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FULL PAPER

3T diffusion-weighted MRI of the thyroid gland with reduced distortion: preliminary results

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Objective: Single-shot diffusion-weighted (DW) echo planar imaging (EPI), which is commonly used for imaging the thyroid, is characterised by severe blurring and distortion. The objectives of this work were: 1, to show that a reduced-field of view (r-FOV) DW EPI technique can improve image quality; and 2, to investigate the effect of different reconstruction strategies on the resulting apparent diffusion coefficients (ADCs).

Methods: We implemented a single-shot, r-FOV DW EPI technique with a two-dimensional radiofrequency excitation pulse for DW imaging of the thyroid at 3T. Images were reconstructed using root sum of squares (SOS) and an optimal-B₁ reconstruction (OBR). Phantom and *in vivo* experiments were performed to compare r-FOV and conventional full-FOV DW EPI with root SOS and OBR.

Results: r-FOV with OBR substantially improved image quality at 3T. In phantoms, r-FOV gave more accurate ADCs than full-FOV. *In vivo* r-FOV always gave lower ADC values with respect to the full-FOV technique irrespective of the reconstruction used and whether only two or multiple *b*-values were used to compute the ADCs.

Conclusion: r-FOV DW EPI can reduce image blurring and distortion at the expense of a low signal-to-noise ratio. OBR is a promising reconstruction technique for accurate ADC measurements in lower signal-to-noise ratio regimes, although further studies are needed to characterise its performance.

Advances in knowledge: DW imaging of the thyroid at 3T could potentially benefit from r-FOV acquisition strategies, such as the r-FOV DW EPI technique proposed in this paper.

Diffusion-weighted (DW) imaging has been shown to provide useful information to differentiate between benign and malignant thyroid nodules [1–4]. Single-shot echo-planar imaging (EPI) is the most frequently used technique because of its relative immunity to motion-induced phase errors, high signal-to-noise ratio (SNR) and short acquisition times [5,6]. However, single-shot DW EPI images are characterised by blurring along the phase-encode direction due to T_2^* decay and are sensitive to off-resonance effects (B_0 inhomogeneities, susceptibility gradients, eddy currents, chemical shift) because of the narrow bandwidth in the phase-encode direction.

The use of parallel imaging has been shown to provide considerable improvement in terms of image quality [7,8]. However, residual aliasing artefacts and noise amplification are often observed with clinically available phased-array coils combined with reduction factors >2 , which limits the applicability of this technique. Another strategy to reduce the number of acquired *k*-space lines and, hence,

off-resonance-induced phase errors and blurring, is to restrict the excitation phase field of view (FOV) to the anatomy of interest, so that the EPI echo-train length can be reduced without fold-over artefacts. This technique has been shown to be particularly effective for imaging the spine [9–12] and has also been reported in the carotid artery [13], the optic nerve [14] and the prostate [15]. Unlike parallel imaging, the phase FOV can be arbitrarily reduced to frame the anatomy of interest with an SNR penalty which is generally lower than that associated with parallel imaging combined with the same reduction factor. However, depending on the approach adopted in excitation to limit the FOV, contiguous multi-slice imaging may not be possible [11], or may require additional pulses for outer-volume suppression [12], or may be characterised by a variable SNR as a function of the number of slices [9]. Saritas et al used a two-dimensional (2D) spatially selective radiofrequency (RF) pulse in combination with a 180° re-focusing pulse to limit the excitation FOV in the phase-encode direction and allow contiguous multi-slice imaging

[10,16,17]. The same pulse sequence also provided fat suppression without the need for additional pulses by chemically shifting the fat profile outside of the imaging volume refocused by the 180° pulse.

A well-known drawback of quantitative DW imaging in inherently low-SNR situations, as in the case of reduced-FOV (r-FOV) techniques, is the systematic underestimation of apparent diffusion coefficients (ADCs), which can result from rectified noise being included into the fitting procedure if magnitude averaging is used [18]. In the original work from Saritas et al, complex averaging was used after correcting the raw data for eventual shot-to-shot motion-induced phase errors at the expense of an increased complexity of the reconstruction [10,19]. Optimal- B_1 image reconstruction (OBR) using array coils [20], *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 [21], suggesting a potentially useful application of this technique to reconstruct inherently low-SNR r-FOV images for subsequent ADC mapping.

In this work, we used a 2D spatially selective RF pulse similar to the one designed by Saritas et al [10,16,17] to perform contiguous, multi-slice, fat-suppressed, r-FOV, single-shot DW EPI of the thyroid gland at 3T [22], together with an eddy-current-compensated diffusion-encoding scheme [23] and B_1 -optimal reconstruction, to reduce the noise floor resulting from magnitude averaging [20]. Phantom experiments were performed to validate the ADC measurements, and *in vivo* applications in healthy volunteers were evaluated and compared with the standard full-FOV single-shot DW EPI pulse sequence.

METHODS

Phantom experiments and *in vivo* studies were performed on a 3.0T whole-body MR system (Signa HDx Twinspeed, GE Healthcare, Waukesha, WI) with a maximum gradient strength of 40 mTm^{-1} and a maximum slew rate of $150 \text{ mTm}^{-1} \text{ s}^{-1}$ for the zoom gradient. Body-coil transmission and one of the two arms of a receive-only, bilateral, four-channel, phased-array carotid coil (PACC, Flick Engineering Solutions, Winterswijk, the Netherlands) were used in all the experiments.

Pulse sequence

The pulse sequence used in this work was a dual spin echo, DW single-shot EPI pulse sequence with a 2D spatially selective RF pulse (Figure 1). The overall pulse sequence design was similar to the one described by Saritas et al [10]. The 2D RF pulse was spatially selective in the slice and phase-encode direction. The two slice-selective refocusing pulses suppressed the signal from the replicates of the periodic 2D excitation while the fat profile was shifted outside of the volume refocused by the 180° pulses, so that fat suppression was achieved without additional pulses.

An echo-planar excitation- k -space trajectory [24,25] was used to produce a 90° flip angle over a $4 \times 50 \text{ mm}$ (slice $\times$ phase) slab. 28 blips in the slice direction and time-bandwidth products of 3 and 12 in the slow and fast directions, respectively, were used. This resulted in the periodic excitation lobes being 3.7 cm apart, so that nine 4-mm-thick contiguous slices could be acquired

Figure 1. Reduced field of view diffusion-weighted single-shot echo planar imaging (EPI) pulse sequence. 2D, two-dimensional; FE, frequency encoding; PE, phase encoding; RF, radiofrequency; SS, slice select; TE, echo time.

within a single multi-slice experiment. The duration of the 2D RF pulse was 23 ms and the peak B_1 was 0.16 G. All the pulse design parameters were chosen to (a) obtain the desired excitation profiles in the slice and phase directions, (b) achieve adequate separation of the periodic excitation lobes so that enough contiguous slices could be acquired to cover the thyroid gland, (c) shift the fat signal outside of the volume refocused by the subsequent slice-selective 180° pulses. A diffusion-encoding scheme similar to the one suggested in [23] was used with the dual-spin-echo pulse sequence to compensate for eddy currents. Diffusion gradients were applied in the superior–inferior, left–right and anterior–posterior directions. Dual (0, 500 s mm^{-2}) and multi- b -value acquisitions (0, 20, 50, 100, 300, 600, 900, 1200 s mm^{-2}) were performed. A conventional EPI readout with maximum gradient strength and 62.5% partial k -space coverage were used to minimise the effective echo time.

Image reconstruction

An OBR, taking into account the different coil–element B_1 fields [20,21], was performed by using a non-accelerated, *i.e.* $R=1$, SENSitivity Encoding (SENSE)-type parallel imaging reconstruction (ASSET; GE Healthcare, Waukesha, WI) [26]. In order to avoid residual fold-over artefacts from the ASSET reconstruction, the calibration pulse sequence was modified to allow the use of saturation pulses to suppress the outer volume signal in the phase-encode direction. Standard root sum-of-squares (SOS) reconstruction was also performed for comparison. In both cases, magnitude averaging over all the repetitions was performed.

Phantom experiments

Spherical phantoms (2.5 cm diameter) containing deionised water and n -tridecane were used to validate the ADC measurements. Both phantoms were placed in an expanded polystyrene box to minimise temperature variations. Temperature monitoring was performed using a thermocouple placed inside the box. ADCs were measured at 18°C (ambient air temperature). All the experiments were performed at the water and n -tridecane Larmor frequencies. The r-FOV pulse sequence was compared with a full-FOV, single-shot, dual spin echo DW EPI

pulse sequence. The two sequences were identical except for the excitation pulse. SOS and OBR (ASSET $\times 1$) were applied to both data sets. Imaging parameters for the r-FOV pulse sequence were: TE=78 ms; TR=3000 ms; slice thickness=4 mm; FOV=22 $\times$ 5.5 cm; imaging matrix=256 $\times$ 64; 16 signal averages. The imaging parameters for the conventional single-shot EPI case were: time to echo (TE)=81 ms; time to repeat (TR)=2200 ms; slice thickness=5 mm; FOV=22 $\times$ 22 cm; imaging matrix=256 $\times$ 256; 16 signal averages. All the experiments were performed using the dual and multi- b -value acquisitions. ADC values were calculated from the isotropic DW images. A non-linear least-squares algorithm was used to estimate the ADC values from the multi- b -value data.

In vivo imaging

Six healthy volunteers were imaged with the r-FOV and full-FOV DW single-shot (SS) EPI pulse sequence. Approval was obtained from the local Research Ethics Committee and all participants gave written informed consent. 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 (CHESS pulse) and a water-selective spectral-spatial excitation pulse. Spatial saturation bands were used to remove the signal from overlying fat and nearby tissues. As for the phantom experiments, ADC values were calculated from the isotropic DW images and a non-linear least-square algorithm was used for the multi- b -value data. In this case, $b < 100 \text{ s mm}^{-2}$ were not included in the fit to avoid perfusion-related signal contamination [27]. Regions of interest (ROIs) were drawn to encompass normal thyroid tissue.

Statistical analysis

Phantom experiments were reported in terms of mean ADC $\pm$ standard deviation (SD), with mean and SD computed over three consecutive slices. For the *in vivo* part of the study, a comparison of the different combinations of acquisition-reconstruction techniques (full-FOV and SOS, r-FOV and SOS, full-FOV and OBR, r-FOV and OBR) for the dual- and multi- b -value case was performed across the six volunteers who took part in the study (see Figure 4). ADC values as obtained from the dual- b -value acquisition and the different acquisition strategies and reconstruction techniques were then compared on a per-subject

basis, by reporting the mean and SD over multiple slices for each subject and each combination of acquisition-reconstruction method (see Figure 5).

RESULTS

Phantom experiments

Reference values for water and n -tridecane ADC values at 18 °C, 1.94 and $0.62 \times 10^{-3} \text{ mm}^2 \text{ s}^{-1}$, respectively, were obtained by linear interpolation of data reported in Tofts et al [28]. Table 1 shows ADC values for water and n -tridecane as obtained from the dual- and multi- b -value acquisitions for both r-FOV and full-FOV with both SOS and OBR. The relative error with respect to the reference values was consistently lower for the r-FOV technique, with the exception of the water case, when multiple b -values and SOS were used (-9.3% for r-FOV and 7.2% for full-FOV). Figure 2 shows the signal intensity decay curve as a function of b -value together with the mean background noise level for a representative ROI in the water phantom. SOS gave a higher mean background noise level with respect to OBR, which caused the ADC value to be underestimated ($\text{ADC} = 1.71 \times 10^{-3} \text{ mm}^2 \text{ s}^{-1}$). The use of OBR on the same data set reduced the mean background noise level, resulting in a signal decay curve above the noise floor for all b -values and hence a more accurate ADC value ($\text{ADC} = 1.98 \times 10^{-3} \text{ mm}^2 \text{ s}^{-1}$; 5.1% relative error with respect to the reference). Excluding the highest b -value ($b = 1200 \text{ s mm}^{-2}$) from the fit gave a similar value with OBR ($\text{ADC} = 2.01 \times 10^{-3} \text{ mm}^2 \text{ s}^{-1}$) and a higher value, closer to the one obtained with OBR and the complete data set, when SOS was used ($\text{ADC} = 1.81 \times 10^{-3} \text{ mm}^2 \text{ s}^{-1}$). Table 1 also shows that SOS and OBR gave approximately the same result when only two b -values were used, while the difference in ADC values was more pronounced when r-FOV multi- b -value acquisitions were performed. This was owing to the b -value used in the dual- b -value case ($b = 500 \text{ s mm}^{-2}$) for which both SOS and OBR gave a mean background noise level well below the measured signal intensity, even for the r-FOV acquisition. On the other hand, when higher b -values were used (maximum b -value in the multi- b -value case = 1200 s mm^{-2}) together with the r-FOV technique, OBR was necessary to reduce the mean background noise level and obtain unbiased ADC values, unless the high b -values were excluded from the calculation.

Table 1. Water and n -tridecane ADC (apparent diffusion coefficient) values [$10^{-3} \text{ mm}^2 \text{ s}^{-1}$] at 18 °C reported as mean $\pm$ standard deviation. The relative error with respect to [28] is also reported in brackets

	Dual- b -value ^a		Multi- b -value ^b	
	SOS	OBR	SOS	OBR
r-FOV-water ^c	1.97 $\pm$ 0.02 (1.5%)	2.05 $\pm$ 0.02 (5.7%)	1.76 $\pm$ 0.08 (-9.3%)	2.04 $\pm$ 0.05 (5.1%)
Full-FOV-water	2.08 $\pm$ 0.03 (7.2%)	2.08 $\pm$ 0.03 (7.2%)	2.08 $\pm$ 0.04 (7.2%)	2.10 $\pm$ 0.07 (8.2%)
r-FOV- n -tridecane ^d	0.66 $\pm$ 0.01 (6.4%)	0.66 $\pm$ 0.01 (6.4%)	0.62 $\pm$ 0.02 (0.0%)	0.66 $\pm$ 0.01 (6.4%)
Full-FOV- n -tridecane	0.68 $\pm$ 0.01 (9.7%)	0.68 $\pm$ 0.01 (9.7%)	0.82 $\pm$ 0.11 (32.2%)	0.81 $\pm$ 0.1 (30.6%)

full-FOV, full field of view; OBR, optimal-B₁ reconstruction; r-FOV, reduced field of view; SOS, sum of squares.

^a $b = 0, 500 \text{ s mm}^{-2}$.

^b $b = 0, 20, 50, 100, 300, 600, 900, 1200 \text{ s mm}^{-2}$.

^cReference ADC for water = $1.94 \times 10^{-3} \text{ mm}^2 \text{ s}^{-1}$ [28].

^dReference ADC for n -tridecane = $0.62 \times 10^{-3} \text{ mm}^2 \text{ s}^{-1}$ [28].

Figure 2. Signal intensity ($\pm$ standard deviation) vs b -value for a representative slice across the water phantom acquired with the reduced field of view technique and both root sum of squares (SOS) and optimal- B_1 reconstruction (OBR). The fitted signal decay curves and the mean background noise levels ($\pm$ standard deviation) are also shown. Mean apparent diffusion coefficients were 1.71 and $1.98 \times 10^{-3} \text{ mm}^2 \text{ s}^{-1}$ for SOS and OBR, respectively.

In vivo experiments

The full-FOV images showed severe distortion of the thyroid gland in all volunteers. r-FOV images were substantially less distorted than the corresponding full-FOV images resulting in better delineation of the underlying anatomy. The improved quality of the r-FOV images facilitated ROI placement and generally resulted in better fat suppression when compared with the corresponding full-FOV images. In one case, distortion and displacement of residual fat signal along the phase encode direction in the full-FOV images prevented confident measurement of the ADC. Figure 3 shows full-FOV and r-FOV T_2 weighted and DW images (dual- b -value acquisition, $b=0, 500 \text{ s mm}^{-2}$) acquired in a healthy volunteer and reconstructed with OBR. ADC values obtained with the r-FOV technique were lower than those obtained with full-FOV irrespective of the reconstruction method used, for both dual- and multi- b -value acquisitions (Figure 4). The mean ADC ($\pm$ SD) for dual- b -value with full-FOV was 2.21 ± 0.49 and $2.11 \pm 0.27 \times 10^{-3} \text{ mm}^2 \text{ s}^{-1}$ with SOS and OBR, respectively. Dual- b -value with r-FOV and SOS gave an ADC of $1.34 \pm 0.23 \times 10^{-3} \text{ mm}^2 \text{ s}^{-1}$, whereas dual- b -value with r-FOV and OBR gave an ADC of $1.76 \pm 0.15 \times 10^{-3} \text{ mm}^2 \text{ s}^{-1}$. In view of the results obtained in phantoms, multi- b -value *in vivo* acquisitions were reconstructed with OBR only, giving ADC values of $1.48 \pm 0.25 \times 10^{-3} \text{ mm}^2 \text{ s}^{-1}$ (full-FOV) and $1.10 \pm 0.13 \times 10^{-3} \text{ mm}^2 \text{ s}^{-1}$ (r-FOV). As shown in Figure 4, the use of multiple b -values gave ADCs that were significantly lower than those obtained from dual- b -value acquisitions. In this case, the difference between r-FOV- and full-FOV-derived ADC values was smaller when OBR was used. Figure 5 shows the mean ADC values across different slices computed for each

Figure 3. T_2 weighted (a-c) and isotropic diffusion-weighted ($b=500 \text{ s mm}^{-2}$) images (b-d) acquired in a healthy volunteer with the reduced field of view (a-b) and full field of view (c-d) techniques and optimal- B_1 reconstruction.

subject with a dual- b -value acquisition and different combinations of full-FOV/r-FOV and SOS/OBR. r-FOV images gave lower ADCs than full-FOV irrespective of the reconstruction method used. However, full-FOV was characterised by a larger SD across different slices, probably owing to the higher distortion generally found in this case.

DISCUSSION

DW imaging has been shown to be capable of differentiating between benign and malignant thyroid nodules [1–4]. However, imaging of the thyroid at 3T is challenging owing to the increased sensitivity to off-resonance effects and the close proximity to the trachea, with distortion, blurring and imperfect fat suppression often resulting in problematic interpretation of the images and uncertain ADC measurements.

In this work, we implemented a r-FOV technique similar to the one proposed by Saritas et al [10] and showed through phantom and *in vivo* experiments that the SNR penalty introduced by the smaller FOV can lead to underestimation of the resulting ADC

Figure 4. *In vivo* apparent diffusion coefficients (ADCs) obtained with dual-/multi-*b*-value acquisitions, full field of view (full-FOV)/reduced field of view (r-FOV) and root sum of squares (SOS)/optimal- B_1 reconstruction (OBR).

values, depending on the specific reconstruction used. Phantom experiments showed that when an OBR technique was used, r-FOV gave ADCs that were closer to the reference values than those obtained with full-FOV. *In vivo*, however, r-FOV always gave lower ADC values with respect to the full-FOV technique irrespective of the reconstruction used and whether only two or multiple *b*-values were used to compute the ADC. In particular, including multiple *b*-values in the ADC calculation gave ADC values that were significantly lower than those obtained with the dual-*b*-value approach. These results suggest that the intrinsically lower SNR found *in vivo*, together with the SNR penalty introduced by the r-FOV acquisition, might benefit from the use of lower *b*-values to obtain accurate ADC measurements. Saritas et al recently showed that when the true noise statistics are taken into account, the resulting optimum *b*-value is a monotonically increasing function of the imaging SNR, with the optimum *b*-value in low-SNR cases being considerably lower than the optimal *b*-value obtained under the assumption of gaussian noise [29]. The two acquisition strategies investigated in this work (dual-*b*-value with $b=0, 500 \text{ s mm}^{-2}$ and multi-*b*-value with *b*-values in the range $0\text{--}1200 \text{ s mm}^{-2}$) were chosen on the basis of previous reported protocols for DW imaging in the thyroid [1–4] and no attempt was made to optimise the maximum *b*-value on the basis of the resulting SNR. Other recent reports have claimed significant differences in ADC values when different acquisition strategies (e.g. PROPELLER vs EPI) were compared using the same *b*-values for both techniques [30,31]. At least for the present comparison between full-FOV and r-FOV SS EPI, using the same *b*-values for both full-FOV and r-FOV might have partly biased the ADC measurements.

On the other hand, there is no consensus in the literature on the *b*-values to be used for ADC mapping in the thyroid, and

Figure 5. Average apparent diffusion coefficients (ADCs) over the thyroid gland (three slices) obtained in the six volunteers using full-field of view (full-FOV) and reduced-field of view (r-FOV) acquisitions and root sum of squares (SOS) and optimal- B_1 reconstruction (OBR). In two cases (Vol. 1 and 6), full-FOV and r-FOV gave similar ADC values irrespective of the reconstruction technique used ($p>0.01$). In two cases (Vol. 2 and 3), full-FOV gave higher ADC values than r-FOV when SOS was used ($p<0.01$), whereas more similar results were obtained when OBR was used. In one case (Vol. 5), full-FOV gave a higher ADC value than r-FOV with both SOS and OBR.

a relatively wide range of ADC values in normal thyroid tissue has been reported, reflecting different acquisition techniques and imaging parameters. Previous measurements performed in normal thyroid parenchyma used *b*-values between 0 and 1000 s mm^{-2} and reported ADC values ranging from 1.25 to $3.28 \times 10^{-3} \text{ mm}^2 \text{ s}^{-1}$ [30,31]. As a consequence, it is not clear what should be regarded as the *in vivo* gold standard. Full-FOV images were heavily distorted and there were often additional artefacts caused by displaced residual fat signal. This often hampered interpretation of the images and made positioning of the ROIs problematic. Whereas in the r-FOV case only minor adjustments to the ROIs were necessary between consecutive slices, in the full-FOV case, ROIs had to be manually drawn slice-by-slice, with the result that both the ROIs' size and position were considerably different between adjacent slices (cf. the higher SD shown in Figure 5 for the full-FOV case). The much higher level of distortion in the full-FOV images is therefore another important factor that could have biased the *in vivo* full-FOV measurements and, as such, needs to be accounted for when interpreting the obtained results.

It should also be mentioned that the ADC of healthy thyroid tissue is lower than the ADC of both benign and malignant thyroid nodules [1–4]. This means that measuring the ADC of healthy tissue is considerably more challenging than measuring the ADC in the presence of pathology. In this case, the diffusion-induced signal decay is slower, so that for the range of *b*-values

considered in this work, noise will be less likely to affect the measured ADC values. For this reason, although the ADC measurements obtained with the r-FOV technique might have been biased owing to the low-SNR and rapidly decaying signal observed in healthy thyroid tissue, r-FOV acquisitions remain a very promising alternative to conventional full-FOV techniques for differentiating between benign and malignant thyroid nodules.

Unlike Saritas et al [10], we used an OBR as described by Roemer et al [20] and implemented as a non-accelerated SENSE-type reconstruction to reduce the noise floor and hence improve the accuracy of the ADC calculation. A similar approach was previously described by Graves et al, in the context of T_2 and T_2^* measurements with suboptimal acquisition parameters [21]. An obvious advantage of using a non-accelerated SENSE-type reconstruction with magnitude averaging instead of complex averaging after phase correction is that the former is readily available on most clinical scanners. However, the use of a SENSE-type reconstruction requires an additional calibration scan to obtain the sensitivity maps, which very slightly increases the total study time and could potentially lead to residual artefacts in the r-FOV images if outer volume suppression pulses are not used to match the calibration FOV and the excitation FOV produced by the 2D RF pulse. In addition, as already mentioned, it is not clear whether the discrepancy found between ADCs obtained from r-FOV and full-FOV acquisitions *in vivo* is owing to the full-FOV values being artifactually increased as a result of excessive distortion or the OBR technique not completely compensating for the lower SNR found *in vivo*. Zaharchuk et al [32] recently compared full-FOV and r-FOV single-shot EPI with complex averaging and a refocusing reconstruction for DW imaging in the spine and found the same ADC values with both techniques.

This study has several limitations. For the phantom experiments, the reference data were obtained from data reported in the literature by linear interpolation to the temperature used in

the experiments. Although extra care was taken to ensure careful monitoring of the temperature during the scan, small temperature fluctuations, which might have affected the ADC values, are difficult to exclude. For the *in vivo* experiments, only six subjects took part in the study. Finally, no attempt was made to optimise the maximum b -value used with the r-FOV pulse sequence, which might have biased the ADC measurements. Further studies therefore are necessary to assess to what extent a r-FOV technique like the one described in this paper and a non-accelerated SENSE-type reconstruction strategy, together with the selection of an appropriate b -value taking into account the true noise statistics [29], can provide improved image quality without loss of accuracy in the resulting ADC values.

CONCLUSION

In conclusion, we have shown that a r-FOV DW EPI acquisition obtained by replacing the RF excitation pulse in a conventional DW EPI pulse sequence with a tailored 2D RF pulse and an OBR can provide improved image quality for thyroid imaging at 3T. The resulting ADC values were shown to be accurate in phantoms but lower than the corresponding full-FOV values *in vivo*. Further studies are necessary to clarify the reason for this discrepancy.

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