Correction of in-plane intra-voxel dephasing effects in gradient echo images.

Peter van Gelderen[1], Jacco A. de Zwart[1], Jeff H. Duyn[1]
[1]Advanced MRI Section, LFMI, NINDS, National Institutes of Health, Bethesda, MD, United States

Introduction

Gradient echo (GRE) MRI at high field provides unique anatomical information not available with other contrast. However, its sensitivity to macroscopic field gradients may lead to signal dropouts that result from dephasing effects and confound the microstructural information reflected in signal amplitude and relaxation rate measures derived from GRE data. As the mechanism underlying these dropouts is well understood, correction may be possible. In fact, through-plane dephasing calculation and correction (e.g. [1]), as well as compensation with additional acquisitions (Z-shimming [2]), have been proposed. In-plane effects have also been studied [3-5], and similar to Z-shimming, extra data can be acquired to correct them [6]. Here we present a simple and improved method to calculate the in-plane dephasing effects in order to derive a correction factor for GRE image intensity, without the need for additional data acquisition.

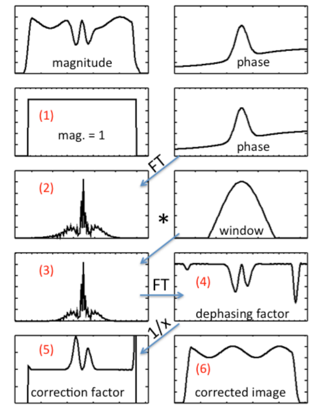

Fig 1. Illustration of the correction procedure. Top row is the original image data, see Methods for explanation of the numbered steps.

Methods

Current methods for calculation of in-plane dephasing effects analyze signal from a single voxel and ignore surrounding tissue. This generally underestimates the dephasing effects, as it ignores the true shape of imaging voxel, given by the point-spread function, which results in mixing of neighboring signal from neighboring voxels. This means that proper calculation of the dephasing effects needs to take into account the entire object.

The proposed solution is to effectively recreate the imaging procedure (Fig 1), starting from a reconstructed, complex image: 1) Create an object with uniform intensity, but with the phase distribution of the image to be corrected; 2) Generate k-space data by applying a Fourier Transform (FT); 3) Multiply with the same apodization window used as in the original reconstruction; 4) Calculate the resulting image intensity by inverse FT; 5) Invert this dephasing factor where sufficiently larger than zero to use as a correction factor; 6) Multiply the magnitude image by this correction factor map.

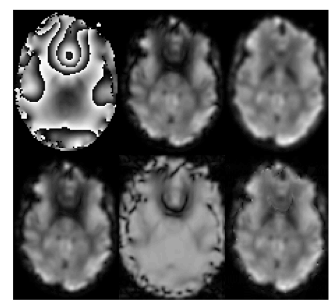

Fig. 2. Correction applied to down-sampled data. A) phase, b) image with and c) w/o dephasing; d) correction with single voxel method, e) dephasing factor and f), correction with proposed method.

Results & Discussion

GRE images were acquired on a healthy volunteer under an IRB approved protocol, on a Siemens Magnetom 7T system with a Nova 32 channel receiver. Six echoes were acquired (TE 7.5-38.5ms). The channels were combined into complex images using a SENSE matrix based on the first echo [7,8]. A Gaussian apodization window was used in the image reconstruction. The resolution was 2x2mm^2; 24 2mm slices were acquired with 1mm gap. The off-resonance frequency was fitted voxel by voxel over the multiple echo times; the phase at the shown TE was recalculated from this fitted frequency.

To demonstrate the proposed method and compare it to the single voxel calculation, one slice at TE 17.5ms was chosen and the complex data blurred to half the resolution (4x4mm^2) on which the dephasing was calculated with both methods. As a reference the magnitude image was blurred as well (with zero phase, this blurring does not suffer from intra-voxel dephasing). The results in Fig. 2 show the proposed correction approximates the reference closely, except for the area where the dephasing is almost complete and brings the signal is close to zero.

The second illustration is performed on the original resolution data with TE 31.5ms, and shown Fig. 3. For this case the phase was unwrapped in 3D to calculate the through-plane dephasing effect as in [1]. The in-plane correction proposed here does not require unwrapping of the phase data. One slice of the image set is shown in Fig. 3, together with the through plane and in-plane correction and their combination. The image uniformity improves notably in the in the frontal area.

The proposed correction method can be applied to 2D as well as 3D GRE data. The phase information can be based on the (complex) image data itself, or some other source of field information. In the implementation presented here, a uniform image was used as starting point for the calculation of the correction factor. Various improvements are possible, including the use of the uncorrected image as starting point, and an iterative approach with sequential updates of the starting point with corrected images. The method fails in areas where dephasing results in near-zero signal. As no extra data is acquired, substantial signal loss cannot be recovered, and this method does not improve the signal to noise. The magnitude of the in-plane correction is dependent on voxel dimension and, for isotropic voxels, is of similar amplitude as the through-plane correction.

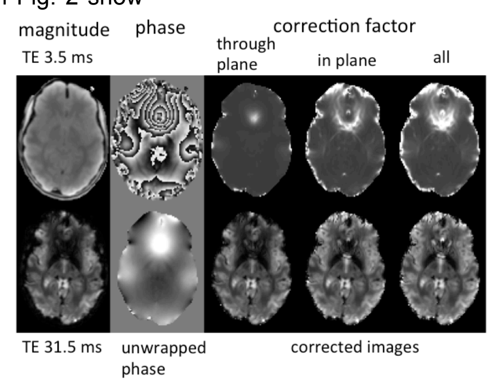

Fig 3. Example of GRE image correction, the three right most columns showing the through plane, the proposed in-plane, and combined corrections of intra-voxel dephasing.

References:
[1] Fernandez-Seara MA, Wehrli FW. Magn Reson Med 2000;**44**:358-366.
[2] Frahm J, Merboldt KD, Hanicke W. Magn Reson Med 1988;**6**:474-480.
[3] Reichenbach J. et. al., J Magn Imaging; **7**:266-279.
[4] Yablonskiy DA. Magn Reson Med 1998;**39**:417-428.
[5] Hwang D, Kim DH, Du YP. Neuroimage 2010;**52**:198-204.
[6] Bakker C, Seppenwolde J, Vincken K, Magn Reson Med 2006; **55**: 92-97.
[7] Pruessmann K, et.al., Magn Reson Med 1999; **42**: 952-962.
[8] de Zwart J, et.al. . Magn Reson Med; **48**: 1011-1020.