Fat quantification by modeling the variation in signal amplitude with TE

M. Bydder[1], M. S. Middleton[1], P. D. Gatehouse[2], A. D. Chavez[1], C. B. Sirlin[1]

[1]Radiology, University of California, San Diego, CA, United States, [2]Cardiac Magnetic Resonance Unit, Royal Brompton Hospital, London, United Kingdom

Introduction

In magnetic resonance imaging (MRI), the signals from fat and water exhibit characteristic temporal variations that can be separated using several measurements taken at different time-points following excitation (TE). The temporal variation comprises phase interference between the two tissues as well as the T2* decay of the individual tissues. The classic Dixon technique uses two images acquired at specific TEs, namely "in phase" (IP) and "opposed phase" (OP), which are subsequently combined to create the water image (IP+OP)/2 and the fat (IP-OP)/2 image (1).

To avoid effects due to B_0 field inhomogeneity, the original Dixon method uses amplitude images although more recent methods use both amplitude and phase and so require modeling of the field map (2,3). As given in (3), the signal s in a voxel or region of interest (ROI) at echo time t is the summation of signals from the n component tissues, Eq 1:

$$s(t) = \sum_{j=1}^{n} A_j \exp(i\omega_j t)\exp(\nu_j t)\exp(i\psi t) \qquad [1]$$

where A_j is a measure of the proton density of component j, ω_j is the precession frequency/chemical shift of component j, ν_j is the T2*-related decay rate for component j and ψ is the field map. The difficulty with this modeling approach is that there are a large number of unknowns and many possible fits ("local minima") corresponding to different combinations of parameters. Some ways to reduce the burden on the fitting is to set the phase of A_j to zero, assume values for ω_j based on prior knowledge, estimate ψ by a separate procedure and ignore T2* decay since this is negligible for sufficiently closely spaced TEs. The validity of these approaches depends on many factors and is not easy to assess in general.

Theory

To avoid some of the difficulties outlined above, the amplitude of the signal is taken to avoid estimating the field map and the T2* decay terms are included in the model. The amplitude of Eq 1 is given by Eq 2, which is the interferometery equation reported in (4):

$$|s(t)| = \sqrt{\sum_{j=1}^{n}\sum_{k=1}^{n} |A_j||A_k| \exp(\nu_j t + \nu_k t)\cos(\omega_{jk} t + \delta_{jk})} \qquad [2]$$

where $\omega_{jk} = \omega_j - \omega_k$ is the difference in precession frequency and $\delta_{jk} = \angle A_j - \angle A_k$ is the difference in phase between individual components. Note that ψ is eliminated, which is the purpose of taking the amplitude. Letting $n = 2$ and $\delta = 0$ then Eq 2 simplifies to:

$$|s(t)| \approx |A_1|\exp(\nu_1 t) + |A_2|\exp(\nu_2 t)\cos(\omega t) \qquad [3]$$

Equation 3 has been used to model the measured signal in the present study. Although it is tempting to associate A_1 with water and A_2 with fat, such a distinction is not possible since the two components in Eq 3 are completely interchangeable. This is the down-side of taking the amplitude, although a similar ambiguity exists with complex data when ψ is not known. One way to distinguish fat and water is to make a further assumption; that water signal > fat signal or T2* of water > T2* of fat. Initial results from the present study indicate the latter is reasonable.

Methods

As part of a liver exam, patients were imaged using a multiple gradient echo sequence with 6 echos (1.5 T Siemens Symphony): TR 122, $\alpha = 45^\circ$, 256×160 matrix, BW 500 Hz/pixel, scan time 20 s. Fat and water signals were estimated by the 2-point Dixon method and also by the 6-point numerical modeling of Eq 3 using a fixed ω. Processing was performed offline in MATLAB.

Results

Figure 1 shows the measured signal in an ROI drawn inside the liver (green circles) as a function of TE. Data points are [62 111 63 83 58 70] at TE [2.2 4.4 6.6 8.8 11.0 13.2] ms. Using the 2-point Dixon method the water signal can be estimated to be 86.5 and the fat signal 24.5 giving a fat/water ratio of 0.28.

The lines are best fit curves to Eq 3 for different values of ω. The results are given in Table 1. Fitted values are relatively insensitive to ω. It is found that one of the components has a T2* of ~7 ms. Attributing this to fat, the fat/water ratio is calculated to be 0.44. This reveals there is a large discrepancy with the result from the 2-point method, which neglects T2* effects.

Figure 1 A plot of signal amplitude in liver as a function of TE (Eq 3).

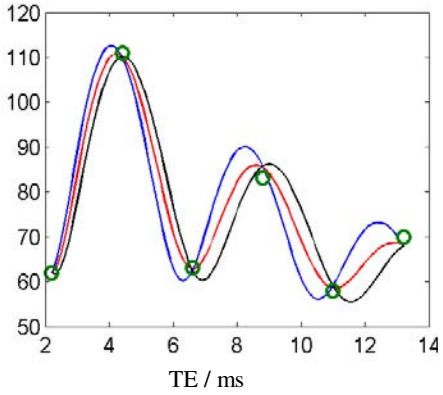

Table 1 Values of the fitted parameters for different ω. The residual norm for each fit is shown, indicating $\omega = \pi/2.2$ gives the best fit.

	$\omega = \pi/2.1$ (blue)	$\omega = \pi/2.2$ (red)	$\omega = \pi/2.3$ (black)
A_1	101.5 ± 3.4	102.7 ± 2.9	101.8 ± 3.3
A_2	41.2 ± 5.3	45.0 ± 4.7	41.9 ± 5.2
$T2_1$*	26.5 ± 2.9	26.0 ± 2.4	26.4 ± 2.8
$T2_2$*	8.77 ± 1.83	6.79 ± 0.87	8.26 ± 1.57
Error	14.8	10.2	13.7

Discussion

Quantification of fat is increasingly of interest in MRI (5). This study has found that T2* effects, which are generally ignored in Dixon methods, can cause an underestimation of the fat/water ratio when the T2* values are short. The T2* values of fat and water in liver have been found consistently to be in the range (5 – 10) ms and (20 – 30) ms, respectively. These T2* values for water are the same as those measured in patients with no fat infiltration.

References

(1) Dixon W. Simple proton spectroscopic imaging. Radiology 1984;153:189 **(2)** Glover G. Multipoint Dixon Technique for Water and Fat Proton and Susceptibility Imaging. J Mag Res Imag 1991;1:521 **(3)** Reeder SB, Wen Z, Yu H, Pineda AR, Gold GE, Markl M, Pelc NJ. Multicoil Dixon Chemical Species Separation With an Iterative Least-Squares Estimation Method. Magn Reson Med 2004;51:35 **(4)** Ford JC, Wehrli FW. In vivo Quantitative Characterization of Trabecular Bone by NMR Interferometry and Localized Proton Spectroscopy. Magn Reson Med, 1991;17:543-551 **(5)** Kovanlikaya A, Mittelman SD, Ward A, Geffner ME, Dorey F, Gilsanz V. Obesity and fat quantification in lean tissues using three-point Dixon MR imaging. Pediatr Radiol, 2005;35:601