Image based physiological noise correction for perfusion-based fMRI

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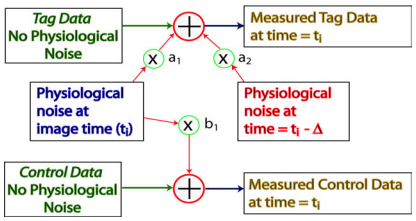
Introduction

Physiological fluctuations are often a dominant source of noise in fMRI experiments, especially at higher field strengths. A number of methods have been developed for the reduction of physiological noise in fMRI experiments. These include image based retrospective correction (RETROICOR) [1], *k*-space based retrospective correction (RETROKCOR) [2], and navigator echo based correction (DORK) [3]. At present the application of these methods has been focused primarily on experiments using blood oxygenation level dependent (BOLD) contrast. However, perfusion-based fMRI using arterial spin labeling (ASL) is becoming increasing popular because of its potential to better localize functional activation to the sites of neuronal activity [4]. In this work we investigate four extensions of RETROICOR to ASL. While significant improvement in statistical power were observed for each method, we found the greatest improvement resulted when 1) physiological noise is estimated separately for tag and control images and 2) the contribution of physiological fluctuations during the tagging process to noise in the tag images is included.

Theory/Background

Figure 1: Simplified Linear Model of an ASL experiment



In a typical ASL experiment tag and control images are required in an interleaved manner. Tag images are acquired at a delay on the order of 1 second after an inversion pulse that inverts the magnetization of arterial blood, while control images are acquired at the same delay after a control pulse that leaves the arterial magnetization fully relaxed. In the analysis of ASL experiments, it is useful to consider the tag and control image sets separately [5]. A simplified model of an ASL experiment is depicted in Figure 1. The measured tag and control images are assumed to be the sum of a noiseless image plus the physiological noise occurring at image acquisition time t_i. The relative impact of the physiological noise on tag and control image acquisition is modeled by the constants a_1 and b_1, respectively. To model the possible effects of cardiac pulsations and respiration on the inversion process, an additional term for physiological noise at time t_i - Δ is added to the tag image with a weight of a_2 , where Δ is the temporal delay between the physiological noise and the image. This term also models the possible modulation of CBF by respiration [6]. In order to estimate the effects of physiological noise using RETROICOR, a 2nd order Fourier series is expanded in terms of cardiac and respiratory phases that are calculated from cardiac and respiratory activity recorded during the experiment [1]. The weights of the Fourier terms can then be estimated with a general linear model. In applying RETROICOR to ASL data, we consider 4 possible methods. In method 1 we assume that the noise affects the tag and control images equally ($a_1=b_1$) and there is no delayed tag term ($a_2=0$). This can be considered to be the direct application of RETROICOR to ASL. Method 2 allows for the possibility that the noise affects the tag and control images differently ($a_1 \neq b_1$), while methods 3 and 4 are modified versions of 1 and 2, respectively, that include a delayed tag term ($a_2 \neq 0$). In order to assess the relative performance of the different methods, an F-statistic can be computed for a general linear model in which a reference waveform (smoothed block design) is treated as the regressor of interest and the physiological noise terms are treated as nuisance terms [5]. In addition, a constant and linear term are included as nuisance terms. The F-statistic has the useful property of explicitly taking into account the reduced degrees of freedom due to adding more nuisance terms.

Methods

Three oblique 8mm slices through the visual cortex where imaged while the subject was shown a full-field, 8 Hz radial flickering checkerboard (block design comprised of 4 periods of 30/30 seconds on/off). The design was repeated two times over 5 subjects. Imaging data were acquired on a Varian 4T whole-body system with a head transmit coil and a surface receive coil (Nova Medical) placed under the occipital lobe. Data were acquired using a PICORE-QUIPSS II [7] sequence with an EPI readout, interleaving of tag and control images, and TR = 2s, TI1/TI2 = 700/1400 ms, TE = 27 ms, θ = 90, FOV 24cm, 64x64 matrix. Image data were co-registered to minimize the effects of subject motion. Cardiac pulse and respiratory effort data were monitored using a pulse oximeter (NONIN) and a respiratory effort transducer (BIOPAC), respectively. Physiological data were sampled at 40 samples per second. F-statistics for each of the methods were calculated for each pixel within a region of interest (ROI) defined to encompass the visual cortex. To determine the optimal delay Δ to use with methods 3 and 4, the F-statistic was calculated for uncorrected pixels that showed significant perfusion ($p < 0.05$) at delay times varying between 0s to 1.5s at intervals of 25ms. The optimal Δ's for each method were chosen to maximize the average F-statistic over each slice.

Results and Discussion

Figure 2: Average F Statistic vs Delay

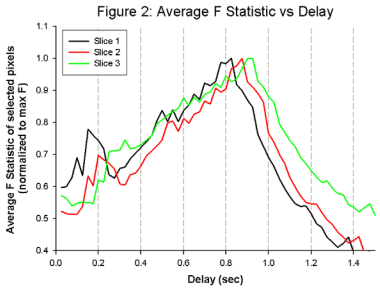
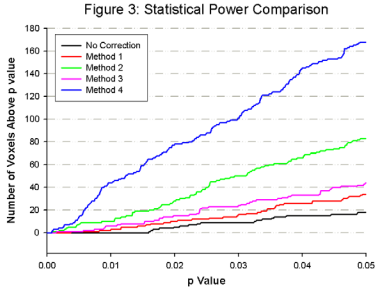


Figure 3: Statistical Power Comparison



Results are shown for subject 1. Similar results were observed for the other subjects. Figure 2 shows the average F statistic value (normalized to max F-statistic) versus delay time, Δ. It shows that the F statistic for method 4 is maximized when the physiological noise is estimated at delays of 850 ms, 900 ms and 950 ms for slices 1, 2, and 3, respectively, which is consistent with the inter-slice delay of 50 ms. Nearly identical values were obtained for method 3. The optimal delay time is consistent with the temporal range of the arterial bolus created by the QUIPSS II pulse sequence, which spans the time between the inversion pulse (Δ = 1400ms) and the QUIPSS II saturation pulse (Δ= 700ms). To compare the proposed methods, Figure 4 plots the number of voxels (within a ROI) that have an F-statistic above a threshold corresponding to p values between 0 and 0.05. For any given p value, method 4 provided the largest number of significant voxels. It is also worth noting that method 3 is an improvement over method 1. These findings indicate that: 1) tag and control images are affected by physiological noise differently, and 2) physiological fluctuations that occur during the tagging process significantly impact the tag images. Although the methods described here are image-based, the results may also prove to be useful for non-image-based methods. For example, acquisition of an additional navigator echo prior to the tagging pulse may improve the performance of navigator-based techniques.

References: [1] Glover, G.H. et al., *MRM* 44:162-167, 2000 [2] Hu, X. et al., *MRM* 34:201-212, 1995 [3] Pfeuffer, J. et al., *MRM* 47:344-353, 2002 [4] Luh W.M. et al., *MRM* 44:137-143, 2002 [5] Liu, T.T. et al., *Neuroimage* 16:269-282, 2002 [6] Wise, R.G. et al., *Proc 10th ISMRM* p. 216, 2003 [7] Wong, E.C. et al., *MRM* 39:702-81, 1998.