New Approach for Estimating dR2* in fMRI

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Synopsis

We introduce a framework in which ΔR_2* in an fMRI experiment can be estimated directly from a single echo, estimate of a baseline image and a baseline R_2*. This approach uses a linearized model of the difference signal between the non active signal and active signal during a task along the time course. ΔR_2* maps, are estimated at each time point using the Conjugate-Gradient (CG) method. Dynamic estimate images of ΔR_2* were created for a study of motor activation using a finger tapping task. These maps were also thresholded to yield the expected areas of activation in primary motor cortex.

Introduction

For functional MRI (fMRI), accurate estimates of R_2* maps can be useful for quantifying dynamics of blood volume and oxygenation, during task performance. However, conventional estimates of R_2* maps involve assessing fractional signal changes which include a division by magnitude of the image, making it sensitive to noise and edges. More recently, R_2* maps have been estimated from multi-echo data, by doing a log-linear or non-linear fit on the maps at each echo and for each time point [1,2], however, this can lengthen the total acquisition time for a slice. We propose an approach using a single T_2*-weighted image combined with a model in which R_2* relaxation effects are incorporated in the signal equation. Separating the R_2* changes into a static baseline map and a dynamic ΔR_2* map, we linearize the signal equation and reconstruct the ΔR_2* maps iteratively using CG.

Method

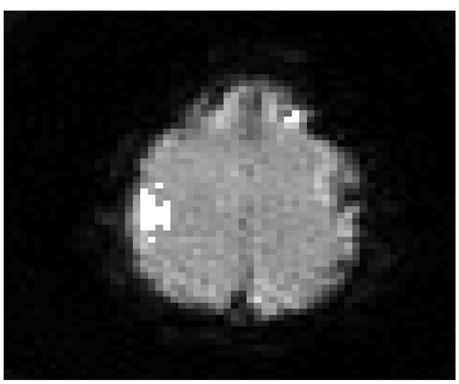
The signal equation used here is given by

$$s(t) = \int f(r) e^{-tR_2^*(r)} e^{-i2\pi(k(t)\cdot r)} dr ,$$

where the R_2* decay is modeled as an exponential decay. With BOLD inducing changes in the R_2* map, the map will decrease in amplitude by a small fraction (<10%) relative to the baseline map. Using this, we can write R2* as the difference of the baseline map and ΔR_2*. Since the decay due to ΔR_2* is relatively small, we linearize that part using Taylor's expansion. Thus we can formulate the signal equations with and without functional changes (BOLD effect). Using the difference of the two and accounting for noise we model the difference of raw data, i.e. non-active vs. active. That system is then iteratively solved for ΔR_2* using the CG method with regularization.

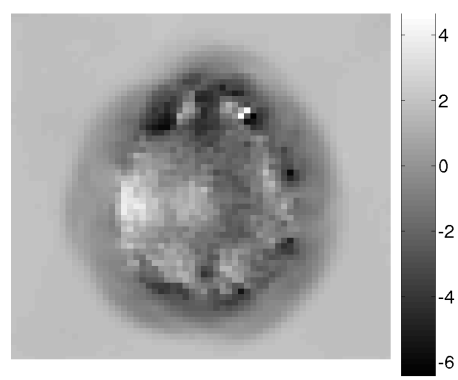
We applied this scheme to human data (block task), acquired using a spiral, as follows,

1. To get the baseline image for the system equation, we averaged the 10 first spiral acquisitions (no task), and estimated it. We applied a log-linear fit to a single multi-echo readout to estimate the baseline R_2*. From this we built our system equation.
2. At each subsequent time point, we took the difference of the raw data and the averaged time points from before, and iteratively solved the system equation (12 iterations).
3. To get an estimate for the ΔR_2* map, we took all of the time points with and without activity (from block design), averaged over all and took the difference.



Results

Using a block designed task on GE 3T scanner, spiral acquisition, with TE = 25ms, the results can be seen in the figures shown. On the right, we can see the estimated ΔR_2* along with a quantative scale showing ΔR_2* in Hz. To the left we can see activation area for the block task, based on thresholding of the ΔR_2* map.



Discussion and Conclusions

We have demonstrated a framework in which ΔR_2* can be estimated directly from a single echo, estimate of a baseline image and a baseline R_2*. This approach allows for the dynamic estimation of ΔR_2* in blocked or event related fMRI studies. We demonstrate that the ΔR_2* maps can be thresholded to define areas of activation defined on changes in signal properties that relate to physiological parameters such as blood oxygenation and volume rather than statistical significance. This work is based on a model in which the difference signals are linearized which, in turn, allows the use of the rapid conjugate gradient estimation approach. Since this approach uses the complex raw data, there are a number of possible extension to this approach, for example, the simultaneous estimation of dynamic field shifts due to respiration and other effects. The use of raw data however, has some potential disadvantages, the most serious being correction for the effect of head motion. Given an estimate of head motion, in-plane motions can be addressed through simple modifications in the system matrix, however through-plane motions may require motion effect to be explicitly estimated along with the ΔR_2*.

References

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