T1 Effect in Fat Quantification Errors in RF Saturation and IDEAL Gradient Echo Imaging

H.C. Chang1, C-J. Juan1, Y-C. K. Huang1, and H-W. Chung3

1Applied Science Laboratory, GE Healthcare Taiwan, Taipei, Taiwan, 2Department of Radiology, Tri-Service General Hospital, Taipei, Taiwan, 3Department of Electrical Engineering, National Taiwan University, Taipei, Taiwan

Introduction
Quantification of fat content using MRI provides important information for clinical diagnosis such as in hepatic steatosis [1]. Several groups have used different MRI imaging techniques to estimate liver fat content including spectroscopic approach [2], opposed-phase imaging [2,3], RF saturation [1,3] and three-point Dixon IDEAL (iterative decomposition of water and fat with echo asymmetry and least-squares estimation) [2]. However, tissue relaxation, in particular T1, is a potential source of bias in fat quantification unless corrections are performed [4]. In this experimental work using phantoms with long and short T1 values, we report the quantitative influences of the T1 effect on fat quantification when using RF saturation and IDEAL methods.

Materials and Methods
On RF saturation measurements, the fat-fraction was computed from MR images with and without fat-saturation according to the definition of fat fraction and water fraction as fat = (1 - I_{sat}/I_{unsat}) (1) and f_{water} = I_{water}/I_{unsat} (2), where I_{sat} and I_{unsat} are signal intensities with and without fat saturation, respectively. For IDEAL, complex data acquired from gradient echo at different TRs were reconstructed separately for fat and water images according to the original IDEAL algorithm for multi-coil data acquisition [5]. From the calculated fat and water images the fat-fraction was obtained as, fat_{IDEAL} = I_{fat}/(I_{fat} + I_{water}) (3) and fat_{IDEAL} = I_{fat}/(I_{fat} + I_{water}) for SPGR, respectively, where the subscript “sub” represents fat or water substance. The true fat fraction was therefore fat_{true} = (1-exp(-TR/T1_{sub})*sin(α))/(1-exp(-TR/T1_{sub})*sin(α) + cos(α)) for FSE and according to S_{sat} = M_{0sub} (1-exp(-TR/T1_{sat})) for IDEAL fat-water separation method. T1 and T2 relax times were measured using a 1.5T MRI scanner (Signa HDx, GE) using an 8-channel head coil: FSE-T1-weighted images with and without fat-saturation (TE/TR=1.87/3.43/4.99ms, TR=10ms, flip angle=10°), FSE-PD-weighted images with and without fat-saturation (TE/TR=1.68/4000ms, ETLE=8, slice thickness=60mm), and IDEAL fat images were determined using Eqs.(1,3) and water fraction maps (Figs.1f & 1j) determined from RF saturation & IDEAL method using Matlab 7.0 on a personal computer.

Results
The experiment data versus true fat fractions using FSE-T1, FSE-PD, and SPGR were shown in Figs.1a & 1b for the short and long T1 phantoms, respectively. The solid lines with different colors represented the computed bias in fat fraction measurements in each sequence, showing good agreement with the experimental measurements. The maximum bias for the short-T1 phantom was 12.1% for FSE-T1, 0.37% for FSE-PD and 6.7% for SPGR, respectively, and were much higher in the long-T1 phantom (23.2% for FSE-T1, 1.3% for FSE-PD, and 17.8% for SPGR, respectively). Figs.1c, 1d, 1g, & 1h showed, respectively, the experimental measurements. The maximum bias for the short-T1 phantom was 12.1% for FSE-T1, 0.37% for FSE-PD and 6.7% for SPGR, respectively, and were much higher in the long-T1 phantom (23.2% for FSE-T1, 1.3% for FSE-PD, and 17.8% for SPGR, respectively). Figs.1c, 1d, 1g, & 1h showed, respectively, the experimental measurements. The maximum bias for the short-T1 phantom was 12.1% for FSE-T1, 0.37% for FSE-PD and 6.7% for SPGR, respectively, and were much higher in the long-T1 phantom (23.2% for FSE-T1, 1.3% for FSE-PD, and 17.8% for SPGR, respectively). Figs.1c, 1d, 1g, & 1h showed, respectively, the experimental measurements. The maximum bias for the short-T1 phantom was 12.1% for FSE-T1, 0.37% for FSE-PD and 6.7% for SPGR, respectively, and were much higher in the long-T1 phantom (23.2% for FSE-T1, 1.3% for FSE-PD, and 17.8% for SPGR, respectively). Figs.1c, 1d, 1g, & 1h showed, respectively, the experimental measurements. The maximum bias for the short-T1 phantom was 12.1% for FSE-T1, 0.37% for FSE-PD and 6.7% for SPGR, respectively, and were much higher in the long-T1 phantom (23.2% for FSE-T1, 1.3% for FSE-PD, and 17.8% for SPGR, respectively). Figs.1c, 1d, 1g, & 1h showed, respectively, the experimental measurements. The maximum bias for the short-T1 phantom was 12.1% for FSE-T1, 0.37% for FSE-PD and 6.7% for SPGR, respectively, and were much higher in the long-T1 phantom (23.2% for FSE-T1, 1.3% for FSE-PD, and 17.8% for SPGR, respectively).

Discussion & Conclusion
Our results suggest fat quantification is affected by the T1 value of the tissue in a nonlinear manner, with large bias encountered in long-T1 than short-T1 tissues. While T1 relaxation is an obvious factor that could influence accuracy in fat quantification, previous studies seldom considered the T1 effects [3]. The results from our study show that the T1 bias in fat-fraction measurements depends on scan parameters (T1 or PD) and sequences (FSE vs. GRE), with PD-weighted images better immune to T1 biasing errors. In clinical situations where relaxation parameters may alter pathologically, therefore, T1 effects should be included in fat-fraction quantification using RF saturation and IDEAL.

Reference

Figure 1. (Right) T1 bias calculated (solid lines) and measured (markers) for short-T1 (a) and long-T1 (b) phantoms, showing good agreement. (Left) Images acquired for the short-T1 phantom with FSE-PD fat-sat (c), FSE-PD without fat-sat (d), IDEAL water image (g), & IDEAL fat image (h), along with fat fraction maps determined from RF saturation (e) & IDEAL method (f) and water fraction maps determined from RF saturation (f) & IDEAL method (j) suggest that T1 correction is needed for accurate fat-fraction measurements.