Gradient Echo Plural Contrast Imaging (GEPCI)

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INTRODUCTION: In magnetic resonance imaging different versions of spin echo (SE) pulse sequences are used to create T2 contrast or to measure T2 relaxation rate constant. The major reason to use SE techniques stems from the refocusing nature of the 180° pulses which substantially reduce the influence of unwanted magnetic field inhomogeneities on MR signal. However these sequences suffer from motion artifacts and imperfections in slice profiles resulting from the presence of the 180° refocusing pulses. More importantly SE techniques suffer from restrictions on RF power deposition. These factors limit clinical applications of SE techniques especially at high field MR imaging. It was shown previously in phantom studies that the influence of the magnetic field inhomogeneities on MRI signal can be removed by making use of a multi-gradient-echo approach in combination with special signal post-processing procedures (1). This proposed method which we refer to as Gradient Echo Plural Contract Imaging (GEPCI) allows generation of T2, T2*, T1 and spin density images in a single scan with the advantage that all such images are naturally co-registered. Other advantages of the technique are (i) fast imaging (TR on order of 50ms), (ii) low flip angles results in low RF power deposition and better slice profiles, (iii) both 2D and 3D versions can be implemented, and (iv) method is insensitive to the RF inhomogeneities. Here we demonstrate *in vivo* application of the proposed technique at 3T.

THEORY: Traditionally, it is assumed that the free induction decay (FID) signal after and RF excitation behaves as exponential function:

$$S(TE) = S(0) \cdot \exp(-R2^{*} \cdot TE) \qquad (1)$$

where *R2* = R2 + R2'* is relaxation rate constant that has contributions from both tissue-specific *R2* effects and *R2'* effects of magnetic field inhomogeneities including those that are not tissue specific. Therefore separation of tissue-specific characteristics described by *R2* relaxation from those, described by tissue-nonspecific contributing to *R2'* relaxation is an important for clinical and scientific applications.

It was previously demonstrated theoretically and on phantoms (2) that part of an FID signal decay due to *macroscopic* field inhomogeneities is an even function of TE and can not be described in terms of exponential decay. Specifically,for short times *TE*, it can be approximated by gaussian or polynomial functions. It has also been shown (3) that the reversal part of FID signal due *mesoscopic* field inhomogeneities is also a gaussian function for short times. Therefore, for short times (TE < characteristic time) FID decay can be well approximated as a polynomial or gaussian functions and the most general solution to the problem is given as (2):

$$S(TE) = S(0) \cdot \exp(-R2 \cdot TE).(1 - \sum \alpha_m \cdot TE^{2m}) \qquad (2)$$

This result is remarkably different from the one described by Eq. [1], where contribution of mesoscopic and macroscopic field inhomogeneities was incorrectly assumed to be exponential. This non-exponential behavior of the MRI signal in Eq. [2] can be used to separate desirable information about *R2* relaxation from the effects of mesoscopic and macroscopic field inhomogeneities. In many practical cases the coefficients α_m are small and only the first two terms of the series are important, hence the series can be approximated as a first order polynomial.

METHODS: We have implemented a 2D and 3D multi-gradient-echo sequence on Siemens 3T Magnetom Allegra system. *In vivo* images were acquired with the following parameters: in plane resolution of 1×1 mm^2 (matrix size = 256×256), slice thickness = 5 mm, number of slices = 14 and TR = 1 sec. 11 gradient echo images were acquired with TE(0) = 7ms and echo spacing of 4ms. R2 (T2 = 1/R2) maps were obtained by fitting Eq. [2] (the series was terminated at the second order polynomial) to the time domain course of signal intensity on pixel by pixel basis. Reference T2 weighted turbo-SE images with the same resolution were also acquired at TE = 88ms and TR = 4 sec.

RESULTS: Fig 1 shows example of curve fitting of data by Eq [1] and Eq [2]. The graph shows that exponential function for FID signal decay Eq [1] does not accurately fit the experimental data whereas the Eq [2] fits the data very well. Figure 2a shows T2 map obtained from the method described above and the scale on the map ranges from 0 to 120 ms. Figures 2b, and 2c show T1 weighted and T2* weighted images from the same data set demonstrating that multiple contrast images can be obtained in the same sequence. Figure 2d is the reference T2 weighted image obtained by turbo-spin echo sequence and shows contrast that is in agreement with the T2 map (Fig 2a).

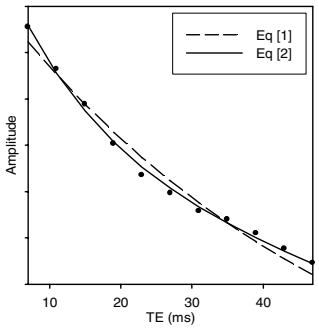

Fig 1. Example of fitting model to data. The GEPCI proposed method (Eq [2]) fits the data very well.

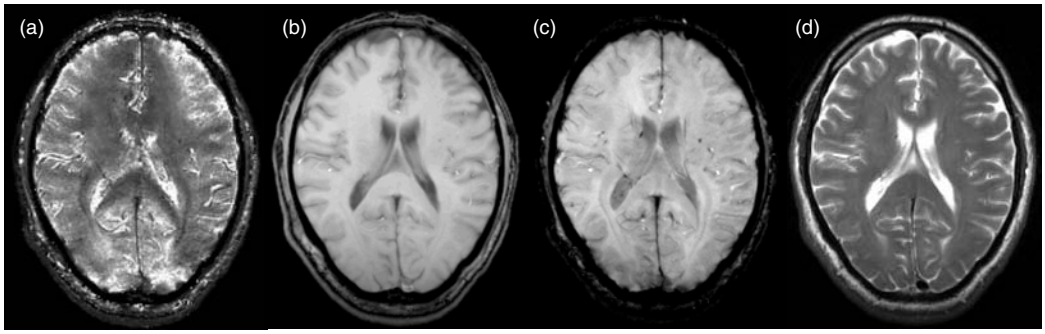

Fig 2. (a) T2 map obtained by fitting the data with Eq [2]. (b) T1 weighted image. (c) T2* weighted image. All three images were acquired in the same sequence. (d) Reference T2 weighted image from turbo-spin echo sequence.

CONCLUSIONS: We have demonstrated that quantitative evaluation of tissue T2 relaxation time constant can be obtained by gradient echo MRI without using refocusing 180° RF pulses. The technique does not require any special hardware modification and can be implemented on any clinical MRI scanner. High resolution imaging is also favorable for this technique because smaller voxels result in less magnetic field inhomogeneity across an imaging voxel and thus will improve the fitting results. Magnetization preparation can be implemented in the sequence to introduce T1 contrast in the GE images and/or to eliminate the signal from CSF resulting in plural contrast images in the same sequence.

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