# Physiological artifact suppression in multi-shot data using covariance-map-enhanced navigator correction

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**Purpose**: MR signal stability, and thus image quality, is significantly affected by physiological noise[1]. Fluctuations linked to the respiratory cycle are typically the dominant source in the brain, increasing in significance with $B_0$. This can be compensated using navigator echoes[2] (in post processing) as well as with real-time shimming[3], in which shims and center frequency are dynamically updated throughout the experiment. The latter shows superior performance but requires hardware modifications. We propose to use a pre-scan similar to real-time shimming[3] to determine the spatial dependence of periodic temporal signal fluctuations throughout the brain, and to correct the effects of such fluctuations in image reconstruction rather than through hardware adjustments.

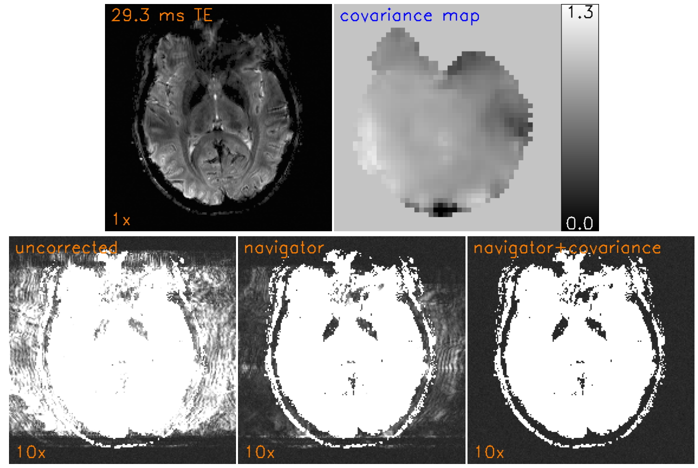


Figure 1: Simulation result for the case where experimental navigator and covariance data from a 2D multi-gradient experiment on a healthy human subject were used. The top row shows the final image (derived using the proposed method) and covariance map. Bottom row shows images (scale ×10) obtained when no correction is applied, after conventional 0th-order navigator correction, or with the proposed method.

**Theory:** Field fluctuations are assumed to be linear with respiration depth (e.g. chest position) and to show a constant spatial distribution, allowing the effects to be decomposed in a spatial and time dependent component: $\Delta B(r,t) = C(r) \cdot A(t)$.

This can be generalized to any signal fluctuation with constant spatial distribution $C(r)$. Although assumed in the case of respiration, $A(t)$ does not have to be linear. The covariance map $C(r)$ can be derived from a calibration scan, where $A(t)$ is recorded in parallel with its effect on the MR images (e.g. a series of phase maps from EPI). Once $C(r)$ is known, knowledge of $A(t)$ suffices to calculate the effect on subsequent MR acquisitions, which then can be corrected by incorporating the induced fluctuation, $C(r) \cdot A(t)$, into the image encoding matrix.

**Methods & Results**: Due to the way that multi-shot MR images are acquired, this correction is not immediately straightforward, requiring either the inversion of the modified encoding matrix or an iterative procedure. An iterative, conjugate gradient method[4] (CGM) approach is needed for large data (e.g. 3D acquisitions), since computational effort for matrix inversion scales to the third power with the number of phase encoding steps. We implemented both approaches, with our CGM implementation assessing performance in k-space to allow coil-combination (and SENSE reconstruction) to be integrated.

Preliminary data were acquired at 7 T (12-echo GRE; 1-s TR; 256×192 matrix; 8 slices; 1×1×2 $mm^3$; 32-channel coil). After reconstruction with a 0th-order navigator correction, background signal (outside the brain) was set to 0. Low-level noise was added. Artificial k-space data were then generated, accounting for fluctuating navigator phase (taken from the navigator echo in this experiment) and the covariance map for each phase-encode step. The covariance map was derived from image phase in a fast, low-resolution GRE scan (70 repetitions; 64×48 matrix; 8 slices; 4×4×3 $mm^3$) after sorting k-space lines according to navigator phase to obtain a 30-bin dataset as function of $A(t)$. Inclusion of this covariance map into the navigator correction through the proposed method substantially reduced ghosting artifacts as is evident in areas outside the brain (see Figure 1).

A Monte-Carlo simulation was also performed using the above approach to investigate the effect of changing the amount of covariance inhomogeneity (range 0.0–7.75× the experimentally measured deviation of $C(r)$ from 1.0) and navigator accuracy (random noise was added to navigator phase, range 0–1× SD of fluctuation in experimental navigator). No noise was added to the reference image to simplify assessment of residual artifact. Figure 2 shows the level of artifact (signal outside of object, known to be zero in source data) as a function of covariance inhomogeneity and navigator noise level. Results show that the proposed method performs well for all levels of covariance inhomogeneity, whereas conventional navigator correction fails with increased inhomogeneity. Increases in navigator noise level similarly affect conventional navigator correction and the proposed method.

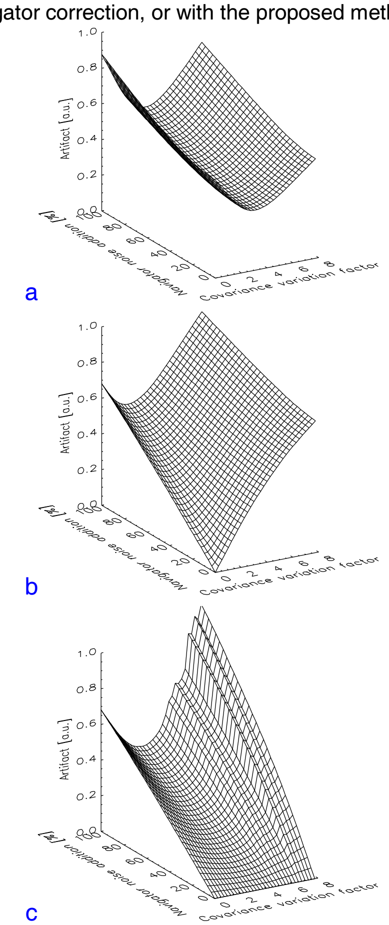


Figure 2: Results of Monte Carlo simulations. The level of covariance map inhomogeneity (relative to the experimental covariance map, see Figure 1) was changed, and noise was added to the navigator. Artifact is computed for uncorrected data (a), conventional navigator correction (b) and the proposed method (c).

**Discussion**: The covariance map $C(r)$ can be computed using a variety of reference signals, e.g. respiratory bellows[3], MR navigator echoes, or any other reference that describes a periodic source of temporal signal fluctuation, $A(t)$, such as mean signal phase over time in the brain in a series of images. Shot-to-shot assessment of a similar reference during a subsequent experiment allows higher-order correction of these imaging data. Although such correction is in principle possible with higher order navigators, these have practical limitations that can be avoided with the proposed method. The Monte Carlo simulation indicates that this can be done in a robust fashion.

**References**: [1]Hu, MagnResonMed 1995(34):201; [2]Ehman, Radiology 1989(173):255; [3]van Gelderen, MagnResonMed 2007(57):362; [4]Hestenes, JResNatlBureauStandards 1952(46):409