Standard compared to optimized mDIXON liver fat fraction imaging

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Purpose With liver steatosis rapidly becoming epidemic, there is need of quantitative, non-invasive fat measurement techniques [1, 2]. While ^{1}H-MR Spectroscopy (^{1}H-MRS) is considered the reference standard, it lacks spatial coverage and is only accurate and reproducible in the hands of experienced spectroscopists. Whole liver, parametric fatmaps resolve this problem and cover the entire liver, even allowing segmental calculation of fat fractions [2]. Recently, a modified Dixon scheme (mDIXON) with flexible TEs was introduced, allowing high resolution 3D isotropic water- and fat-only images to be obtained in a single breath hold [3, 4]. However, preliminary reports have shown a bias in the fat fractions obtained with mDIXON, with a 4% lower limit for a reference ^{1}H-MRS value of 0% [5]. It is unclear whether this lower limit is due to T1-weighting, the use of only two echoes or limitations in the reconstruction module. In this study we compared standard mDIXON ($mDIX_{STAND}$) with an optimized (i.e. longer TR and deliberately chosen asymmetric TEs) scan protocol ($mDIX_{OPT}$) in patients and phantoms to improve accuracy at low fat fractions.

Methods: Fourteen subjects were recruited with informed consent in a board approved study in non-alcoholic fatty liver disease (NAFLD) and non-alcoholic steatohepatitis (NASH) patients. Fat phantoms were constructed according to Hines et al. ranging from 0 to 60% fat [6]. Examinations were performed at 3.0T (Ingenia, Philips Healthcare, Best, The Netherlands). Single voxel (20×20×20 mm3) multi-echo STEAM ^{1}H-MRS was performed with TR/TE1/ΔTE of 3500/10/5ms and either 5 (in vivo) or 10 (phantom) echoes. Post-processing was performed as detailed by Yokoo et al [7]. mDIXON (W/IP/OP/F) images were obtained for $mDIX_{STAND}$ with a 3D fast gradient echo acquisition (FA=10,TR=5.4ms, TE1/ΔTE=2.11/1.0ms, 3 echoes, FOV=360×306×192mm, acq. matrix=240x203x48, recon. matrix=384×384×96, acq. time = 17s) and for $mDIX_{OPT}$ using an optimized 2D fast gradient echo (FA=5, TR=100ms, TE1/ΔTE=2.11/0.76ms with 3 (in vivo) or 4 (phantoms) echoes, FOV=384×304×180mm, acq. matrix=192×152×18, recon. matrix=192×152×18, acq. time = 19.8s). Fat fractions maps were calculated voxel-by-voxel from the reconstructions (F/F+W). ^{1}H-MRS voxel positions were automatically co-localized to the correct anatomical position on the fat fraction maps for proper comparison using a home-written MATLAB script. Correlations between ^{1}H-MRS (as reference) and mDIXON values were assessed with Spearman's Correlation Coefficients and linear regression lines.

Results: Examples of W/F/Fatmaps of the phantom and in vivo measurements are shown in Fig. 1A-E. As shown in Fig. 2A-D, $mDIX_{STAND}$ correlated (almost) perfectly with ^{1}H-MRS values in both phantoms (R_S:1.0,P<0.001) and humans (R_S:0.97,P<0.001), but clearly overestimated fat content showing non-zero intercepts. $mDIX_{OPT}$ had a similarly good correlation with ^{1}H-MRS values. Moreover, in phantoms using $mDIX_{OPT}$ resolved the overestimation with the intercept changing from 6.69 (95%-CI: 4.6 - 8.8) to 0.67 (95%-CI:-0.26 - 1.6). However, for in vivo liver measurements the overestimation was still present (compare the marked intercepts in Figs 2B and 2D).

Discussion and Conclusion: $mDIX_{STAND}$ is a fast 3D isotropic method capable of generating high resolution water-only, in/out of phase and fat-only images in a single breath hold. Fat-fraction maps showed excellent correlations with ^{1}H-MRS determined true fat fractions but especially low fractions (near the 5.6% threshold for diagnosing steatosis) tend to be overestimated [9]. This overestimation of fat fractions was resolved using the optimized $mDIX_{OPT}$ scan protocol but in phantoms only and not for in vivo liver measurements. Differing T1-values of tissue and phantom material may be one of the causes for this finding. Given its larger spatial coverage and isotropic and higher resolution, the standard mDIXON protocol appears to be preferable over the theoretically optimized protocol.

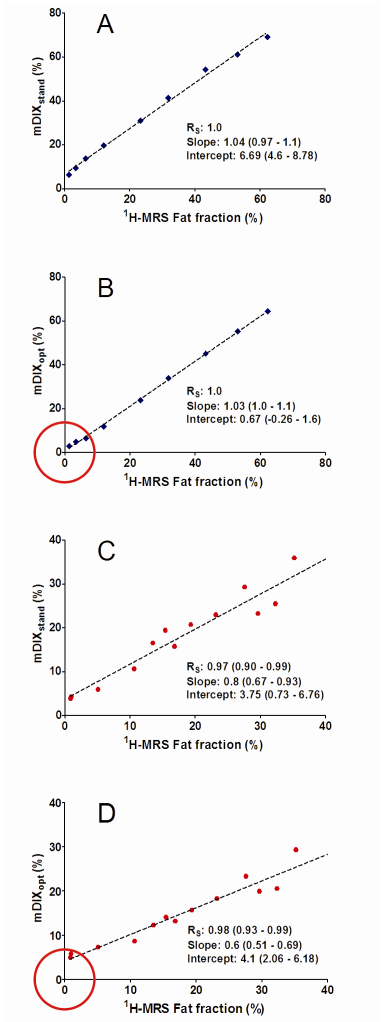

Fig. 2A-D. Scatter plots of ^{1}H-MRS derived fat fractions (x-axes) and mDIXON derived fat fractions (y-axes) for phantom (A-B) and in vivo measurements (C-D).

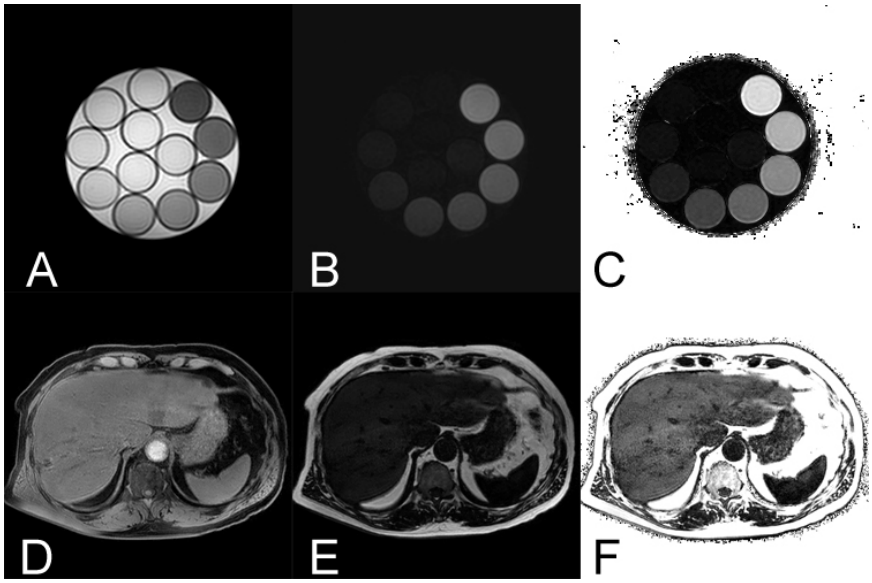

Fig. 2A-D. Signal decay versus TE for PDEs (2A-B) and PMEs (2C-D).

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