Dual-Echo Dixon Imaging with Unrestricted Choice of Echo Times

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Introduction

Two-point Dixon methods are particularly attractive in rapid imaging applications. However, they commonly restrict the choice of echo times by requiring at least one of them being in phase [1,2]. This compromises their scan efficiency and renders them sometimes even slower than three-point Dixon methods [3]. In this work, a novel two-point Dixon method is described that basically eliminates all constraints on the echo times, and its performance is compared to that of existing two-point Dixon methods in abdominal imaging.

Methods

The composite signal S in image space at echo time TE is modeled by

$$S_n = \left(W + F\mathrm{e}^{\mathrm{i}\theta_n}\right)\mathrm{e}^{\mathrm{i}\varphi_n},$$

where W and F are the water and fat signal in image space, θ is the dephasing angle between them, and φ is a common phase. First, potential values of the phasor that corresponds to the phase error $\Delta\varphi := \varphi_2 - \varphi_1$ are calculated. For this purpose, a major and a minor signal component are derived from the magnitude of S at the two echo times [2]

$$M_{1/2} = \frac{1}{2}\left[\sqrt{\frac{|S_1|^2(\cos\theta_2-1)-|S_2|^2(\cos\theta_1-1)}{\cos\theta_2-\cos\theta_1}} \pm \sqrt{\frac{|S_1|^2(\cos\theta_2+1)-|S_2|^2(\cos\theta_1+1)}{\cos\theta_2-\cos\theta_1}}\right],$$

and the conjugate complex product of the signal equation for the two echo times is solved, yielding

$$\mathrm{e}^{\mathrm{i}\Delta\varphi_{1/2}} = \frac{S_1^* S_2}{\left(M_{1/2}+M_{2/1}\mathrm{e}^{-\mathrm{i}\theta_1}\right)\left(M_{1/2}+M_{2/1}\mathrm{e}^{\mathrm{i}\theta_2}\right)}\,.$$

Then, one of these values is selected based on the assumption of spatial smoothness of the phasor. Any of a number of existing strategies may be applied for this purpose [1,2,4]. Finally, W and F are re-estimated given S_1, S_2, and the phasor. Either they are considered as real and the system of equations

$$\begin{aligned} W^2 + F^2 + 2WF\cos\theta_{1/2} &= |S_{1/2}|^2, \\ W^2 + F^2\mathrm{e}^{\mathrm{i}\Delta\theta} + WF\left(\mathrm{e}^{\mathrm{i}\theta_2} + \mathrm{e}^{-\mathrm{i}\theta_1}\right) &= S_1^* S_2 \mathrm{e}^{-\mathrm{i}\Delta\varphi}, \end{aligned}$$

is solved, or they are considered as complex and the two signal equations are solved for $W' = W\mathrm{e}^{\mathrm{i}\varphi_1}$ and $F' = F\mathrm{e}^{\mathrm{i}\varphi_1}$.

Abdominal imaging on volunteers was performed on 1.5 T and 3.0 T scanners (Philips Healthcare, Best, Netherlands) with 16 and 32 element receive coils and a 3D spoiled multi-gradient-echo sequence. Typical protocol parameters included a coverage of 370 x 260 x 240 mm^3, a slice thickness of 3 mm, and a flip angle of 10°. Scans were completed in single breathholds in less than 20 s.

Results

The relative noise in water images produced with the proposed method is quantified for a range of echo times in Fig. 1. The shown simulated values reflect the worst case, in which no spatial smoothing of the phasor is performed, for a selected water-fat ratio. They decrease towards shorter and longer echo spacings, as known from existing methods [2], and additionally close to the two diagonals defined by $\Theta_1 + \Theta_2 \bmod 2\pi = 0$.

In Fig. 2, the proposed method (lower row) is compared to existing methods (upper row) [1,2] at 1.5 T. Given the minimum TR for an in phase echo time, it allows the use of lower bandwidths and thus provides up to 60% more SNR in this case. Alternatively, it permits the use of shorter TRs and thus accelerates scans up to 1.6-fold in this case (1.3-fold in the shown example).

The freedom of choice is exploited to enhance the spatial resolution at 3.0 T in Fig. 3. While there is no significant penalty for an in phase echo time at 1.5 mm, it gets substantial at 1.0 mm (20% longer TR) and even prohibitive (70% longer TR) if additionally an out phase echo time is demanded [1].

Conclusions

Removing the restrictions on the choice of echo times that are imposed by existing two-point Dixon methods provides more flexibility in the selection of protocol parameters and thus enables shorter scan times, higher spatial resolution, and increased SNR in rapid imaging applications.

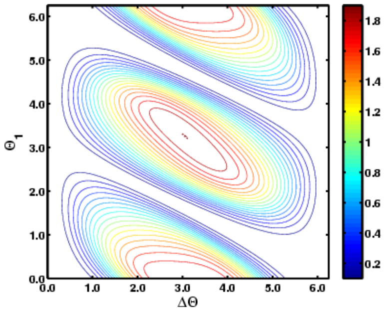
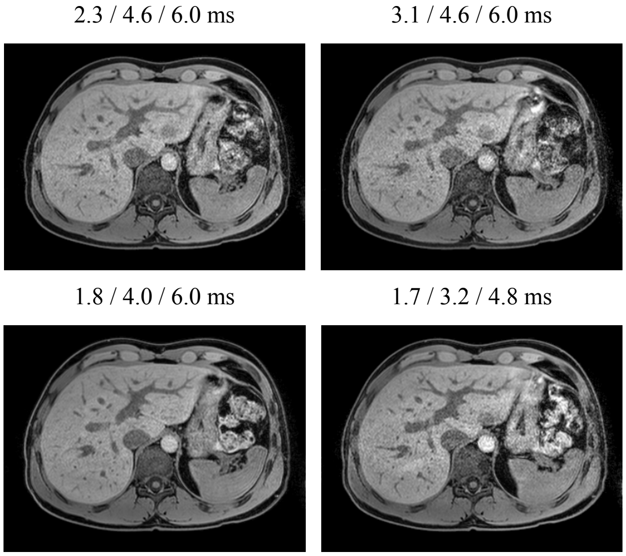


Fig. 1. Noise propagation in the separation. Plotted is the effective number of signal averages as function of the dephasing angle Θ_1 at the first echo time and the increment in the dephasing angle $\Delta\Theta$ between the two echo times.

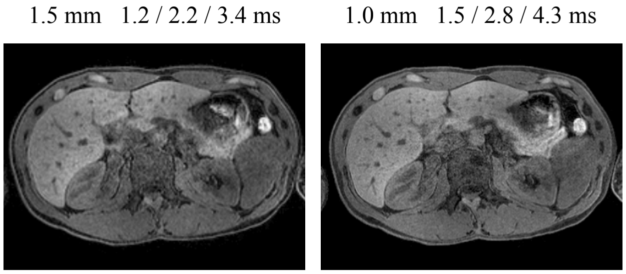

2.3 / 4.6 / 6.0 ms

3.1 / 4.6 / 6.0 ms

1.8 / 4.0 / 6.0 ms

1.7 / 3.2 / 4.8 ms

Fig. 2. Comparison of water images produced from different two-point acquisitions at 1.5 T with corresponding methods. Stated are the two echo times and the repetition time.


1.5 mm 1.2 / 2.2 / 3.4 ms

1.0 mm 1.5 / 2.8 / 4.3 ms

Fig. 3. Water images produced from different two-point acquisitions at 3.0 T with the proposed method. Stated are the spatial resolution, the two echo times, and the repetition time.

References

1. Ma J. Magn Reson Med 2004; 52:415-419 2. Xiang QS. Magn Reson Med 2006; 56:572-584. 3. Eggers H, et al. Proc ISMRM 2009; 2705. 4. Yu H, et al. Magn Reson Med 2005; 54:1032-1039.