Reduced Field of View Diffusion-Weighted Imaging of the Thyroid Gland Using 2D RF Pulses and Optimal B_1 Reconstruction

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Introduction

Diffusion-weighted (DW) imaging has been recently shown to provide useful information to differentiate between benign and malignant thyroid nodules [1-3]. However, conventional single-shot (SS) DW echo-planar images often suffer from severe distortion due to susceptibility variations, eddy currents and chemical shift. Reduced field of view (r-FOV) techniques can be used to alleviate off-resonance-induced phase errors by increasing the velocity of k-space traversal in the phase direction in cases where the anatomy of interest allows substantial reduction of the FOV in one dimension. These techniques, however, while alleviating distortions, result in decreased SNR, which can lead to systematic underestimation of apparent diffusion coefficients (ADCs) as a result of rectified noise being included into the fitting procedure when magnitude averaging is used. Optimal B_1 image reconstruction (OBR) using array coils [4], i.e. conventional non-accelerated parallel-imaging reconstruction, has been shown to improve the accuracy of T_2 and T_2* quantification in low-SNR cases [5], suggesting a potentially useful application of this technique to reconstruct inherently low-SNR r-FOV images for accurate ADC mapping. In this work, we used a 2D spatially selective RF pulse similar to the one designed by Saritas *et al.* [6,7] to perform contiguous, multi-slice, fat-suppressed, r-FOV, SS-DW-EPI of the thyroid gland at 3T. OBR and conventional sum-of-squares (SoS) reconstruction were used to reconstruct both r-FOV and full-FOV SS-DW-EPI images and the resulting ADC values were compared in phantom experiments and healthy volunteers.

Methods

Phantom experiments and in vivo studies were performed on a 3.0T whole-body MR system (Signa HDx, GE Healthcare, Waukesha, WI) with a maximum gradient strength of 40mT/m and a maximum slew rate of 150mT/m/s. Body-coil transmission and one of the two arms of a receive-only, bilateral, 4-channel, phased-array carotid coil (2 channels; PACC, Machnet BV, Eelde, The Netherlands) were used in all the experiments. The r-FOV pulse sequence used in this work was a dual spin echo, DW-SS-EPI pulse sequence with a 2D spatially selective RF excitation pulse (Fig.1) similar to the one reported by Saritas *et al.* [6]. Diffusion gradients were applied in the superior-inferior, left-right and anterior-posterior directions and b-values of 0 and 500s/mm^2 were used. A conventional EPI readout with maximum gradient strength and 62.5% partial k-space coverage were used. A non-accelerated SENSE-type parallel imaging reconstruction (ASSET, GE Healthcare, Waukesha, WI) [8] taking into account the different coil-element B_1 fields [4,5] was used. SoS reconstruction was also performed for comparison. In both cases magnitude averaging over all the repetitions was used.

<u>*Phantom experiments:*</u> Spherical phantoms (2.5cm diameter) containing de-ionized water and n-tridecane were used to validate the ADC measurements. ADCs were measured at 18.5ºC. The r-FOV pulse sequence was compared with a full-FOV, SS, dual spin echo DW-EPI pulse sequence. The two sequences were identical except for the excitation. Sum-of-squares and OBR (ASSET×1) were applied to both data sets. Imaging parameters for the r-FOV pulse sequence were: TE=78ms; TR=3000ms; slice thickness=4mm; FOV=22×5.5cm; imaging matrix=128×32; 16 NEX. Imaging parameters for the conventional SS- EPI case were: TE=81ms; TR=3000ms; slice thickness=4mm; FOV=22×22cm; imaging matrix=128×128; 16 NEX.

<u>*In vivo evaluation:*</u> Six healthy volunteers were imaged with the r-FOV and full-FOV DW-SS-EPI pulse sequence. Imaging parameters were the same as for the phantom experiments. For full-FOV imaging, fat saturation was achieved using both a spectrally selective saturation pulse and a water-selective excitation pulse. Spatial saturation bands were also used to remove signal from overlying fat and nearby tissues. Mean ADC values (±standard deviation) were evaluated on a per-subject basis.

Results and Discussion

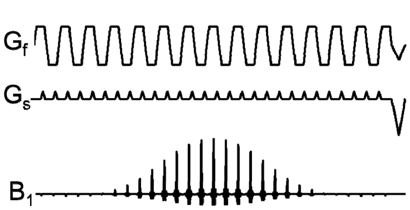
Phantom experiments (Table 1) showed that accurate ADC measurements could be performed with both the r-FOV and full-FOV techniques. SOS and OBR gave very similar results due to the high SNR achieved in phantom. In vivo, r-FOV images with OBR showed considerably less distortion than the corresponding full-FOV images (Fig.2). Fig. 3 shows that when SoS reconstruction was used, r-FOV data systematically underestimated ADC values. When OBR was used, the difference between ADC measurements performed on r-FOV and full-FOV images was no longer significant (p>0.01) in all but one case (Vol.5). In conclusion, we showed that r-FOV imaging using 2D RF pulses and OBR can be used for accurate ADC measurements in the thyroid.

TABLE 1

	Water (ref. ADC* = 1.94§)		n-tridecane (ref. ADC = 0.62)	
	SOS	**OBR**	**SOS**	**OBR**
r-FOV ADC	1.97±0.02	2.05±0.02	0.66±0.01	0.66±0.01
Full-FOV ADC	2.08±0.03	2.08±0.03	0.68±0.01	0.68±0.01

* Toft PS et al. MRM 2000; 43:368; §ADC values in 10^{-3}mm^2/s.

References: [1] Erdem G et al. JMRI 2010; 31:94; [2] Bozgeyik Z at al. Neuroradiology 2009; 51:193; [3] Razek AA et al. AJNR 2008; 29:563; [4] Roemer PB et al. MRM 1990; 16:192; [5] Graves MJ et al. JMRI 2008; 28:288; [6] Saritas EU et al. MRM 2008; 60:468; [7] Saritas EU et al. ISMRM 2010; p.189; [8] King KF et al. ISMRM 2000; p.153.



- Slow dir. = Slice dir.; Fast dir. = Phase encode dir.;
- 4mm (slice) × 50mm (phase) slab;
- 28 blips in the slice dir. and time-BW products of 3 and 12 in the slow and fast dir., respectively;
- Pulse width = 23ms;
- Peak B_1 = 0.16 Gauss;
- Inherent fat suppression achieved by shifting the fat profile outside of the refocused volume [6].

Figure 1

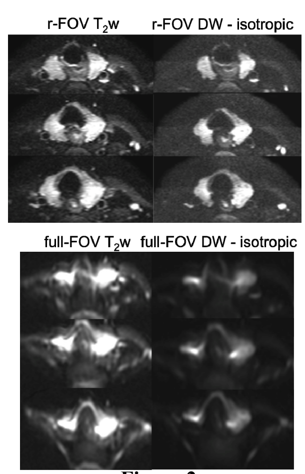


Figure 2

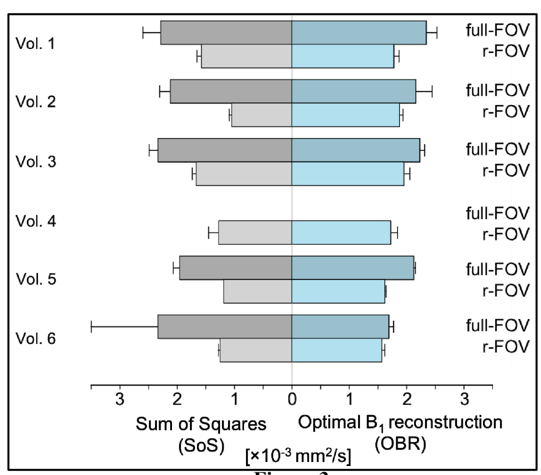


Figure 3