related to the precession angle[4]. It has been known that a high resolution magnetic field mapping is possible by use of the modified FADE sequence which is in effect an interferometry technique. By use of the proposed interferometry technique scan time of less than 10 seconds has been achieved for the inhomogeneity field mapping. The field inhomogeneities encountered in MRI can be largely categorized by two; one caused by the magnet geometry of limited dimension and the other is inhomogeneity induced by inhomogeneous distribution of the susceptibility inside the object to be studied by NMR. The former is object-invariant and its effect to imaging can be predicted if once measured. The effect of the latter, however, cannot be predicted because of the fact that the induced field inhomogeneity is dependent on the spatial distribution of inhomogeneous susceptibilities inside the object [5]. Therefore, object-by-object based field measurement should be performed to predict or correct the effect of field inhomogeneities including susceptibility induced one. Accurately measured field inhomogeneity data can be used for the correction, calibration, and adjustment of a certain experiment such as chemical shift and susceptibility imaging. Further, advanced spectroscopic imaging techniques recently developed require true human *in-vivo* field inhomogeneity maps, i.e., field inhomogeneity maps in the presence of the object which includes susceptibility and chemical shift effects, within a reasonably short imaging time.

When this technique is applied to dynamic functional studies which are based on the oxygenation level dependent susceptibility effect change, it is shown that the method is highly sensitive to the susceptibility effect but relatively insensitive to the inflow effect. As is known, although we have obtained susceptibility affected signal changes with the conventional gradient echo (CGE) technique with small flip angle and long echo time, the results so far obtained are relatively poor in signal to noise ratio and appear to be contaminated by the large fraction of inflow effect [6,7]. It is, therefore, difficult to distinguish, at least in the gradient echo technique, the change of oxygen level alone from the inflow effect in both the visual cortex and small veins. In this paper, a modified SSFP sequence referred to termed as SSFPI (SSFP Interferometry) sequence is proposed and applied to functional imaging in an attempt to quantitatively observe the susceptibility effect without perturbations such as the backgrounds and inflow effect.

THEORY AND METHOD

In Fig. 1, the SSFP fast gradient echo imaging sequence of both multi echo and double echo data acquisitions are shown. When the FID and echo signals obtained simultaneously, it is known as FADE (Fast Acquisition Double Echo). For the sake of simplicity, let us consider FADE sequence only [4]. By adjusting the time duration of the reading gradient in the SSFP sequence, FID and echo signals can be combined so that the result becomes an interference pattern or an interferogram. The magnetization M is then given by (1),

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

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. (1) is a function of the free precession angle θ. Therefore, the field inhomogeneity can be measured provided that a set of suitable parameters are given, namely, the flip angle α, E_1, and E_2. Since the free precession angle θ is the time integration of the magnetic fields existing in the given imaging situation (including the field inhomogeneity $\Delta B(x,y)$), it is given by

$$\theta = \int_0^{TR} \gamma \Delta B(x,y)\,dt = \gamma \Delta B(x,y) TR \quad (2)$$

Since $\int_0^{TR} G_x(t)dt$ and $\int_0^{TR} G_y(t)dt$ terms shown in Fig. 2 will be canceled out, the free precession angle θ will be a sole function of the field inhomogeneity $\Delta B(x,y)$. Then the voxel signal intensity M(x,y) given in Eq. (1) will be a relatively pure function of the field inhomogeneity, which appears as a periodic function of the precession angle θ. The θ dependent signal intensities are plotted in Fig. 3 for the

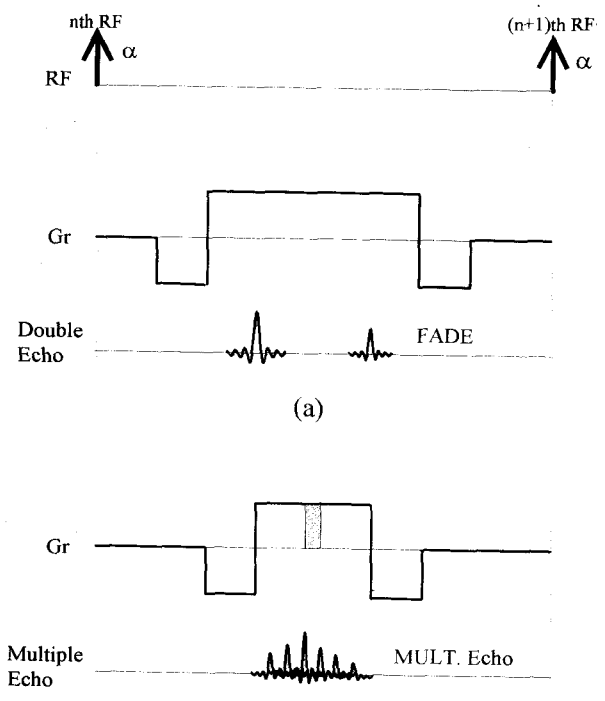

Fig. 1

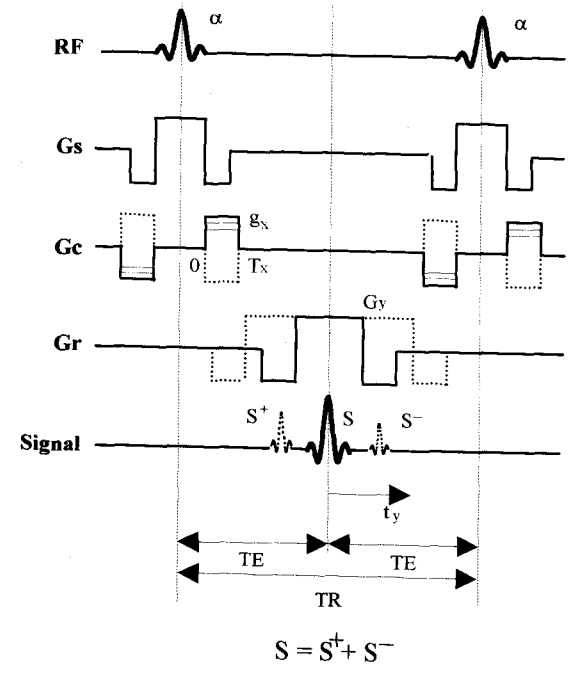

Fig. 2

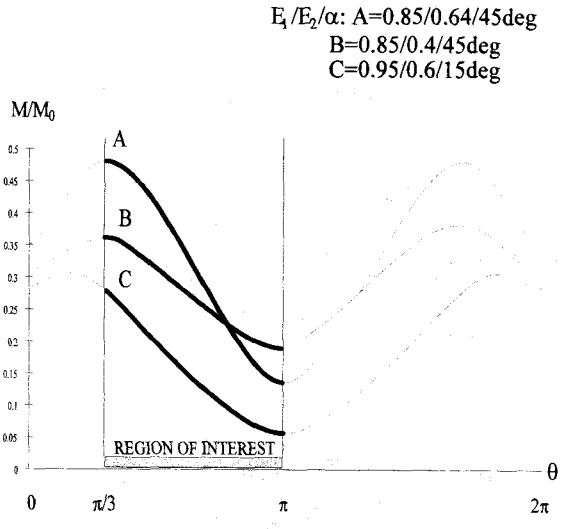

Fig. 3

between $0 \sim 2\pi$. As shown in Fig. 3, signal variations can be calculated for different values of α, E_1, and E_2. Although it shows high degree of amplitude variation for different $E_1/E_2/\alpha$ combinations (3 different combination), all of them show strong sinusoidal variations as a function of θ. By suitable adjustment of the TR a linear relation between strength of the inhomogeneity and signal intensity can be obtained. Therefore, if a field inhomogeneity induced by susceptibility effect lies within the linear region (region of interest) of the curve, the susceptibility effect can be measured directly. Since the method is fast and direct, it can be useful for the measurement of susceptibility effects in fMRI.

EXPERIMENTS AND RESULTS

A series of experiments were performed on a 2.0 Tesla whole body system with a head coil. Functional imaging by photic stimulation was carried out with the SSFPI technique on normal volunteers. For the functional imaging, time course studies with a number of experiments were performed by varying the flip angle α from 20° to 90° with TR/TE =58/30 (msec). A single oblique slice having a slice thickness of 10(mm) was selected with 128X128 image size. The imaging plane which is parallel to the calcarine fissure was carefully selected to provide the time-course data of the signal intensities. TR value was carefully selected so that variation lies in the linear region for SSFPI fMRI. Visual activation was performed by photic stimulation using a home made 8Hz LED checker board. A set of 15 images were obtained with a time interval of 9 sec. The first 5 and last 5 images were obtained during rest period and images 6 to 10 were obtained during the photic stimulation. Time course-signal processing was carried out using the correlation technique for each pixel [8]. The box-car waveform was used as the reference waveform [8]. The correlation coefficient (cc) varied between +1 and -1. A threshold value of TH is selected within 0 and +1. The pixel which is on the condition such as $cc \geq TH$ are superimposed on a image acquired during the rest period. Figure 4 shows the correlation data superimposed images with TH=0.6 and

the time course data varying flip angle from 90° to 20° with TR/TE = 58/30(msec). The sensitivity of the susceptibility effect difference is increased with decreasing flip angle as shown in Fig. 4. In conclusions, the modified SSFP fast imaging is used to obtain an interferogram which represents the field inhomogeneity and its application to functional imaging. The results obtained show that the field inhomogeneity maps derived from the proposed method can effectively be used for the measurement of the oxygenation process during stimulation using the linear region discussed. Considering the speed of the method (short scan time) and the sensitivity obtained, the proposed method can be utilized for functional MR imaging in a much more direct fashion without interference from, for example, the inflow effect.

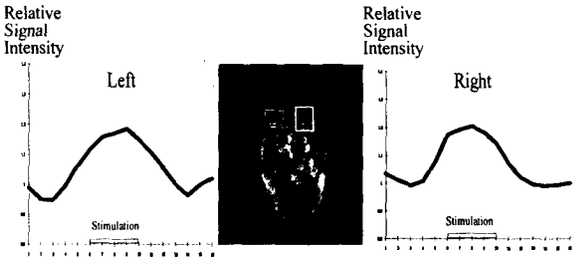

Fig. 4

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FIGURE CAPTIONS

Fig. 1 The SSFP fast gradient echo imaging sequence of both multi echo and double echo data acquisitions are shown. The applied RF pulse, reading gradient, and the various generated echoes are shown. The shading area represents the difference time between FID and echo.

Fig. 2 A modified SSFP fast gradient echo imaging pulse sequence for the field inhomogeneity mapping. By suitable adjustment of the reading gradient, an interference pattern of the FID and echo signals can be obtained. This interferogram directly represents the field inhomogeneity.

Fig. 3 Normalized voxel signal intensities as a function of the free precession angle θ which is directly proportional to the field inhomogeneity. This example is set between 0 to 2π for three sets of given parameters α, E_1, E_2.

Fig. 4 The SSFPI activated data with cc≥0.6 superimposed on an image acquired at a "rest" period and the time course data with flip angle of 20° and TR/TE=58/30 (msec). The time course data of the regions indicated by squares on the images are plotted at the sides to demonstrate the susceptibility dependent functional imaging capability of the proposed SSFPI technique.