Magnetic Field Monitoring for Improved Phase Contrast Flow Quantification

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INTRODUCTION:

Magnetic resonance imaging provides the unique potential to quantitatively measure coherent (flow) as well as incoherent (diffusion) motion information. Therefore, strong gradient pre-pulses are incorporated into the pulse sequence to encode motion information either into the phase (flow), or the amplitude (diffusion) of the detected signals. The associated intense gradient switching, however, induces eddy currents, which can cause significant encoding errors in the subsequent data acquisition part of the pulse sequence. For the case of quantitative flow imaging this problem is known to cause significant errors with the consequence of potentially misleading interpretation [1, 2]. The problem of eddy current artifacts in motion encoding exists on well-calibrated modern MR systems with actively shielded gradients and gradient waveform pre-emphasis.

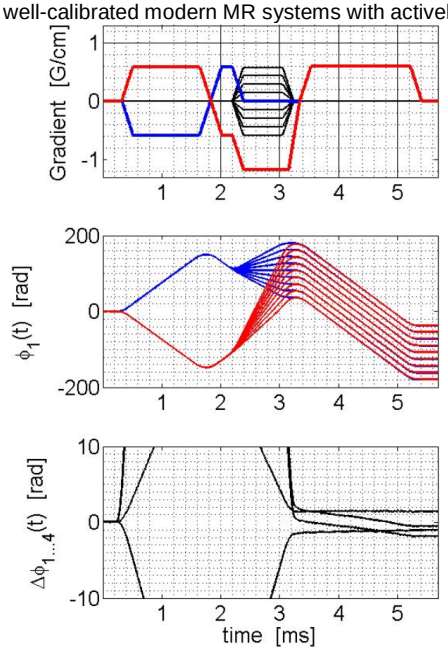

Fig. 1: MFM phase evolution measurement for 9 (/128) PC phase encoding steps: red positive polarity, blue negative polarity.

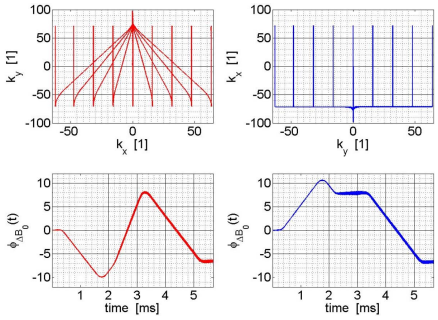

Fig. 2: MFM-measured k-space trajectory and phase offset for 9 (/128) phase encoding steps: red pos. polarity, blue neg. polarity.

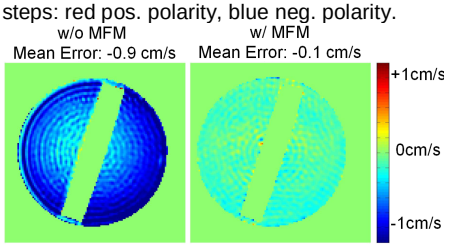

Fig. 3: Significant improved PC flow quantification using MFM (right).

In the presented work a Magnetic Field Monitoring (MFM) approach, similar to ones described by Mason et al [3] and more recently by Pruessmann et al [4], is used for real-time monitoring of the actual (physical) magnetic fields applied during the execution of the pulse sequence. In the subsequent post-processing the obtained information is then used to correct for imperfections caused by Eddy currents and concomitant field effects.

METHODS:

A MFM hardware setup has been developed based on small NMR probes acting as local magnetic field sensors [5, 6]. The probes consist of a small water droplet (~ 1mm) closely surrounded by a solenoid coil for high SNR signal detection. Susceptibility-matching methods [7] have been applied, in order to minimize T2*-induced signal decay and hence maximizing MFM tracking times. Four NMR probes are placed in close proximity to the imaging object. Time-resolved image encoding information can be extracted from the NMR probe's phase evolution $\phi_n(t)$ according to:

$$\phi_n(t) = \varphi_{n,offset} + \varphi_{\Delta B0}(t) + \mathbf{r}_n\,\mathbf{k}(t) + O(\mathbf{r}_n^{\,2}), \quad \varphi_{\Delta B0}(t) = \gamma\int_0^t \Delta B_0(\mathbf{r},t')dt', \quad \mathbf{k}(t) = \gamma\int_0^t \mathbf{G}(\mathbf{r},t')dt' \qquad (1)$$

with $\Delta B_0(t)$ and $\mathbf{G}(t)$ the main magnetic field offset and gradient field, respectively. The NMR probe's spatial position $\mathbf{r}_n$, as well as their time-constant phase offset $\phi_{n,offset}$ are obtained from fast single FID-based calibration scans. Gridding image reconstruction is then performed based on the MFM measured phase variation $\varphi_{\Delta B0}(t)$ and the k-space trajectory $\mathbf{k}(t)$ according to [8]:

$$\text{image}(\mathbf{r}) = \int \text{data}(\mathbf{k})\, w(\mathbf{k})\, e^{-i\varphi_{\Delta B0}(\mathbf{k})}\, e^{-i\mathbf{k}\mathbf{r}}\, d\mathbf{k} \qquad (2)$$

with w(**k**) the k-space density compensation function. In this way encoding errors caused by any kind of magnetic field imperfections (concomitant field, Eddy currents, etc.) are intrinsically accounted for.

A gradient echo based phase-contrast sequence has been designed for a nominal VENC of 50 cm/s. The flow encoding bi-polar gradient was applied along the frequency encoding direction and was combined with the frequency encoding prewinder gradient (cf. Fig. 1).

All imaging experiments were performed on a GE Signa Excite 3T system (GE Healthcare, Milwaukee, WI) using an 8-channel head array coil and a cylindrical water-filled phantom.

RESULTS and DISCUSSION:

Magnetic field monitored PC imaging has been performed for the pulse sequence shown in Fig. 1. The middle plot shows measured phase evolutions for one NMR probe $\phi_1(t)$ and both polarities of the bipolar gradient. The bottom plot illustrates the phase difference $\Delta\phi_n(t)$ between the positive and the negative polarity for all four NMR probes utilized. Ideally, the phase difference $\Delta\phi(t)$ after the application of the bi-polar gradients should approach zero again. The appearance of residual phase offsets and variations for t > 3.4 ms indicates magnetic field imperfections in form of Eddy currents and concomitant field effects.

Figure 2 illustrates the real-time magnetic field monitored phase variation $\varphi_{\Delta B0}(t)$ and k-space trajectory **k**(t) obtained using Eq. (1). Quantitative flow maps were calculated based on standard phase difference processing. Figure 3 compares the quantitative flow map obtained without MFM (images reconstructed based on the nominal (prescribed) gradient waveforms) with the one obtained using MFM (images reconstructed according to Eq. (2)). For the considered static water phantom, MFM clearly demonstrates a significantly improved PC flow quantification by reducing the mean flow error from -0.9 cm/s to -0.1 cm/s. The flow profile obtained using MFM is approximately clean of spatially constant and linear errors and thereby consistent with the spatially linear model described by Eq. [1]. Further improvement can be expected from using more magnetic field sensors together with a spatially higher-order image encoding model (cf. Eqs. (1-2)).

Alternatively to the proposed correction method based on MFM monitored gridding reconstruction (2), a simpler approximate phase correction scheme might be applied. To this end a net phase correction map is estimated at the echo time only and subtracted from the corresponding standard reconstructed images. This approximate phase correction has the advantage of allowing a computationally more efficient incorporation of spatially higher-order field imperfections.

REFERENCES:

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