

Influence of Spectral Model and Signal Decay on Hepatic Fat Fraction Measurements at 3 T with Dual-Echo Dixon Imaging

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Introduction

Methods to quantitatively assess fatty infiltration of the liver with MRI have received considerable interest lately, because they promise a superior diagnosis and monitoring of hepatic steatosis [1]. Gradient-echo sequences with chemical shift encoding have proven to allow a rapid and robust mapping of water and fat, but the accuracy of derived fat fractions is affected by various confounding factors, such as the spectral composition of fat and transverse relaxation [2,3]. While the acquisition of more echoes generally increases the susceptibility to these effects, it also facilitates the calibration of suitable corrections [4]. At the same time, it entails inserting a multi-echo scan into clinical examinations, which currently include dual-echo scans only, and compromising on resolution and coverage due to breath-hold limitations. Therefore, the influence of the spectral model and of signal decay on fat fractions estimated from dual-echo imaging is studied in this work, and a comparison with six-echo imaging is made.

Methods

An opposed-phase (OP) first and an in-phase (IP) second echo time are conventionally selected in dual-echo imaging. The fat fraction f is, for $f < 0.5$, estimated by

$$f = (S_{IP} - S_{OP})/(2S_{IP}),$$

where S_{OP} and S_{IP} denote the measured signal magnitude. Assuming a multi-peak spectral model of fat predicting relative amplitudes of a pure fat signal of $c_{OP}, c_{IP} \leq 1$, OP and IP echo times are slightly different, and f is more correctly determined by

$$f = (S_{IP} - S_{OP})/((1 + c_{OP})S_{IP} - (1 - c_{IP})S_{OP}).$$

The bias introduced by using a single- instead of a seven-peak spectral model is plotted in Fig. 1 [5]. In the relevant range of fat fractions, a systematic underestimation is seen for both choices of OP and IP echo times, which can be minimized by selecting more flexible TEs, such as 1.4/2.6 ms [6]. Common T_2^* decay of water and fat signals has a similar effect as the spectral composition of fat at short echo times, but the bias is larger for smaller fat fractions, as evident from Fig. 2.

Imaging was performed on patients in single breath-holds on a 3 T scanner (Philips Healthcare, Best, The Netherlands) with a 16-element receive coil and a 3D T_1 -weighted multi-gradient-echo sequence [7]. Typically, $TE_1/\Delta TE/TR$ were 1.4/1.1/3.7 ms and 1.3/0.9/7.1 ms, the resolution was 1.5 mm and 2.3 mm, and the coverage in SI direction was 350 mm and 250 mm for two and six echoes, respectively.

The water-fat separation optionally included a multi-peak spectral model and transverse relaxation in the signal model. For a quantitative analysis, a region of interest of about $50 \times 50 \text{ mm}^2$ was manually defined in the right lobe of the liver in a selected slice. Vasculature was excluded by histogram-based thresholding of the water signal, and bias from noise was minimized by averaging of the complex water and fat signals before deriving fat fractions.

Results

Representative fat fraction maps from a selected examination are shown in Fig. 3. The fat fraction increases substantially from 27.4% to 34.7% with the use of a multi-peak spectral model for six echoes, but only slightly from 33.1% to 33.5% for two echoes in this example. Considering the expected underestimation of the fat fraction due to T_2^* decay, the results obtained with a multi-peak spectral model correspond well. Moreover, the comparatively minor difference between the results for two echoes agrees with the predicted small bias at the chosen echo times.

Discussion and Conclusion

Using a precalibrated spectral model, the bias in fat fractions from the spectral composition of fat can be removed in dual-echo imaging. In addition, the bias from T_2^* decay can be reduced by assuming a typical T_2^* value for liver tissue. Considering both, results from dual- and six-echo imaging seem to be comparable for fatty infiltration of the liver. Furthermore, a detection, or even a crude quantification, of T_2^* is in principle possible in dual-echo imaging in the absence of fat. Clinically, shorter T_2^* values for liver tissue are consistent with other disease processes, such as hemochromatosis, in which fat is generally not present in significant quantities. Consequently, a separate multi-echo scan may only be required in case of deposition of both fat and iron in the liver. The decision on its inclusion in a clinical examination may be based on a dual-echo scan, employing other image analysis methods [8].

References

1. Lee SS, et al. J Hepatol 2010; 52:579-585.
2. Hussain HK, et al. Radiology 2005; 237:1048-1055.
3. Reeder SB, et al. J Magn Reson Imaging 2009; 29:1332-1339.
4. Yu H, et al. Magn Reson Med 2008; 60:1122-1134.
5. Ren J, et al. J Lipid Res 2008; 49:2055-2062.
6. Eggers H, et al. Proc ISMRM 2010; 770.
7. Perkins TG, et al. Proc ISMRM 2010; 556.
8. Bashir MR, et al. Proc ISMRM 2010; 554.

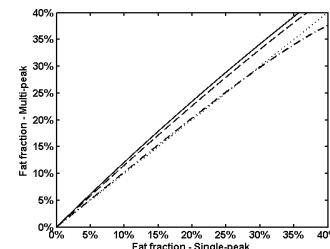


Fig. 1. Bias in fat fractions derived from dual-echo imaging at 3 T with ideal (solid) and actual (dashed) opposed- and in-phase echo times, and with flexible echo times (dashdotted), using a single-peak spectral model. The line of unity (dotted) is shown for reference.

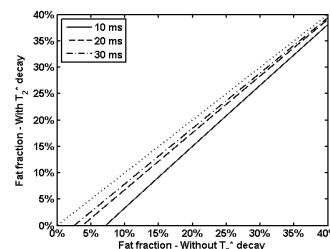


Fig. 2. Bias in fat fractions derived from dual-echo imaging at 3 T with actual opposed- and in-phase echo times, ignoring T_2^* decay. The line of unity (dotted) is shown for reference.

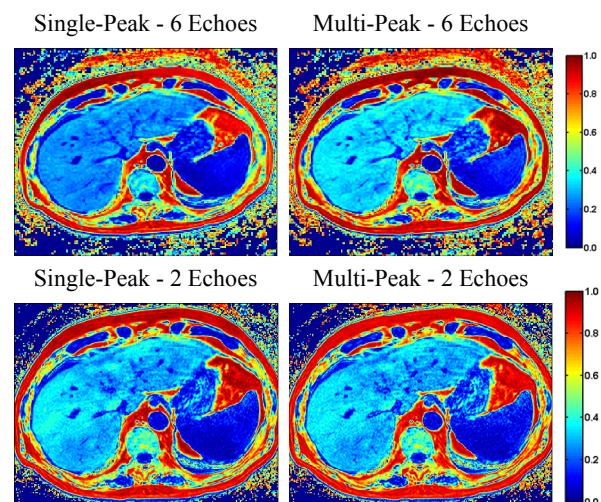


Fig. 3. Fat fraction maps estimated from six- and dual-echo imaging, using a single- and a multi-peak spectral model of fat in the separation.