Snapshot field monitoring enables correction of slow field perturbations in high-resolution brain MRI

Signe Johanna Vannesjo[1], Bertram J. Wilm[1], Yolanda Duerst[1], Benjamin E. Dietrich[1], David O. Brunner[1], Christoph Barmet[1,2], Thomas Schmid[1], and Klaas P. Pruessmann[1]

[1]*Institute for Biomedical Engineering, University and ETH Zurich, Zurich, Switzerland,* [2]*Skope Magnetic Resonance Technologies, Zurich, Switzerland*

Introduction: Most MR imaging applications rely on a stable magnetic field during the experiment. Changes in the spatial distribution of tissue susceptibility caused by patient motion, e.g. breathing or movement of extremities, give rise to dynamic field perturbations. These effects scale with the main magnetic field strength and have been reported to affect brain MRI, such as T2*-weighted gradient echo imaging, at 7T [1-3]. It has been shown that artifacts stemming from such drifts can be strongly reduced or eliminated by concurrent field monitoring with NMR field probes [4,5]. However the field evolution mesaurable by a probe is limited by T2* decay as well as gradient dephasing [6]. Gradient dephasing inherently limits the observable maximal value in k-space, and thereby image resolution. If the unknown field changes are relatively slow however - as is the case for breathing-related field perturbations - it can be sufficient to assume a constant field perturbation during each readout. Thus a single field measurement per readout using a short probe acquisition suffices to describe the field perturbation, thereby eliminating the image resolution limitation for field monitoring. In this work, the feasibility of this approach is tested for high-resolution T2*-weighted gradient echo imaging at 7T.

Methods: All measurements were performed on a whole-body 7T Philips Achieva system. T2*-weighted gradient-echo images were acquired in the transverse plane (FOV 230x230 mm^2, TR 800 ms, TE 25 ms, slice thickness 2 mm) on a phantom (0.3 mm in-plane resolution) and in two healthy volunteers (0.75 mm and 0.3 mm resolution, resp.) in a caudal and cranial region of the brain, respectively. Image data acquisition was performed using a 32-channel head coil (Nova Medical, Wilmington, USA). For the in-vivo scans, the volunteers were instructed to breathe deeply throughout the scan. In the phantom experiments a bottle filled with mineral oil was shifted back and forth along the z-axis at breathing frequency, to simulate breathing-related field effects. The resulting field fluctuations were measured with 16 ^{19}F NMR probes (diameter 1.8 mm, T2 30 ms) placed on the outside of the receive coil [5,6]. The probes were excited at 12.5 ms after phantom or brain excitation, respectively, for each gradient echo line. After digitization on a custom spectrometer [7], the field in each probe was calculated by a linear fit of the probe's phase evolution during 2 ms acquisition. Using the field information of 16 probes, a 0th-3rd order spherical harmonic field model was fitted for each acquisition. For image reconstruction, the nominal k-space trajectory was calculated from the imaging parameters. The measured fields were translated into linear phase offsets in the effective phase encoding. Images were reconstructed as a least squares solution for the calculated encoding, including the 0th- 3rd-order phase evolutions, using a conjugate-gradient based iterative algorithm [8,9]. For comparison, images were also reconstructed with nominal encoding only.

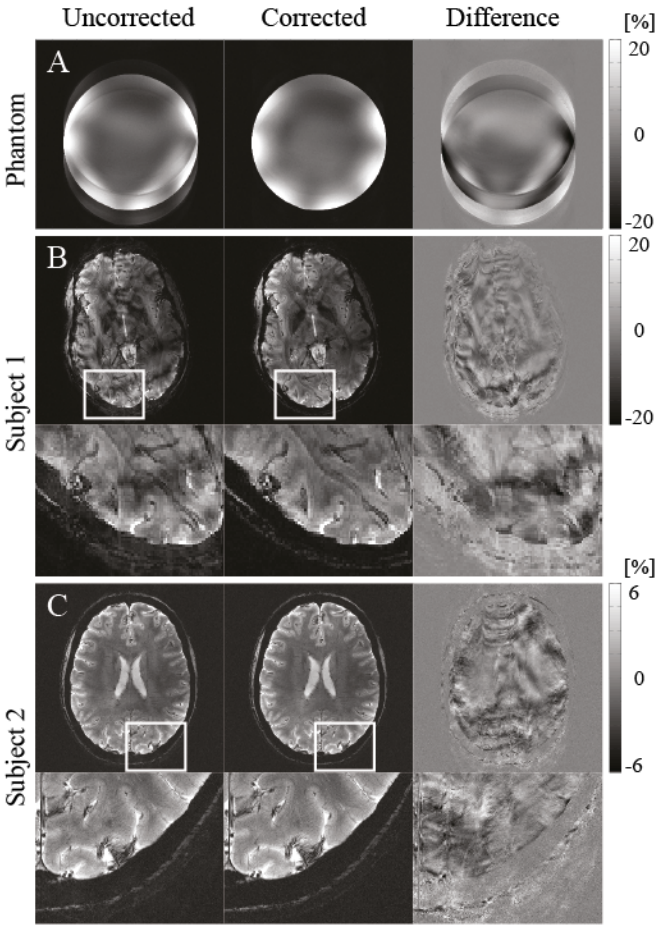

Fig 2: Images reconstructed with the nominal encoding only (uncorrected); including the measured 3rd-order fields (corrected); and the difference between the reconstructions (in % of image max).

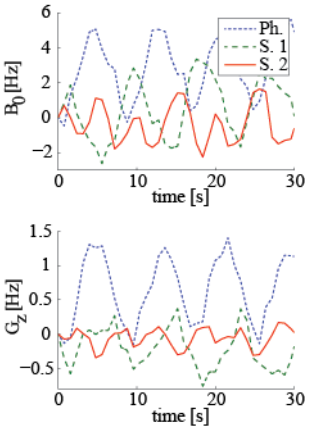

Fig 1: Measured field fluctuations in the 0th-order (top) and in the z gradient (bottom), scaled to Hz at the position of the respective slice.

Results and Discussion: A selection of the measured fields in the 0th-order (B_0) and in the linear z field term (G_z) is shown in Fig 1. Despite the strong field perturbations induced during the phantom experiments, good image quality of the field monitored reconstructions was obtained (Fig 2A), with minimal residual ghosting in the corrected images. In the phantom, a relative difference of up to 26% of the maximum image magnitude between uncorrected and corrected images is observed. In the uncorrected caudal in vivo image (subject 1), strong ghosting artifacts and intensity modulations over the image are apparent. The artifacts can be well corrected when using the monitored field data (Fig 2B), with a difference between the reconstructions of up to 22%. For the more cranial slice (subject 2) (Fig 2C), the artifacts are more subtle, which can be expected due to the greater distance to the thorax. However also in this case, incorporating the field measurements in the reconstruction yields visibly less ghosting and more homogenous signal intensity within the brain, with a difference between the reconstructions of up to 6%.

Conclusion: We introduced a variant of field monitoring which can be used for the correction of slow field perturbations that can be assumed to be constant during each encoding step. Unlike full k-space monitoring, the approach is not limited by signal decay of the field probes and thus is applicable also to high-resolution imaging. The method was successfully applied for the correction of a commonly used T2*-weighted gradient-echo sequence at 7T, which otherwise can be subject to strong breathing-related artifacts.

References: [1] van de Moortele et al, MRM 2002;47:888-895 [2] van Gelderen et al, MRM 2007;57:362-368 [3] Versluis et al, NeuroImage 2010;51:1082-1088 [4] Vannesjo et al, Proc. ISMRM 2012; p.216 [5] Barmet et al, Proc. ISMRM 2010; p.216 [6] Barmet et al, MRM 2008;60:187-197 [7] Dietrich et al, Proc. ISMRM 2012; p.700 [8] Pruessmann et al, MRM 2001;46:638-651 [9] Wilm et al, MRM 2001;65:1690-1701