Characterization of Small Liver Lesions using Partial Volume Corrected T2 Estimates Obtained from Highly Undersampled Radial Fast Spin Echo Data via PURIFY

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Introduction: The characterization of small focal liver lesions (diameter<1.5 cm) can be challenging due to the contamination from the liver within the imaging slice. This partial volume effect (PVE) is the main source of error in small lesion classification based on diffusion- or T2-weighted imaging (1, 2). Our group has developed a radial Fast Spin Echo (radFSE) method for T2 mapping of the liver (3). The method is motion insensitive and provides T2 maps of the abdomen with high spatial and temporal resolution from data acquired in a single breath hold. RadFSE has been used for the successful characterization of focal liver lesions. Here we present a novel *Partial volUme coRrected roI-based T2 Fitting of highY undersampled data* (PURIFY) technique which combines radFSE with a joint bi-exponential fitting algorithm (4) to estimate the T2 of lesions in the presence of partial volume from highly undersampled data. The method is fast (data are acquired in a breath hold) and provides accurate T2 estimates for the characterization of small focal liver lesions.

Theory: The signal from a voxel containing a mixture of lesion and background tissue (i.e. liver) can be modeled by a bi-exponential decay:

$$s(TE) = I_l e^{-TE/T2_l} + I_b e^{-TE/T2_b} + \varepsilon(TE), \quad [1]$$

where I_l, I_b are the signal intensity of the lesion and background at echo time TE =0, $T2_l$ and $T2_b$ are the corresponding T2 values, $\varepsilon(TE)$ are independent and identically distributed Gaussian noise. To estimate $T2_l$ from Eq. [1] a bi-exponential fit is required. However, bi-exponential fit suffers from large uncertainty of the fitted parameters when noise is present (4).

TABLE 1. Phantom results

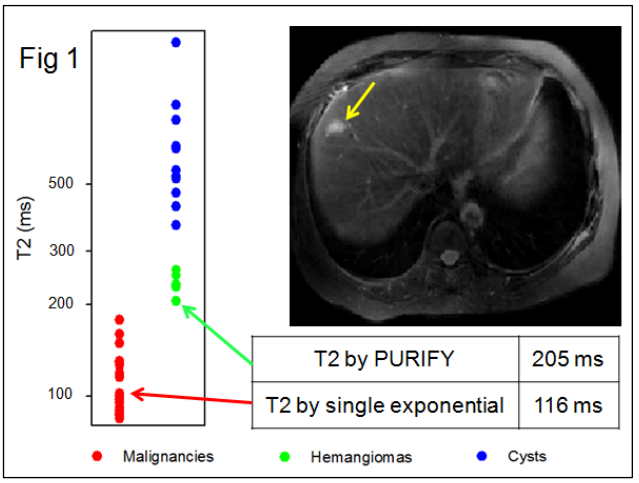
		Estimated T2 (ms)			
		6mm bulb	8mm bulb	10mm bulb	13mm bulb
Gold standard T2 = 69.0 ms	PURIFY	64.7	65.0	69.7	71.0
	Single exponential	49.3	52.5	55.3	59.0
Gold standard T2 = 179.8 ms	PURIFY	166.0	164.5	177.4	183.9
	Single exponential	68.1	96.6	111.9	122.7

To overcome this problem we use an ROI-based joint bi-exponential fitting approach. In this approach the assumption is that within a small ROI the lesion and background T2s are homogeneous. Thus, we can constrain $T2_l$, $T2_b$ on each voxel inside the ROI to two global quantities $\bar{T}2_l$, $\bar{T}2_b$ as shown below:

$$\arg\min_{I_l^1,\cdots,I_l^M,I_b^1,\cdots,I_b^M,\bar{T}_{2l},\bar{T}_{2b}} \{\sum_{n=1}^{N}\sum_{m=1}^{M} \| I_l^m e^{-TE_n/\bar{T}2_l} + I_b^m e^{-TE_n/\bar{T}2_b} - s_m(TE_n) \|^2\}. \quad [2]$$

In Eq. [2], I_l^m, I_b^m are the signal intensity of the m^{th} voxel inside the lesion's ROI at TE=0 and $s_m(TE_n)$ is the signal from the m^{th} voxel at TE_n. This algorithm has been shown to be more accurate than a simple ROI averaging technique (4).

The *s(TE)* images that are used in Eq. [2] are reconstructed using a principal component based reconstruction algorithm recently developed by our group (5). This model-based algorithm yields accurate T2 decay from highly undersampled data and accounts for the multi-component nature of the decay curves. The combination of the reconstruction algorithm for highly undersampled data and the ROI-based joint fitting approach yields PURIFY.



Methods: Data were acquired with radFSE on a 1.5T GE scanner. A total of 256 radial views (256 readout points per view) were collected with ETL=16 to yield 16 highly undersampled TE data sets (16 radial lines per TE). The TE points were equispaced by 9 ms to cover the range of 9-144 ms. TR = 1.2-1.4 s, receiver bandwidth = ±31.25 kHz, NEX=1, slice thickness = 8mm. Since PURIFY relies on accurate bi-exponential decay curves, the existence of stimulated echoes in FSE acquisitions, as a result from slice profile imperfections, is destructive. Stimulated echoes were minimized by increasing the thickness of the refocusing slice as demonstrated in (6).

Results: PURIFY was tested on a phantom composed of four NMR tubes ending in a spherical bulb to represent small lesions. The inner diameters of the bulbs varied from 6-13 mm. The tubes were filled with Magnevist with concentrations of 0.15mM and 1.5 mM to yield T2s representative of malignant (69.0 ms) and benign (179.8 ms) liver lesions. To enforce partial volume the tubes were inserted in a background bath of 4mM Magnevist (T2=39.1 ms) to represent the liver parenchyma. The in-plane resolution was 0.5 mm/pixel. The gold standard T2 values were obtained using fully sampled single-echo spin-echo data from the NMR bulbs without background (i.e. no PVE).

As shown in Table 1 the T2 estimates obtained with PURIFY from highly undersampled data in the presence of PVE are similar to the gold standard. On the other hand, single exponential fitting (where no PVE is taken into account) underestimated T2s by 15-30% for bulbs with $T2_{true}$=69.0 ms and 30-60% for the bulbs with $T2_{true}$=179.8 ms compared to the gold standard.

In vivo results are shown in Fig. 1 where the plot shows T2 values obtained from radFSE in a set of 41 patients with larger liver lesions (>1.5 cm in diameter) (3). The image shows data for a subject with a small hemangioma. Note that when a single exponential fit is used the T2 of the hemangioma falls within the range of malignant lesions (false positive) because the PVE due to liver contamination is not taken into account. When the PURIFY approach is used the lesion is within the range of benign lesions (true negative). It is noteworthy to mention that the acquisition of data used for the PURIFY reconstruction only took 22s.

Conclusions: In this work, we demonstrated that PURIFY yields accurate T2 estimates in the presence of partial volume. This enables the characterization of small lesions which are typically contaminated with background tissue (as in the liver) using highly undersampled data which can be acquired in just a breath hold.

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