Image-based determination of the fat signal model for Dixon water and fat imaging

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Introduction: Most Dixon water and fat imaging methods assume a simple fat spectral model by which the fat signal is represented as a single resonance peak with a constant peak amplitude at different echo times[1,2]. The actual complexity of the fat spectrum, however, can result in inaccurate Dixon processing and incomplete fat-water separation. For a more accurate description, some investigators used high-resolution MR spectrum of fat (usually acquired at a very high field-strength that is different from that used for imaging[3]) and modeled the fat signal as a linear combination of contributions from different spectral components of fat[4]. In this work, we propose a simple method to determine a general fat signal model directly from the images acquired for Dixon processing. Compared to the spectral approach, our method can account not only for the different spectral components of fat, but also for other complicating factors such as the pulse sequence and scan parameters.

Method: In our method, the signal of a complex image from a Dixon acquisition at echo time TE is expressed as:

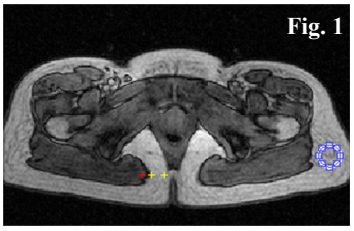
$$S = \left(W + \delta F e^{i\theta}\right)P \qquad \text{(Eq. 1)}$$

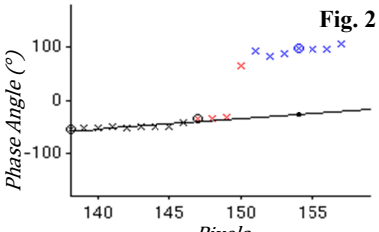
in which W and F are the scalar amplitudes of water and fat in a given pixel. δ and θ are the amplitude scaling factor and the phase of the fat signal, respectively. For a given pulse sequence and a set of scan parameters, both δ and θ are assumed to be a function of TE. P is a phasor or a complex number with unit amplitude. The phase of P includes contributions to the signal phase other than the chemical shift (e.g., magnetic field inhomogeneity), and therefore is generally assumed spatially smooth. Note that in the original single peak model for fat, it is assumed that $\delta = 1$ and $\theta = \alpha\, TE$, where α is a constant proportional to the chemical shift of fat (approximately 3.35ppm).

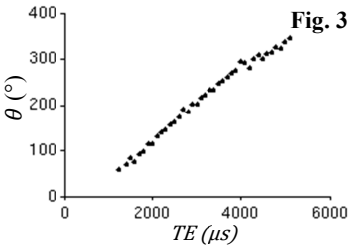
In order to determine δ and θ as a function of TE, we collected a series of image datasets from the lower pelvic region of a subject. We used a 3D dual-echo fast spoiled gradient echo sequence that allowed for user-adjustment of the echo times. These images included a fixed minimum TE_1 of 1.2ms with a range of TE_2s (from 2.9ms to 5.2ms with an increment of 0.1ms), as well as a fixed TE_2 of 4.6ms with a range of TE_1s (from 1.2ms to 2.9ms with an increment of 0.1ms). All the images were acquired on a clinical 1.5 Tesla whole body scanner using the built-in birdcage body RF coil. The following scan parameters were used: flip angle = 12°, receiver bandwidth = ±83.33kHz, acquisition matrix = 192×160, FOV = 42cm, slice thickness = 5mm, and a total of 32 slices in approximately 13 seconds. The scanner settings (receiver gains and transmit/receive center frequency) were fixed throughout the experiment.

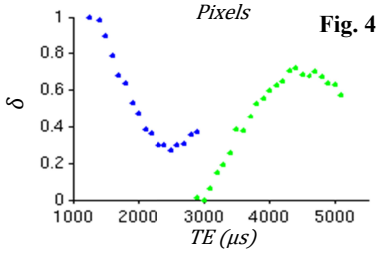
For image analysis, we developed an interactive tool in Matlab that allowed us to visually select three pixels on a straight line. Using this software tool, we visually identified a homogeneous region of subcutaneous fat and an abutting region of muscle on each image. We then placed the first two data points in the fat region and the third point in the muscle region (yellow and red points in Fig. 1). The software then performed a linear fitting of the signal phase using all the data points within the first two selected data points (using the Ann-Cho algorithm that minimizes the impact of noise and possible phase wraps[5]). Based on the linear fitting, the software reported the extrapolated phase value at the third data point. Assuming that the fat region is pure fat, the muscle region is pure water and that the phase of the phasor in Eq. 1 is linear within the selected range, the value of θ for the image is given by the difference between the actual and the extrapolated phase angles at the third pixel point (Eq. 1). For δ, we measured the average signal amplitude from an ROI in a pure fat region (blue circle in Fig. 1) that is consistently placed across all the images of the different TEs. These measured δ values were then normalized to the measurement from the image of the shortest TE.

Results: Using the proposed method and the acquired data, we obtained a total of 410 δ and θ values (205 from the 1st echo and 205 from the 2nd echo, and 5 different slices for each combination of the two echo times). For a given combination of TEs, the values for δ and θ were then averaged. Fig. 3 and Fig. 4 display the averaged δ and θ as a function of TE respectively. Within the TE-range of our data, we found that θ remained approximately linear with TE. In contrast, δ was oscillatory as a function of TE. Interestingly, we also found that δ from the images of the first echo (blue data points in Fig. 4) and from the images of the second echo (green data points in Fig. 4) seemed discontinuous as a function of TE. We assume that this discontinuity reflected some additional signal attenuation from the gradients used for the first echo.









Discussion: Accurate fat signal model is important for Dixon water and fat imaging. However, many confounding factors exist and all the signal dependence cannot be accounted for even assuming the fat spectrum can be accurately measured. For example, because different fat spectral components have different T1 and T2 relaxation times, the signal of each of these components may vary with TE differently depending on the specific pulse sequence, scan parameters, and the field strength used for imaging. Our proposed method determines the variation of the amplitude and phase of the fat signal from the same data that are collected for Dixon imaging. In general, different measurements are needed for different pulse sequences and scan parameters. However, since the variation between the different subjects and different anatomy is small[3], the results from a given pulse sequence and similar scan parameters can be applied prospectively to different patients.

References: [1] Dixon WT. Radiology 1984;153(1):189-194. [2] Ma J. J Magn Reson Imaging 2008;28(3):543-558. [3] Ren J, Dimitrov I, Sherry AD, Malloy CR. J Lipid Res 2008;49(9):2055-2062. [4] Eggers H, Brendel B, Duijndam A, Herigault G. Magn Reson Med 2011;65(1):96-107. [5] Ahn CB, Cho ZH. IEEE Trans Med Imaging 1986;6:32-36.