Addressing phase errors in quantitative water-fat imaging at 3 T using a time-interleaved multi-echo gradient-echo acquisition

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Target audience: Basic scientists and clinical researchers working on water-fat imaging

Purpose: Chemical shift encoding-based water-fat imaging using multi-echo gradient-echo sequences has been emerging for non-invasive organ fat quantification [1] using either complex-based [2] or magnitude-based [3] techniques. Although complex-based techniques have shown superior noise performance compared to magnitude-based techniques [4], complex-based techniques suffer from different types of phase errors, which can induce significant fat fraction (FF) bias [5-9]. Phase errors have been previously addressed by using hybrid methods combining complex- and magnitude-based fitting [5,6], by using reference data [7] or by modeling the phase error terms [8,9]. Most of this work has focused on single-shot acquisitions (where all echoes are acquired in a single TR) using either a fly-back readout (at 1.5 T [5, 6]) or a non-fly-back readout (at 3 T [9]). Time-interleaved multi-echo gradient-echo (TIMGE) acquisitions (where echoes are acquired in multiple TRs) are highly desirable at 3 T, as they enable high spatial resolution while maintaining short echo time steps. However, their phase error behavior has not been previously addressed. The present study aims to address and correct phase error effects in TIMGE acquisitions.

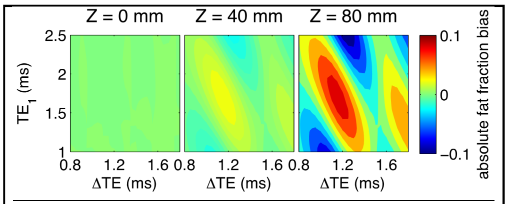

Fig. 1: Simulated absolute FF bias due to concomitant dephasing for different slice off-center locations Z. The estimated FF bias depends on TE$_1$, ΔTE and Z.

Methods: Phase error decomposition & correction: A typical transverse 3D TIMGE sequence was considered acquiring in total six echoes with constant echo spacing (ΔTE) in two interleaves (three echoes per interleave) using fly-back gradients. Phase errors include terms that can be corrected using a reference scan with reverse gradient readout polarity ($\Delta\Phi_R$). Remaining phase errors that are not corrected by reversing the gradient readout polarity include spatially varying terms due to concomitant gradients ($\Delta\Phi_C$) and a spatially constant interleave offset ($\Delta\Phi_{IO}$). Therefore the total phase error is $\Delta\Phi_{total} = \Delta\Phi_R + \Delta\Phi_C + \Delta\Phi_{IO}$. A preparation phase module was performed before the actual sequence, acquiring the echoes without phase encoding once with the original and once with flipped readout gradient polarity. The information of the phase difference was used to perform a 1D phase correction on the acquired data and to remove the $\Delta\Phi_R$ phase term. $\Delta\Phi_C$ was calculated based on the gradient waveforms according to [10]: $\Delta\Phi_C(x,y,z) = A z^2 + B (x^2 + y^2) + C xz + D yz$ with x, y and z being the physical coordinate system of the scanner, while A, B, C and D are calculated scalars representing the spatially varying dephasing. $\Delta\Phi_{IO}$ was estimated based on down-sampled data, which has already been corrected for $\Delta\Phi_R$ and $\Delta\Phi_C$ effects, using a formulation similar to [8,9] fitting for a constant phase offset for the second interleave.

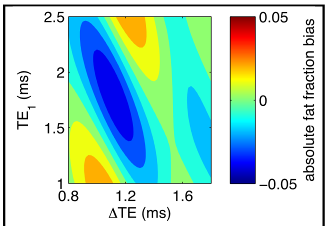

Fig. 2: Simulated absolute FF bias due to constant phase shift between the two interleaves of 0.05 rad for a nominal FF of 0 %.

Simulations: a) The influence of the concomitant field phase error on FF estimation was simulated for 0% nominal FF as a function of TE$_1$ and ΔTE using the described sequence. Concomitant coefficients for a transverse scan were estimated based on the gradient waveforms in the readout direction (L/R) only (B = C = D = 0, $\Delta\Phi_R = \Delta\Phi_{IO} = 0$). b) The influence of a 0.05 rad phase offset between the echoes of the two interleaves was simulated on FF estimation. ($\Delta\Phi_R = \Delta\Phi_C = 0$)

MRI measurements: A water phantom with 0 % FF was scanned on a 3 T Ingenia (Philips, Best, Netherlands) scanner using an 8-channel extremity coil using the described sequence at different resolutions in the frequency encoding direction with Δx of 1.0 / 1.1 / 1.2 / 1.3 / 1.45 mm leading to ΔTEs of 1.34 / 1.25 / 1.21 / 1.10 / 1.01 ms, respectively. All other parameters were kept constant: TE$_1$ = 1.8 ms, TR = 14 ms, FOV = 140 x 140 x 220 mm, voxel size = Δx x 2 x 2 mm, flip angle = 20°. A liver scan of a healthy volunteer was performed using posterior and anterior coil arrays with TE$_1$ = 1.57 ms, ΔTE = 1.21 ms, TR = 9 ms, flip angle = 5°, voxel size = 1.25 x 2.5 x 6 mm, FOV 400 x 240 x 192 mm. Noise bias effects in FF were corrected using the magnitude discrimination approach [11]. A single-voxel STEAM MRS was performed with TE = 10/15/20/25 ms (used for T_2 correction), TM 16 ms, TR = 3.5s, VOI = 30 x 30 x 20 mm, sweep width = 5000 Hz (single breath hold).

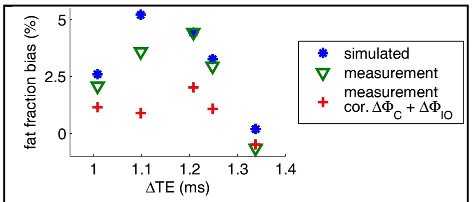

Fig. 3: FF bias measured correcting for $\Delta\Phi_R$ (green triangle), and $\Delta\Phi_R + \Delta\Phi_C + \Delta\Phi_{IO}$ (red cross) in the phantom experiment with ΔTEs of 1.34 / 1.25 / 1.21 / 1.10 / 1.01 ms and TE$_1$ = 1.8 ms at Z = 67 mm off-center. The blue asterisk shows simulated FF bias assuming that only $\Delta\Phi_R$ is corrected.

Results: The simulation in Fig. 1 shows that the absolute FF bias due to the concomitant field phase error varies for the described sequence depending on TE$_1$, ΔTE and Z off-center. At Z = 80 mm an absolute FF bias up to ~10 % can be observed for specific TE$_1$ and ΔTE. A small systematic phase offset $\Delta\Phi_{IO}$ between the two interleaves also results in an underestimation of the FF of 5% at certain TE$_1$ / ΔTE combinations. (Fig. 2) The phantom experiment shows a FF bias in accordance with the simulation if $\Delta\Phi_C + \Delta\Phi_{IO}$ were not corrected (green triangle and blue asterisk in Fig. 3). Correcting for all three phase factors shows smallest FF bias. (Fig. 3) Fig. 4 shows the *in vivo* and one phantom result after the different phase correction steps. The MRS T_2-corrected liver FF was equal to 1 %.

Discussion & Conclusion: The present results show that the concomitant field phase error and interleave offset have a non-linear effect on complex-based water-fat separation and have to be considered especially for off-center slices. The tendency of the predicted FF bias from simulation could be confirmed in the phantom experiment. The MRS-determined liver FF of 1% is also in agreement with the imaging FF after correction of all phase errors. **In conclusion, complex-based water-fat separation using TIMGE has to consider additional phase error effects that cannot be corrected using only a reference scan with reversed gradient readout polarity.**

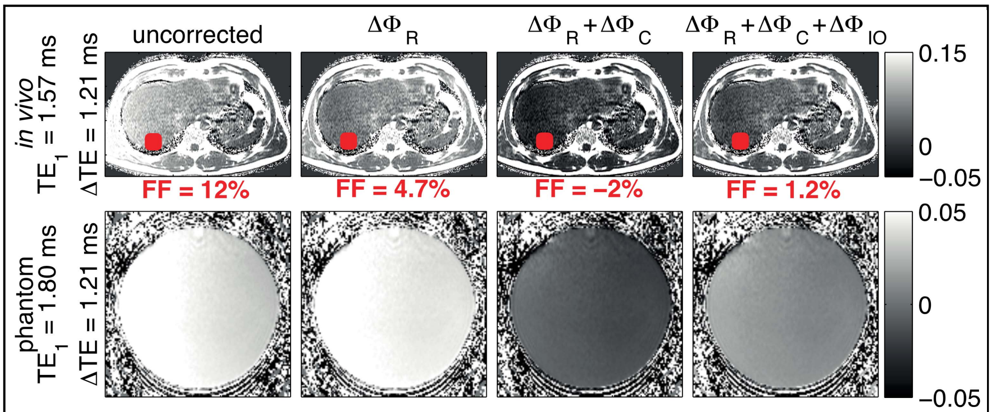

Fig. 4: FF maps: *In vivo* results (first row) at Z = 75 mm (TE$_1$ = 1.57 ms / ΔTE = 1.21 ms) and phantom results (second row) at Z = 61 mm (TE$_1$ = 1.08 ms / ΔTE = 1.21) ms. Columns show the result after given phase corrections. The given fat fractions in red match the red VOI of the performed MRS (FF = 1 %). The $\Delta\Phi_R$ term eliminates a linear FF bias in the frequency encoding direction. The $\Delta\Phi_C$ and $\Delta\Phi_{IO}$ terms primarily show a constant in-plane behavior.

References: [1] Reeder, J Magn Reson Imag 34:729, 2011, [2] Yu, Magn Reson Med 60:1122, 2008, [3] Bydder, Magn Reson Imag 26:347, 2008, [4] Hernando, Magn Reson Med 64:811, 2010, [5] Yu, Magn Reson Med 66:199, 2011, [6] Hernando, Magn Reson Med 67:638, 2012, [7] Yu, J Magn Reson Imag 31:1264, 2010, [8] Eggers, Proc. ISMRM 2008, p. 1364, [9] Peterson, Magn Reson Med 71:219, 2014, [10] Bernstein, Magn Reson Med 38:300, 1998, [11] Liu, Magn Reson Med 58:354, 2007.

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