Assessment of bone marrow fat fraction in the presence of trabecular bone: initial comparison of water-fat imaging with single-voxel MRS

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Introduction: Bone marrow fat quantification has been proposed as a useful tool in understanding the relationship between osteoporosis and bone marrow adiposity [1] and in characterizing cellularity for radiation dosimetry in cancer patients [2]. Single-voxel MR spectroscopy has been the technique traditionally used to measure fat content in localized regions of the vertebral bodies and the proximal femur [1,3]. However, the distribution of bone marrow fat content can be spatially heterogeneous and there is a growing interest in applying chemical shift-based water fat imaging techniques for measuring bone marrow fat content with high spatial resolution [2]. Quantitative water-fat imaging techniques have been emerging [4,5] and have shown excellent agreement with single-voxel MRS in estimating fat content in different body parts [6]. However, the application of water-fat imaging in the bone marrow remains technically challenging due to the presence of trabecular bone. Trabecular bone shortens the T_2* of both water and fat components and constitutes a third phase with no or small visible MR signal [7,8] at the TEs of clinically accessible gradient echo sequences. The present study aims to perform a preliminary comparison of bone marrow fat fraction measurements using clinically accessible water-fat imaging and single-voxel MRS.

Methods: MR measurements: The left proximal femur of seven young healthy volunteers was scanned on a 3 T whole-body GE scanner using an 8-channel cardiac coil. A 6-echo 3D SPRG sequence was used for water-fat imaging with TR/TE/ΔTE=9.8/2.1/1.0 ms, flip angle=2°. 2-5 locations were selected per subjects to perform single-voxel (12x12x12 mm^3) MRS (resulting in a total of 25 acquired locations) based on a STEAM sequence with parameters: TR=6, TE=15/20/25/30 ms, 9 averages per TE, 4096 data points, 5 kHz acquisition bandwidth, 2.4 kHz RF pulse bandwidth, and two acquisitions at two different center frequencies (one on the main fat peak and one on the water peak).

Fat quantification: The IDEAL algorithm combined with a water-fat signal model accounting for the multiple peaks of the fat spectrum and a single T_2* correction [4,5] was used to derive fat fraction maps for visualization. Water-fat decomposition was also performed for the average complex SPGR signal over the ROIs of the MRS acquisition in order to compare mean fat fraction values between water-fat imaging and MRS.

The two MR spectra acquired with different center frequencies were combined to compensate for chemical shift displacement effects in the MRS voxel localization. The MR spectra were fitted using Gaussian lineshapes and taking into account the chemical structure of triglycerides to constrain the area of fat peaks in the water region as a given ratio of the main fat peak area [9]. MRS fitting was performed with frequency domain methods by considering a single water peak (not accounting for short T_2* species) and by considering a narrow and a broad water peak (accounting for short T_2* species). T_2 correction was then performed and the short T_2* species signal was finally excluded in the computation of fat fraction in the method accounting for short T_2* species

Results: The fitting of the MRS data using the model not accounting for short T_2* species shows poor agreement with the experimental MR spectra in the frequency region between 3 ppm and 4 ppm (arrows in Fig. 1a and 1b). The fitting of the MRS data using the model accounting for short T_2* species shows good agreement with the experimental MR spectra even in the frequency region between 3 ppm and 4 ppm (arrows in Fig. 1d and 1e).

The presently employed 6-echo SPGR sequence samples the signal at TE > 2ms and therefore most of the SPGR signal of the short T_2* species has already significantly decayed (Fig. 2b). For TE > 2 ms, there is also very good agreement between the time evolution of the T_2*-corrected MRS signal with either accounting or not accounting for short T_2* species. However, for TE < 2 ms there is a difference between the T_2-corrected MRS signal accounting for short T_2* species and the T_2-corrected MRS signal not accounting for short T_2* species.

Fig. 3 shows a significant underestimation of the fat fraction using the MRS model not accounting for short T_2* species with the fraction using the 6-echo SPGR data. There is no significant bias in fat fraction estimation using the MRS model accounting for short T_2* species with the fraction using the 6-echo SPGR data (Fig. 3).

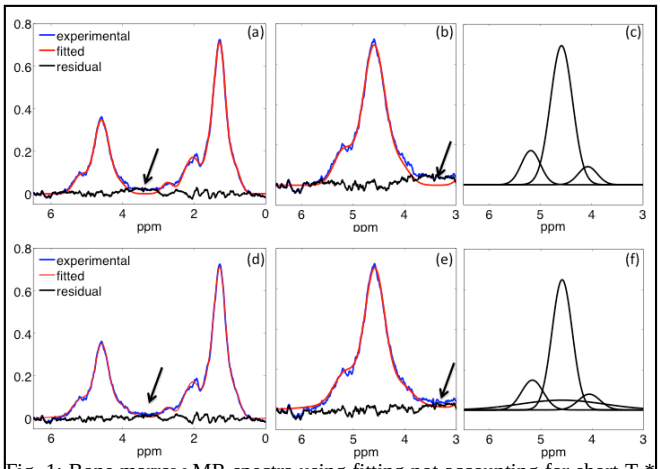

Fig. 1: Bone marrow MR spectra using fitting not accounting for short T_2* species (a-c) and accounting for short T_2* species (d-f): (a) and (d) full spectra, (b) and (e) spectra zoomed around water region, (c) and (f) superposition of decomposed modeled peaks around water region.

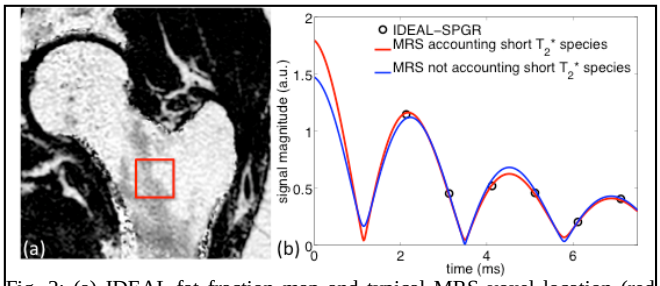

Fig. 2: (a) IDEAL-fat fraction map and typical MRS voxel location (red box), and (b) time evolution of magnitude signal: IDEAL-SPGR experimental data, T_2-corrected MRS signal accounting for short T_2* species, and T_2-corrected MRS signal not accounting for short T_2* species.

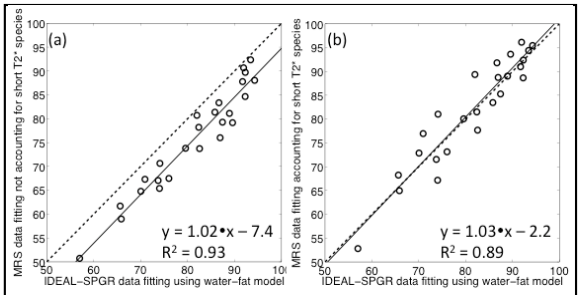

Fig. 3: Fat fraction results comparison between: (a) water-fat imaging and MRS with fitting not accounting for short T_2* species, and (b) water-fat imaging and MRS with fitting accounting for short T_2* species (dashed line represents the unity).

Discussion: A rigorous comparison of the bone marrow fat fraction measurement between single-voxel MRS and water-fat imaging should consider the effect of short T_2* species. MRS in general starts sampling the FID signal at very short acquisition times and is therefore sensitive to the presence of short T_2* species. Clinically accessible gradient echo imaging measures the signal at TEs longer than 1-2 ms and is therefore much less sensitive to the presence of short T_2* species [7,8]. When the MRS model does not account for the presence of the short T_2* species, the area of the short T_2* species ends up being considered within the water peak area inducing an underestimation of the MRS-derived fat fraction compared to the imaging-derived fat fraction. Therefore, the present preliminary data suggest that in order to compare the fat fraction measurements between MRS and imaging, the MRS data processing should use a model accounting for short T_2* species and exclude the determined area of the short T_2* species in the calculation of the MRS fat fraction. After accounting for the effect of short T_2* species a good agreement was presently found between the water-fat imaging and MRS fat fractions.

References: [1] Griffith, Radiology 236:945, 2005, [2] Pichardo, J Nucl Med 52:1482, 2011, [3] Li, J Magn Reson Imag 33:974, 2011, [4] Yu, Magn Reson Med 60:1122, 2008, [5] Bydder, Magn Reson Imag 26:347, 2008, [6] Meisamy, Radiology 258:767, 2011, [7] Reeder, Proc ISMRM 2011, p. 805 [8] Hu, Int J Bod Comp Res 9:111, 2011, [9] Hamilton, NMR Biomed 24:784, 2011.

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