Simultaneous estimation of I_o, R_2*, and field map using a multi–echo spiral acquisition

Bradley P. Sutton[1], Scott J. Peltier[2], Jeffrey A. Fessler[1], Douglas C. Noll[1]
[1]Dept. of Biomedical Engineering, [2]Dept. of Applied Physics
Functional MRI Laboratory, University of Michigan, Ann Arbor, MI

ABSTRACT

Functional MRI using R_2^* maps may be more quantitative and robust than using T_2^*-weighted imaging. Standard estimation techniques are confounded by magnetic susceptibility and inhomogeneity. Even with standard field corrections, errors in estimation of the field map can persist in the R_2^* and I_o maps. In this work, we perform an iterative simultaneous estimation of the field map, R_2^* and I_o. The results in simulation and human study show that this method is substantially more accurate in determining these parameters than standard estimation schemes.

INTRODUCTION

Recent multi-echo studies have shown an echotime (TE) dependence in components of the fMRI signal, with interest taken in using R_2^* as a measure of functional activation [1]. A standard method to measure R_2^* is to reconstruct images at multiple echo times and fit an exponential decay to the pixel values [4]. The R_2^* maps obtained in this manner are often noisy as the fit is performed on relatively few time points (ie. 4 to 10).

Macroscopic effects of R_2^* and the field map cause degradations and distortions in single-shot gradient echo images, such as as spiral acquisitions. Correcting these distortions can lead to more accurate gradient-echo imaging in general, and more accurate R_2^* maps for functional studies.

The standard method to estimate field maps is to acquire full FOV images at two different echo times and divide the pixel-by-pixel phase difference by the difference in echo times [2]. Both this standard field map estimation and the estimation of R_2^* maps assume that the entire k-space acquisition occurs at the echo time.

To account for interactions between R_2^*, I_o, and field inhomogeneities, we proposed to perform a regularized nonlinear least-squares joint estimation of the I_o image, R_2^* map and field map based on modeling the signal equation.

METHODS

A multi-echo spiral pulse sequence with 4 echo times (TE=4.8/25.28/45.76/66.24ms TR/FA/FOV=500ms/45/20cm, Matrix size=62, 400 time points) was implemented on a GE 3T Signa scanner (GE Medical Systems, Milwaukee, WI). The first readout in the time series had echo times delayed by an additional 2.5ms in order to form a field map in the standard way, using just the first spiral of the sequence at two different echo times. This fieldmap was used as an initial estimate in our iterative algorithm and was also used to correct the time-series images for the standard method using a conjugate phase reconstruction [3, 5].

For our iterative method, the cost function to be minimized is given by:

$$J(I_o,\omega_0,R_2^*) = \frac{1}{2}\sum_i |(y_i - u_i(R_2^*,I_o,\omega_0))|^2 + \beta \sum_{j_1,j_2 \text{ neighbors}} (b(r_{j_1}) - b(r_{j_2}))^2 \quad (1)$$

where $\boldsymbol{y}$ is the data during the readout of all four spirals, $\boldsymbol{u}$ is the modeled signal equation, depending on the estimated spin-density image, I_o, field map, ω_0, and the R_2^* map, and $b(r) = R_2^*(r) + i\omega_0(r)$. The model for the signal equation is given by: $u_i(R_2^*, I_o, \omega_0) = \sum_j I_o(r_j)e^{-i2\pi k_i r_j}e^{-b(r_j)t_i}$, where r_j indicates position and k_i is the k-space trajectory. The last term in the cost function (1) penalizes roughness in the R_2^* and field map. Since the cost function depends on the current estimate of the field map, the R_2^* map, and the image, we alternate reconstructing the image using our current estimate of the R_2^* and field maps and then updating the R_2^* and field maps using the current estimate of the image. We take advantage of the linearity of the image reconstruction problem and use the conjugate gradient method to find an estimate of the image using all of the data from all of the time points. Taking the derivative of the cost function with respect to $\omega_0(r_j)$ and $R_2^*(r_j)$, we can update the estimate of the R_2^* and field maps using gradient descent.

RESULTS

An ellipsoid object was simulated with R_2^* and field inhomogenity to compare various estimation methods when the truth was known. Typical values for gray and white matter R_2^* were used [6]. The results for using a linear fit on the natural log of the data, a nonlinear fit using the Gauss-Newton method, and our simultaneous estimation method are shown in Figure 1. The simultaneous estimation method has reduced the error to around 6% in both R_2^* and I_o by the 20th iteration, which is dramatically better than the standard fit methods, especially for R_2^*. This is further seen in the R_2^* profiles, where error in the field map estimation has resulted in overestimation of R_2^*. Figure 2 shows the results on a typical slice from the human subject.

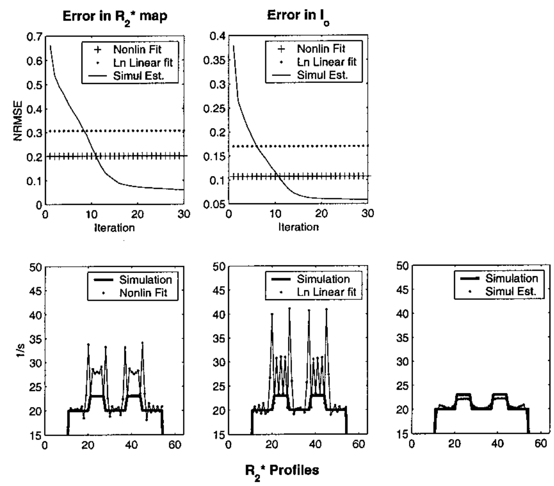

Figure 1: NRMSE in R_2^* and I_o maps along with profiles of R_2^* for the two standard estimation schemes and the proposed simultanesous estimation.

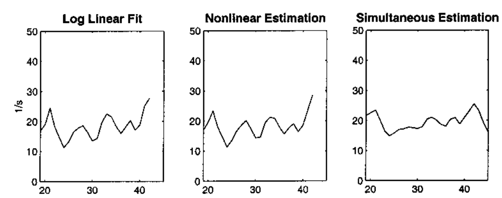

Figure 2: Profiles of R_2^* map for standard and simultaneous estimation in human subject.

DISCUSSION

Our regularized nonlinear least-squares joint estimation method shows increased accuracy in determining R_2^*, field map, and I_o. The method uses the whole timecourse of the k-space acquisition and models the signal equation using current estimates of the parameters. This will aid in accurate quantitation of tissue parameters and detection of BOLD R_2^* modulation.

References

[1] V. G. Kiselev, S. Wiese, and S. Posse. In *Proceedings of 7th ISMRM*, 1999.

[2] K. S. Nayak and D. G. Nishimura. *Magnetic Resonance in Medicine*, 43:151–154, 2000.

[3] D. C. Noll, C. H. Meyer, J. M. Pauly, D. G. Nishimura, and A. Macovski. *IEEE Transactions on Medical Imaging*, 10(4):629–637, 1991.

[4] S. Posse, S. Wiese, D. Gembris, K. Mathiak, and C. Kessler. *Magnetic Resonance in Medicine*, 42:87–97, 1999.

[5] H. Schomberg. *IEEE Transactions on Medical Imaging*, 18(6):481–495, 1999.

[6] J. P. Wansapura, S. K. Holland, R. S. Dunn, and J. W. S. Ball. *Journal of Magnetic Resonance Imaging*, 9, 1999.

This work was supported in part by a Whitaker Foundation Graduate Fellowship