An Adaptive Cardio-Respiratory Filter for MRI Time Series Data

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Introduction

MRI can be used for the study of dynamic processes (e.g. brain activation) through acquisition of time series data. Such data can however be affected by quasi-periodic events unrelated to the process being investigated. Potential artifact mechanisms are blood flow pulsations related to the cardiac cycle (CC), and tissue motion and changes in local magnetic field related to the respiratory cycle (RC). This not only reduces the statistical power of fMRI experiments, but also causes temporal signal correlation of otherwise uncorrelated cortical areas, affecting interpretation of fMRI studies of event-related activity or neuronal connectivity.

Methods for reducing artifacts related to CC and RC have been suggested, e.g. band rejection filtering [1]. However, CC and RC are typically temporally undersampled, complicating the separation of desired from undesired signals. Glover et al. proposed a method (RETROICOR) which operates in image domain and models CC and RC by Fourier-series fitting [2]. Here a model-free variation of this approach is presented, which performs selective averaging to derive an estimate of CC- and RC-related artifacts. The method is validated by evaluating its performance relative to RETROICOR.

Methods

An estimate of the temporal characteristics of the artifact can be obtained with a precision that exceeds the temporal resolution of the MRI data acquisition by sorting and averaging of the acquired MRI data based on their acquisition time relative to the nearest artifact occurrence ('event') (Fig. 1). The period encompassing each event is subdivided in a number of discrete intervals ('bins'). All MRI data with acquisition times falling within the same bin are averaged on a voxel-by-voxel basis, resulting in suppression of signals whose timing does not correlate with artifact events, while amplifying signals that correlate with the artifact. The resulting data are referred to as 'artifact estimate', from which DC signal is removed. Finally, each bin in 'artifact estimate' is subtracted from all MRI data that had been assigned to this bin based on their acquisition time relative to the nearest artifact event.

Performance was evaluated on a GE 3 T scanner. Two datasets were acquired on each volunteer (n=6): a 17-slice dataset with 1.7 s TR and 90º flip angle (366 volumes; 622.2 s acquisition time; 45 ms TE; 2.3×2.3×3.0 mm^3 nominal resolution; labeled 'slow'), resembling a typical fMRI experiment, and a single-slice dataset (100 ms TR; 6222 volumes; 15º flip angle; referred to as 'fast'), aligned to the center slice of 'slow', which ensured unaliased acquisition of CC- and RC-related artifacts. CC and RC were recorded using a pulse oximeter and respiratory bellow (1 kHz sampling).

For comparison, data were separately filtered with RETROICOR (a C-implementation was obtained from the Radiological Sciences Lab at Stanford) but otherwise processed identically.

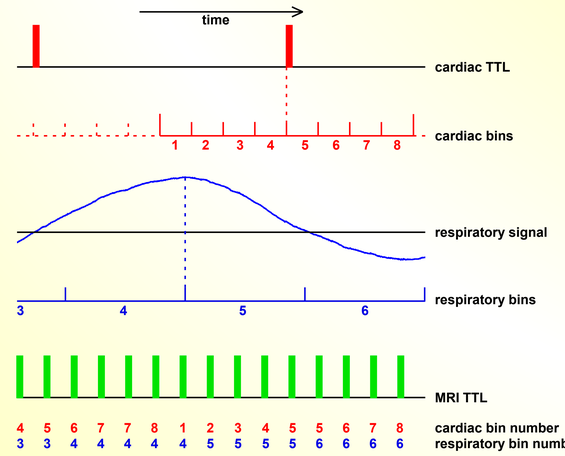

Figure 1: Schematic representation of the filtering method. Both the cardiac cycle (CC) and respiratory cycle (RC) are subdivided into bins. In the example shown here, 8 bins are used for both cardiac and respiratory filtering. Information about CC and RC timing is derived from physiological monitoring data. Filtering for CC and RC is performed independently. MRI data are assigned to a bin based on their acquisition time relative to the nearest event.

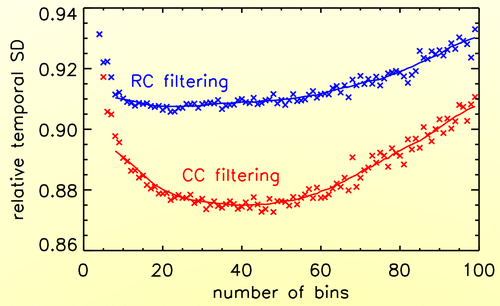

Figure 2: Plot of relative SD_t as function of the number of bins. Red represents CC-, blue RC-filtered data. Symbols are the computed relative SD_t values as a function of the number of bins, the solid lines the result of 16-fold smoothing.

Results

Temporal standard deviation (SD_t) after filtering was computed, corrected for reduced degrees of freedom, and compared to SD_t before filtering as a measure of filtering performance. 'Slow' data were filtered with a range of numbers of bins (Fig. 2), an optimum of 40 for CC and 21 for RC was found.

Performance of 'fast' was evaluated by computing the average spectral intensity in the frequency band containing the prinicipal CC or RC artifact. Example spectra for a single voxel are shown in Fig. 3. Maps of average spectral intensity before and after filtering are shown in Fig. 4 for one volunteer. In addition, 'fast' data were filtered as 17 independent subsets (e.g. volumes [1,18,35,...,6121]), resembling 'slow', before being recombined. This demonstrates that a similar level of artifact suppression can be obtained when the artifact is undersampled (additional peaks at n/1.7 Hz are caused by small differences in baseline correction for each subset).

The level of artifact suppression in all volunteers (867 voxels for CC, 2064 for RC, of which 275 in both CC and RC) is shown in Fig. 5, both for our method and for RETROICOR. Artifact is suppressed to approximately noise level in most CC- and a large number of RC-voxels. Voxels that perform poorly in our method typically also perform poorly in RETROICOR, suggesting that the remaining spectral intensity might be unrelated to CC or RC (e.g. presence of low-frequency fluctuations in RC band, see Fig. 3). Average SD_t reduction in CC voxels was 12.5% for 'slow' (up to 55.7%) and 9.6% for 'fast' (up to 63.3%), compared to 6.3% and 7.8% respectively for RETROICOR. For RC voxels, improvement was 9.3% for 'slow' (up to 52.9%) and 4.9% for 'fast' (up to 25.2%), compared to 8.0% and 4.1% with RETROICOR. Data filtered for both CC and RC (275 voxels) showed improvement of 13.2% for 'slow' and 9.0% for 'fast', RETROICOR yielded 8.2% in both cases.

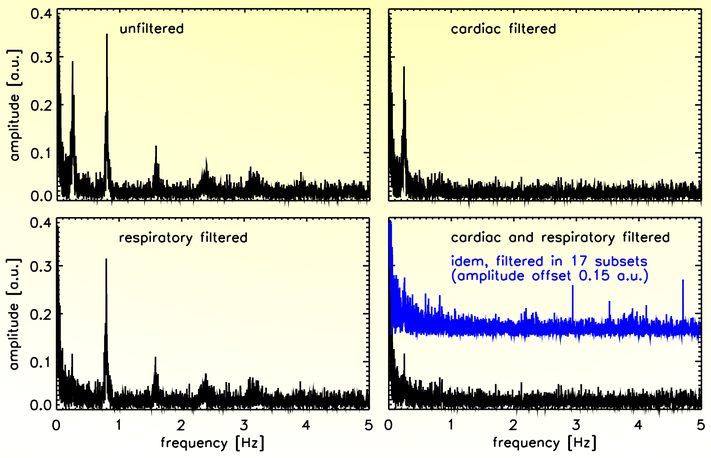

Figure 3: Single-pixel frequency spectra for a 'fast' scan, obtained before and after filtering. The top-left plot shows the frequency spectrum for a pixel in the unfiltered dataset. The top-right plot shows the same pixel after application of only the cardiac filter, the lower-left plot respiratory-only filtered data. The lower-right plot shows the frequency spectrum after application of both filters. The blue line, offset 0.15 units along Y, is the result of filtering for cardiac and respiratory artifacts after splitting the data in 17 subsets (see text for details).

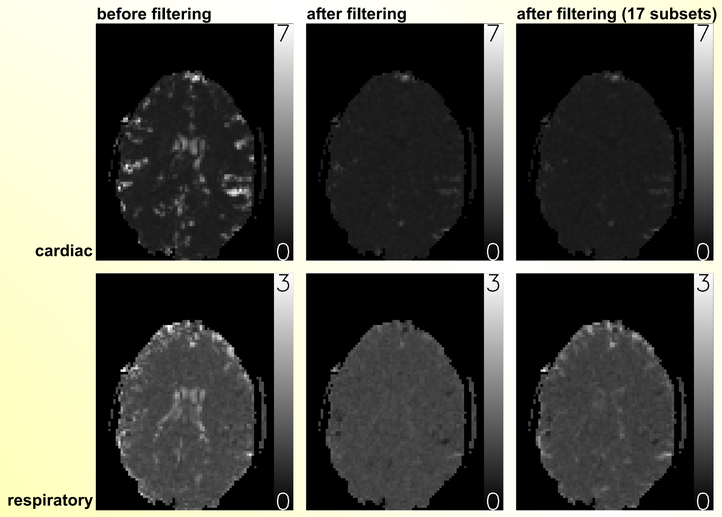

Figure 4: Maps of artifact intensity before and after filtering for a 'fast' scan, showing the average intensity in a frequency band surrounding the principal artifact frequency in the magnitude spectrum before (left column) and after (center and right column) filtering. The top row shows cardiac (1.18-1.48 Hz for this volunteer), the bottom row for respiratory (here: 0.20-0.50 Hz).

Conclusion

A model-free, flexible, adaptive filter for MRI time series data has been demonstrated in an fMRI-like setting. It provides artifact suppression to noise level with no significant effect on signals unrelated to the cardiac and respiratory cycle, even when these artifacts are undersampled in the MRI data acquisition. Under the current experimental conditions and for the level of variation in CC and RC encountered during these experiments the performance of this method is at least equivalent to the RETROICOR method.

References

[1] Biswal B, DeYoe EA, Hyde JS, Magn Reson Med 35:107-113 (1996)
[2] Glover GH, Li T-Q, Ress D, Magn Reson Med 44:162-167 (2000)

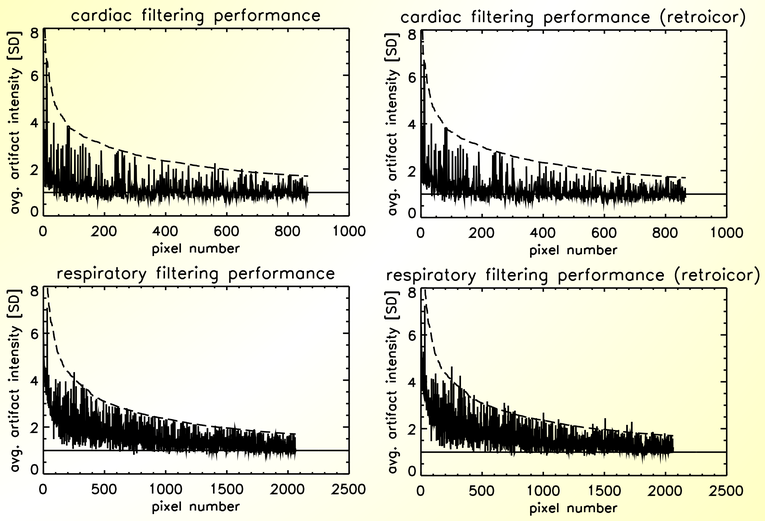

Figure 5: Plots of the achieved artifact reduction for all voxels with a significant cardiac (top row) or respiratory (bottom row) artifact in the 6 volunteers. Data are based on the decrease in average spectral intensity in a 0.15 Hz band encompassing the principal cardiac artifact frequency. The dotted line represents the artifact level before filtering. It is a decreasing curve since voxels were sorted on pre-filtering artifact intensity. The horizontal line at 1 SD indicates noise level. The solid line shows the artifact intensity in the same frequency band after filtering. Left column shows the result of filtering with the method described here, right column the result obtained using the RETROICOR method [2].