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Analysis and correction of field fluctuations in fMRI data using field monitoring

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ABSTRACT

This work investigates the role of magnetic field fluctuations as a confound in fMRI. In standard fMRI experiments with single-shot EPI acquisition at 3 Tesla the uniform and gradient components of the magnetic field were recorded with NMR field sensors. By principal component analysis it is found that differences of field evolution between the EPI readouts are explainable by few components relating to slow and within-shot field dynamics of hardware and physiological origin. The impact of fluctuating field components is studied by selective data correction and assessment of its influence on image fluctuation and SFNR.

Physiological field fluctuations, attributed to breathing, were found to be small relative to those of hardware origin. The dominant confounds were hardware-related and attributable to magnet drift and thermal changes. In raw image time series, field fluctuation caused significant SFNR loss, reflected by a 67% gain upon correction. Large part of this correction can be accomplished by traditional image realignment, which addresses slow and spatially uniform field changes. With realignment, explicit field correction increased the SFNR on the order of 6%.

In conclusion, field fluctuations are a relevant confound in fMRI and can be addressed effectively by retrospective data correction. Based on the physics involved it is anticipated that the advantage of full field correction increases with field strength, with non-Cartesian readouts, and upon phase-sensitive BOLD analysis.

1. Introduction

Functional MRI of the brain typically relies on time series of MR image data with suitable weighting, most commonly based on BOLD (blood-oxygen-level dependent) mechanisms (Bandettini et al., 1992; Ogawa et al., 1990). Brain activity and connectivity are inferred upon from the spatiotemporal signal structure of such time series. Any unrelated signal fluctuations act as confounds that limit the sensitivity of the technique.

Confounds in fMRI are of diverse origin (Murphy et al., 2013). In task-based studies all brain activity unrelated to the task is effectively a confound and its manifestation in fMRI data is often comprised in the notion of physiological noise. Confounds of physiological nature also include signal fluctuations due to respiration or heart rate variation (Chang et al., 2009; Birn et al., 2006; Chang and Glover, 2009), blood vessel pulsation (Mandeep et al., 1999), pulsatile blood flow and the associated subtle bulk motion of the head, as well as any other head motion (Power et al., 2012).

Further MR image fluctuations arise from imperfections of the instrumentation used and the electromagnetic fields involved. Regarding radiofrequency, the net gains of transmit and receive chains immediately affect the signal level of resulting data. Confounding gain changes can arise, e.g., from power amplifier fluctuations, instability of supply voltages, or changes in coil loading due to motion. Baseline and gradient magnetic fields, ranging from DC to few tens of kHz, fluctuate mostly due to imperfections of magnet and gradient hardware. However, low-frequency field perturbations also arise from the magnetic susceptibility of the subject in conjunction with physiological mechanisms, particularly with breathing (Pfeuffer et al., 2002; Raj et al., 2000; Van de Moortele et al., 2002) and, potentially, cardiovas-

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cular action (Gross et al., 2016; Pruessmann et al., 2011).

The present work focuses on this latter class of confounds related to magnetic field perturbations up to the audio-frequency range. Alterations of field strength change the Larmor frequency of nuclear spins and thus their phase accrual after excitation. Yet spin phase is also the carrier of spatial encoding in MRI. Therefore, field perturbations change how MR signal is depicted in resulting images. When the underlying field errors fluctuate over fMRI time series so do the depiction errors, which then act as confounds.

The form and magnitude of depiction errors caused by a given field perturbation depend strongly on the encoding strategy. The most common fMRI readout by far is single-shot 2D echo-planar imaging (EPI), which combines high spatiotemporal resolution with high SNR efficiency, relative robustness against motion, and sharper time assignment than segmented approaches (Bandettini et al., 1992; Mansfield, 1977). For considering the effects of field perturbation in single-shot EPI it is useful to distinguish slow field changes that are approximately static over each single-shot experiment (of typically several tens of ms) and higher-frequency field errors that vary within shots.

Slow field changes arise chiefly from magnet drift, temperature change of magnetised parts (Busch et al., 2014; Foerster et al., 2005), particularly of passive shims, and breathing (Pfeuffer et al., 2002; Raj et al., 2000; Van de Moortele et al., 2002; Windischberger et al., 2002). In single-shot EPI such slow-changing field offsets result primarily in image distortion by shifting image contents in the phase-encoding direction, by a distance proportional to the local field offset. To a smaller degree they also cause ghosting and blurring due to inconsistency of phase increments along odd and even k-space lines (Hennel, 1997). They strongly perturb the image phase, which however concerns fMRI only upon phase-sensitive data analysis (Calhoun et al., 2002; Rowe, 2005; Rowe and Logan, 2004), which is rarely performed to-date. Slow field fluctuations are commonly addressed by two strategies. At the acquisition stage, navigator readouts added to the sequence serve to constantly re-determine the global B_0 (Foerster et al., 2005; Hu and Kim, 1994; Pfeuffer et al., 2002; Splitthoff et al., 2007; Versluis et al., 2010; Ward et al., 2002) or, using a receiver array, a higher-order field model (Splitthoff and Zaitsev, 2009) for data correction. At the image processing stage, varying distortion is partly addressed by co-registration (Andersson et al., 2003, 2001; Ashburner and Friston, 2007; Frackowiak et al., 1995), which is limited, however, to field offsets whose spatial structure matches the distortion model used.

Higher-frequency fields that vary significantly during EPI readouts are almost exclusively driven by gradient operation, with potential contributions from active shimming when performed dynamically (Duerst et al., 2015; Sengupta et al., 2011; van Gelderen et al., 2007). Hardware trade-offs and imperfections give rise to a range of typical errors in these field components. Most prominent among these are the general low-pass behavior of gradient and shim chains, delays, eddy currents, mechanical vibrations, and gain drifts. In single-shot EPI they result in a variety of artifacts, most prominently in ghosting but also in blurring, shearing, and other distortion. When these mechanisms vary over time the related artifacts fluctuate in time series and again become confounds to fMRI. Gradient system imperfection is traditionally addressed by waveform pre-distortion (pre-emphasis) and, for EPI, by data correction based on calibration. The standard calibration approach is to perform additional EPI readouts without phase-encoding blips. Correction settings are then derived from the inconsistencies of the echo train, capturing reproducible imperfections of the frequency-encoding gradient. Such calibration can be performed on a per-scan basis as well as, to sense system changes during a scan, on a per-shot basis by adding calibration echoes at the beginning of each actual EPI readout (Bruder et al., 1992; Hinks et al., 2006; Schmitt et al., 1998).

Navigators and calibration echoes have in common that they rely on NMR signal from the head for field observations. Alternatively, field measurements can also be performed with external NMR sensors, which permit field recording concurrently with image readouts (Barmet et al., 2010, 2009, 2008; De Zanche et al., 2008; Wilm et al., 2011). With this approach, the evolution of B_0 and gradient fields can be captured without requiring additional time or reproducibility of field behavior. Unlike EPI calibration it does not rely on intrinsic repetitiveness of gradient waveforms, permitting field error correction also for, e.g., variable-density EPI and spiral scanning (Kasper et al., 2014; Vannesjo et al., 2016a).

The diversity of types and sources of field perturbations prompts the question which mechanisms dominate in fMRI time series and how large the associated confounds are. Given the different options for addressing field errors it is also important which spatial terms need to be accounted for and at which temporal resolution. A recent study targeted these questions for hardware-related perturbations, performing fMRI scans in a phantom with field monitoring by external sensors (Kasper et al., 2015). In this study relevant variability over time series was observed in both the uniform and gradient field components, exhibiting slow as well as within-shot dynamics. It resulted in image fluctuations ranging between 1% and 10%, depending on spatial order, yet permitted effective retrospective correction using field recordings.

Based on these findings, the goal of the present contribution is to establish how they translate to fMRI in vivo. Specifically, it aims to explore the structure and magnitude of additional field fluctuations of physiological origin, the severity of related image fluctuations, and whether field recording and retrospective correction are equally effective in the in vivo scenario.

2. Methods

Investigation of field fluctuations and their impact on standard fMRI was performed in vivo at 3 Tesla, using the following study design:

- Acquisition of 2D EPI time series in vivo with concurrent field monitoring.
- Extraction of prominent fluctuations of the background field and EPI trajectories using principal component analysis (PCA).
- Spectral separation of physiological field fluctuations from hardware-related perturbations.
- Analysis of image fluctuations caused by field fluctuations of different spatial order and origin.
- Isolation of field-mediated effects from other fluctuations, using simulation.
- Quantification of BOLD sensitivity gained by retrospective field correction.

2.1. Setup

We used the same hardware setup as described in the preceding phantom study (Kasper et al., 2015) to facilitate comparison between phantom and in vivo results. Image data was acquired on a Philips Achieva 3 T system, using an 8-channel head coil array. Field monitoring was performed with an array of 12 transmit/receive field probes (Barmet et al., 2009, 2008; De Zanche et al., 2008) based on ^{19}F NMR for operation concurrent with imaging readouts (Barmet et al., 2010; Wilm et al., 2011). The probe array was mounted on the inside of the head coil as illustrated in Wilm et al. (2015).

2.2. Subjects and imaging protocol

We carried out a total of 18 fMRI sessions, including four healthy subjects (BMI 19–25, two female) after written informed consent and with approval by the local ethics committee. All but one of the subjects underwent three sessions, successively on one day. One subject (subject 1) underwent the three-session protocol repeatedly on three days to examine within-subject variability.