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Relaxation effects in the quantification of fat using gradient echo imaging☆

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Abstract

Quantification of fat has been investigated using images acquired from multiple gradient echos. The evolution of the signal with echo time and flip angle was measured in phantoms of known fat and water composition and in 21 research subjects with fatty liver. Data were compared to different models of the signal equation, in which each model makes different assumptions about the T1 and/or T2* relaxation effects. A range of T1, T2*, fat fraction and number of echos was investigated to cover situations of relevance to clinical imaging. Results indicate that quantification is most accurate at low flip angles (to minimize T1 effects) with a small number of echos (to minimize spectral broadening effects). At short echo times, the spectral broadening effects manifest as a short apparent T2 for the fat component.

Keywords: Fat quantification; T1 relaxation; T2* relaxation; Dixon

1. Introduction

One of the most commonly used magnetic resonance (MR) methods for fat detection and quantification is gradient recalled echo (GRE) imaging using Dixon's two-point technique [1]. This method assumes a simplified two-component system wherein the observed MR signal is the summation of two signal sources: fat protons and water protons, which are characterized by a distinct chemical shift of approximately 3.5 parts per million (ppm). Images are acquired at two echo times (TEs) at which the signals from fat protons and water protons are presumed to be exactly in-phase (IP) and out-of-phase (OP). Since these images are acquired with TEs that are short relative to the T2* of healthy liver (~30 ms at 1.5 T), it is normally assumed that T2* decay is negligible and all signal variation between the two TEs is due to the phase interference of the fat and water protons.

Under these assumptions, the IP image is just the sum of the water and fat signals, and the OP image is the difference between the water and fat signals. The water and fat amplitudes (S_1 and S_2, respectively) can then be estimated by linear combinations of the IP and OP images.

$$S_1 \approx (\mathrm{IP}+\mathrm{OP})/2 \quad (1)$$

$$S_2 \approx (\mathrm{IP}-\mathrm{OP})/2 \quad (2)$$

Nominally, it is assumed that S_1 is water and S_2 is fat, although, in practice, these can only represent the "majority" and "minority" components of the signal, which is the familiar fat–water ambiguity problem [2–4]. Ambiguity is related to B0 field homogeneity, which, in turn, is related to the problem of multidimensional phase unwrapping [5].

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Table 1
A list of the symbols used in the present study

Symbol	Meaning
F	Fat fraction
S	Total signal
S_j	Signal of component j
ω_j	Chemical shift
v_j	R2 relaxation rate (1/T2)
v_{sys}	System R2 relaxation rate
v_j^*	R2* relaxation rate (1/T2*)
μ_j	R1 relaxation rate (1/T1)
ψ	Field map
ϕ	Phase offset
t	TE
T	TR
α	Flip angle

The first equation provides a means of obtaining a water-only image (i.e., fat–suppression contrast) and the second a fat-only image. However, of particular clinical interest is the fat fraction, defined by Eq. 3.

$$F = \frac{S_2}{S_1 + S_2} \tag{3}$$

When the conditions leading to Eqs. 1 and 2 are satisfied, the fat fraction can be estimated by rearrangement of Eqs. 1 and 2.

$$F_{\text{Dixon}} = \frac{IP - OP}{2\text{IP}} \tag{4}$$

The 2-point Dixon technique is clinically used as an indicator for the presence of fat, although it is recognized that Eq. 4 is not valid for many cases of clinical importance where T2* decay is significant [6]. A regular occurrence is that the T2* decay causes signal from the later echo to be lower than the signal from the earlier echo, which leads to spurious negative values for the fat fraction when the IP image is acquired later than the OP image.

In addition to the effects of T2* relaxation, the fat fraction can be shown to be sensitive to differences in T1 relaxation [7], which introduces a dependence on imaging parameters such as repetition time (TR) and flip angle (α). It may be observed that the measured fat fraction increases with flip angle owing to the preferential suppression of longer T1 components. Since imaging parameters are typically optimized for signal-to-noise ratio (SNR) and/or scan time, a given protocol may induce significant T1-weighting bias in the fat fraction estimate.

As well as relaxation effects, fat is known to have a complicated chemical spectrum that contains a number of different spectral components [8]: CH3, CH2, CH2COOR and CH=CH groups at 0.9, 1.3, 2.2 and 5.3 ppm, respectively, which collectively represent the total fat signal. The phase interaction between different fat components can add considerable complexity to the observed signal variation with TE, especially at longer TEs.

The primary goal of this article is to address the limitations of the 2-point method for fat quantification. This is done by starting with a comprehensive model of the MR signal that incorporates T2* relaxation effects, B0 inhomogeneity, spectral complexity and T1 relaxation effects and then introducing simplifications with an examination of the consequences and validity of each step.

1.1. Mathematical models

1.1.1. Comprehensive model

The measured complex signal S at echo time t from a sample composed of n components is a summation of the individual signals from all the components [9],

$$S(t) = \sum_{j-1}^{n} S_j \exp(i\omega_j t)\exp(-v_j^* t)\exp(i\psi t)\exp(i\phi) \tag{5}$$

where the individual terms are defined in Table 1. Strictly, the chemical shift term ω_j depends on the field strength and should have a dependence on ψ; however, ψ is typically expressed in ppm of the main field, and therefore, its effect on ω_j is usually negligible.

The terms $v_j^* = v_j + v_{sys}$ contain a contribution from the T2 of the component (v_j), which is an intrinsic property of the tissue, and the "system" (v_{sys}), which is dependent on the imaging voxel size, susceptibility within the patient and the action of contrast agents [10,11]. When v_{sys} dominates v_j, then v_j^* may be approximated as v_{sys} for all components [12].

As well as the variation with the echo time, the components have a dependence on the repetition time T and flip angle α given by Eq. 6 [9].

$$S_j(T,\alpha) \sim \frac{(1 - \exp(-\mu_j T))\sin\alpha}{1 - \exp(-\mu_j T)\cos\alpha} \tag{6}$$

where the terms are defined in Table 1. Any differences between the T1 of fat and water introduce a dependence of the estimated fat fraction on the TR and/or flip angle, which can subsequently create a dependence on B1 field inhomogeneity. One way to mitigate T1 effects is to use a long TR and/or low flip angle so that full T1 recovery may be assumed; otherwise, T1 relaxation must be taken into account.

Eqs. 5 and 6 provide a complete description of the MR signal $S(t,T,\alpha)$ at any TE, TR and flip angle; therefore, modeling the signal variation with TE and α provides a way to estimate the components S_j and, hence, the fat fraction F. The fat fraction estimated in this manner is a reflection of the relative abundance of water and fat protons only and not of any MR specific parameter. A comparable modeling approach is used in the spectroscopy analysis package jMRUI (e.g., see Ref. [13]) to perform curve-fitting of the free induction decay signal to a generalization of Eq. 5 (in which S_j is a complex variable). The number of data points acquired in spectroscopy is of the order 10^3 compared to the number of unknowns $3n+1$, and so, the approach is