MEASUREMENT OF T1 AND T2 IN MR SPECTROSCOPIC EVALUATION OF FATTY LIVER

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INTRODUCTION

MR spectroscopy (MRS) has been validated to assess liver fat in patients with nonalcoholic fatty liver disease [1] and the general population [2]. Since T1 and T2 for water and fat are different, correction for T1 and T2 relaxation is necessary to improve accuracy [3]. T1 and T2 values from the literature may be used for these corrections, but this may itself introduce inaccuracy. For example, for the case shown in **Figure 1** (upper part; spin-echo sequence, TR=1500 msec, TE=30 msec, α=90°), variation of just 1.0 standard deviation in T1 and T2 values from those reported in the literature could result in as much as an 11% decrease or a 9% increase in the calculated fat fraction. A potentially more accurate method would be to measure T1 and T2 of water and fat directly for each patient. Moreover, measurement of T1 and T2 may reflect underlying pathophysiology, and so may itself be of clinical importance. Alternatively, T1 dependence may be avoided by using "ping" (one excitation, no preparation pulses), long TR, and low flip-angle (α) methods, or a combination of these. We attempted to measure T1 and T2 with MRS for 10 patients being evaluated for fatty liver, and report here the results obtained using these different methods and discuss difficulties encountered with the measurements.

MATERIALS AND METHODS

With IRB approval, we assessed liver fat using MRS (13 MR scans; 9 children and 1 adult) on a 1.5 Tesla Siemens Symphony scanner using included spectroscopy processing software. T2 was estimated by obtaining spin-echo spectra at TE=30, 40, 50, 60, and 70 msec. T1 was estimated in 8 cases by obtaining spectra at TR=1340, 2500, 5000, 7500, and 10000 msec, and in the remaining 5 cases by obtaining breath-hold spin-echo spectra at α=90, 70, 50, 30, and 10 degrees. Measurements of T1 and T2 were considered reliable when: a) Pearson-r correlation coefficients of plots of TR or TE vs. Intensity (I) had $p<0.05$, or b) plots of α vs. Intensity showed a clear maximum. Measurements of T1 and T2 from a water phantom and from a 52%-fat Microlipid (Novartis) phantom were also obtained using a variety of imaging and spectroscopic methods. The various methods of measuring T1 and T2 are compared.

RESULTS

Full width at half maximum (FWHM), and fat fraction: For the 13 MRS scans of human subjects, FWHM of the water peak was 17.5±4.5 Hz (range 11.4 to 25.7 Hz). Fat fraction ranged from 2.7 to 54.8%. FWHM for the Microlipid phantom was 1.76 Hz (water peak). Measurements of the Microlipid fat phantom were used to validate our methods. Sample spectra for a typical patient, and the 52%-fat Microlipid phantom are shown in **Figure 1**.

T2 measurement: In human subjects, plots of TE vs. ln(I) were linear ($p<0.05$) for 13/13 water and 11/13 (summed) fat peaks. Average T2 for the water peak was 39.2±5.0 msec, and of the (summed) fat peaks was 49.6±10.7 msec. These are comparable to values reported in the literature (T2=49±10 msec for water, and 71±9 msec for fat [4]).

T1 measurement - TR method. A fit with $p<0.05$ was obtained in 1/8 cases for water and 0/8 cases for fat. At least in part, this method was unsuccessful because the minimum sequence default TR (1340 msec) was too high, and both water and fat were mostly recovered by that time; in the future we will use a shorter TR for this method.

T1 measurement - α method. Plots of α vs. Intensity showed a clear maximum in 1/5 cases, for both water and fat spectral peaks. This method was largely unsuccessful because the "noise" (i.e., variability) of measurement was large relative to the dependence of Intensity on T1. Recognizing this limitation, we used the lowest TR possible on our scanner, which was 530 msec. A lower TR may have been more successful, but is not yet available on our scanner.

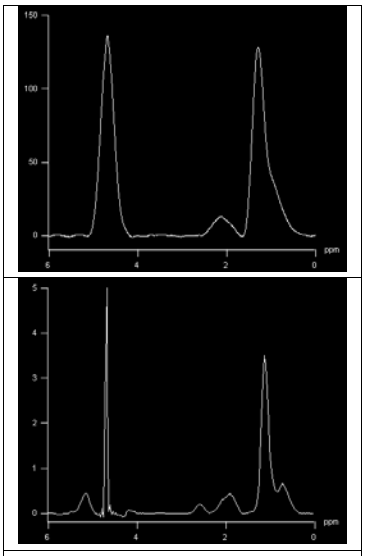

Figure 1. Sample spectra from a typical patient (above), and the 52%-fat Microlipid phantom (below).

DISCUSSION

Accurate fat fraction estimation requires reliable correction for relaxation effects, or avoidance of a dependence on those effects. T2 can be measured reliably, but measuring T1 in-vivo is difficult. Hence, it may be best to utilize sequences that avoid T1 dependence, only using T1 values from the literature when necessary, and to measure T2 directly.

The purpose for which increased accuracy is required, over and above that provided by using T1 and T2 values from the literature, should also be considered. If the most accurate results possible are required, as in a research setting, then measurement of T2 is probably a good idea, and the extra time expended in additional scanning is justified. We do not yet know how much accuracy is required clinically, and so any decision on how much emphasis to place on accuracy for clinical purposes should be deferred until we understand better how this information will be used.

Many other factors may affect estimation of fat percentage by MRS, such as voxel location, patient and treatment factors (motion, breath size consistency during breath-holds, scanner limitations related to pulse sequence parameters, liver iron content, use of an iron-containing contrast agent), method used to process spectrum, level of experience of individual processing spectrum, choice of method to estimate T1 and T2, assumptions about liver fatty acid composition, and model used to estimate corrected fat fraction. Although some of these factors have been discussed in the literature, most are still incompletely understood.

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