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FULL PAPER

Artifact-robust diffusion-weighted whole-body imaging with background suppression at 3 T using improved turbo spin-echo diffusion-weighted imaging

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Objective: To compare single-shot turbo spin-echo (TSE) diffusion-weighted whole-body imaging with background suppression (DWIBS) and echo-planar imaging (EPI) DWIBS to determine the feasibility of direct-coronal TSE-DWIBS.

Methods: All measurements were performed using a 3.0 T MRI scanner. In the phantom study, we compared the contrast ratios (CRs) of tumor-mimicking phantom (tumor) to muscle-mimicking phantom (muscle) and water to muscle and the signal-to-noise ratio (SNR) between TSE-DWIBS and EPI-DWIBS. In the volunteer study, 10 healthy volunteers were whole-body scanned with direct-coronal TSE-DWIBS, direct-coronal EPI-DWIBS (corEPI-DWIBS), and transverse EPI-DWIBS (traEPI-DWIBS). Two radiologists assessed the image distortion, uniformity of fat suppression, overall artifacts, and overall image quality in maximum intensity projection (MIP) images from each DWIBS image using a 5-point scale.

Results: In the phantom study, the CR of tumor to muscle was found to be lower for TSE-DWIBS (10.57 ± 0.45) than for EPI-DWIBS (15.38 ± 0.27), and the

CR of water to muscle was higher for TSE-DWIBS (9.61 ± 0.66) than for EPI-DWIBS (2.52 ± 0.60). The volunteer study revealed good inter observer agreement between TSE-DWIBS and EPI-DWIBS with respect to image distortion, uniformity of fat suppression, overall artifacts, and overall image quality, with weighted Cohen's κ coefficients of 0.91, 0.74, 0.87, and 0.72, respectively. Qualitative analysis scores for image distortion, uniformity of fat suppression, overall artifacts, and overall image quality were significantly higher for TSE-DWIBS than for corEPI-DWIBS or traEPI-DWIBS ($p < 0.05$).

Conclusion: Direct-coronal TSE-DWIBS is robust against magnetic field inhomogeneity. High-quality images without distortion or fat suppression inhomogeneity were obtained.

Advances in knowledge: Many studies on DWIBS have been previously reported; however, these studies used EPI read-out. To the best of our knowledge, no studies using TSE read-out have been reported. In this study, we examined the feasibility of TSE-DWIBS with lesser artifacts than EPI-DWIBS.

INTRODUCTION

Diffusion-weighted whole-body imaging with background suppression (DWIBS) uses short tau inversion recovery (STIR) under free breathing conditions.¹ DWIBS can visualize malignant tumors and abnormal lymph nodes and is useful for determining the response of bone metastasis to treatment.^{2–6} Recently, UK quantitative whole-body WB-DWI technical workgroup was established that provides consensus guidance for the quantification of DWIBS.⁷ Single-shot echo-planar imaging (EPI) is commonly used for diffusion-weighted imaging (DWI)

(EPI-DWI), including DWIBS (EPI-DWIBS), using a 1.5 T or 3 T MRI scanner.⁷ Because, in order to emphasize diffusion, we must minimize the influence of motion such as body motion and pulsation. EPI is a very short acquisition pulse sequence and suitable for DWI.^{8,9} EPI-DWI has several disadvantages, including image distortion and the appearance of chemical-shift and susceptibility artifacts.^{8,10} In 3 T MRI with EPI-DWI, these disadvantages are amplified under conditions with increased magnetic field inhomogeneities.¹¹ DWI with single-shot turbo spin-echo (TSE-DWI) also has disadvantages, including a low

signal-to-noise ratio (SNR) and high image blurring.^{12–14} This technique is used mostly for the evaluation of regions with severe magnetic field inhomogeneity, including the inner ear,¹⁵ head and neck,² and breast.¹⁶

Recently, an improved TSE-DWI sequence was developed that includes a number of improvements: signal averaging in the image space to offset the signal drop owing to phase error in multiple acquisitions, a shorter RF pulse shape that enables shorter echo spacing, and compatibility with parallel imaging (sensitivity encoding) that shortens the data acquisition time and yields high-quality images.^{12,17}

EPI-DWIBS scans transverse images and displays them using maximum intensity projection (MIP) images. The scanning of coronal sections in DWIBS decreases the number of slices required, thereby decreasing the acquisition time. In fact, diffusion tensor imaging using direct-coronal TSE-DWI has been reported.¹⁸ However, coronal section EPI-DWIBS is rarely used because of prominent distortion in the phase-encoding direction. Since transverse section EPI-DWIBS is distorted in the anteroposterior direction, it looks undistorted with the frontal image of MIP, but distortion appears in the lateral image of MIP. The minimal distortion of TSE-DWIBS may render this technique applicable to coronal section acquisition.

Adequate imaging using DWIBS requires suppression of the background signal and amplification of the contrast on the lesion signal.¹ Many background signals are caused by fat and can be suppressed using STIR. Muscle also contributes to background signals. The contrast in tumor and muscle is expected to differ between TSE-DWIBS and EPI-DWIBS because these imaging systems differ with respect to read-out method and apparent diffusion coefficients.^{12,16,19} Studies to determine

whether TSE-DWIBS is an effective imaging method have not been reported.

This study investigates the feasibility of TSE-DWIBS. We compare the contrast and SNR between TSE-DWIBS and EPI-DWIBS and evaluate the image quality between direct-coronal TSE-DWIBS, direct-coronal EPI-DWIBS, and conventional transverse EPI-DWIBS in healthy volunteers.

METHODS AND MATERIALS

Phantom study

Phantom measurements were performed on a 3 T clinical MRI scanner (Ingenia, Philips Healthcare, Best, Netherlands) using a 32-channel flexible anterior coil and an integrated posterior coil. The spherical phantom (90–401 type; NIKKO FINES, Tokyo, Japan) has two sections, one uniform and one with contrast (Figure 1). The uniform section was filled with a polyvinyl alcohol gel having 80% moisture content for measuring the SNR.

To measure the contrast ratio (CR), we made samples of physiological saline (a), physiological saline with agar and sucrose (b), and gadopentetate dimeglumine (Gd-DTPA) (Magnevist; Bayer Schering Pharma, Berlin, Germany) added to solution B (c). (a) simulates water in the body, (b) simulates muscle, and (c) simulates a malignant tumor.^{20,21} The T_1 relaxation times of (a), (b), and (c) are 4095, 2166, and 1160 ms, respectively, and the T_2 relaxation times are 1962, 99, and 88 ms, respectively. The apparent diffusion coefficients of (a), (b), and (c) are 2.2×10^{-3} , 1.5×10^{-3} , and $0.9 \times 10^{-3} \text{ mm}^2 \text{ s}^{-1}$, respectively. These samples were inserted into the contrast section of the phantom.

TSE-DWIBS and EPI-DWIBS imaging of the phantom was performed 10 times using a new shimming procedure before

Figure 1. Images of phantom contrast section (A) and uniform section (B). Samples with nine different contrasts are enclosed in the contrast section (three rows and three columns). The bottom row can be filled with samples of interest. Lower left sample, tumor-mimicking phantom (a); lower centr sample, muscle-mimicking phantom (b); and lower right sample, water (saline) (c). The SNR ratio was measured in the uniform section. Circle region of interests were used for contrast ratio measurement and dotted circle region of interests were used for SNR measurement. SNR, signal-to-noise ratio.

Figure 2. TSE-DWIBS pulse sequence diagram. DWIBS, diffusion-weighted whole-body imaging with background suppression; TSE, turbo spin-echo.

each measurement. The parameters for TSE-DWIBS are as follows: repetition time (TR), 7000 ms; echo time (TE), 52 ms; slice thickness, 5 mm; field of view (FOV), 240 mm; matrix, 96×96 ; parallel imaging (PI) factor, 4.0; b -values, 0 and 800 s mm^{-2} ; number of signal average (NSA), 3; refocusing flip angle (RFA), 80° ; and acquisition time, 1 min 24 s. The TSE-DWIBS sequence diagram is shown in Figure 2. The parameters for EPI-DWIBS were as follows: TR, 5000 ms; TE, 52 ms; slice thickness, 5 mm; FOV, 240 mm; matrix, 96×96 ; PI factor, 2.0; b -values, 0 and 800 s mm^{-2} ; NSA, 3; acquisition time, 1 min 10 s. Both sequences used gradient overplus during the motion-probing gradient (MPG).

Limitations in the specific absorption ratio were encountered during TSE-DWIBS due to the use of greater RFAs. Because an increase in the temperature of the phantom could influence the measurements, the imaging interval of the phantom scan was set to 20 min, and the fan in the bore was maximized. The air-conditioned room temperature was set to 22°C .

Volunteer study

This prospective, single-institution study was approved by the Institutional Review Board of our facility. Informed written consent was obtained from each of the 10 healthy volunteers (seven males and three females; mean age, 30.7 ± 6.5 years; range, 24–47 years). After localizer imaging, direct-coronal TSE-DWIBS (TSE-DWIBS), direct-coronal EPI-DWIBS (corEPI-DWIBS), and transverse EPI-DWIBS (traEPI-DWIBS) were used to image a 920 mm region on the caudal side, with the upper end of the cerebellum as the upper end of the imaging field of view ($320 \text{ mm} \times 3$ stations; 20 mm overlap). The same scanner used in the phantom study was used under free breathing conditions with a 15-channel head coil, a 32-channel flexible anterior coil, and an integrated posterior coil. Imaging parameter details are shown in Table 1. The MIP images of each station with a b -value of 800 s mm^{-2} were generated rotating around the left to right axis at 15 degree increments and the built-in Mobi View™

software (Philips Healthcare, Best, Netherlands) was used to stitch the MIP images (Figure 3).

Imaging analysis

Phantom study

The CR measurements were set regions of interest (ROIs) for samples (A), (B), (C). On the contrast section, the mean signal intensity was measured using three slices. The ROIs constituted 70% of the diameter of the samples. The CRs were calculated as follows: CR of X to Y = $(SI_X - SI_Y) / (SI_X + SI_Y)$, where SI_X and SI_Y are the mean signal intensity of the samples, with X and Y designating water to muscle and tumor to muscle, respectively. In the measurement of SNR, two images were used as one set among a total of 10 images, giving a total of five sets.

The subtraction images were obtained by subtracting the even-numbered images from the odd-numbered images. The SNR for each DWIBS was calculated using the subtraction method, according to the following equation: $\text{SNR} = \text{SI} / \text{SD}$, where SI is the mean signal intensity of the even-numbered images, and SD is the standard deviation of the subtraction images. SI and SD were obtained for a set circular ROI of 70% of the diameter of the phantom as the mean value of three slices of the uniform section. The SNRs were calculated for each DWIBS image obtained at a b -value of 800 s mm^{-2} .

Volunteer study

Two readers (YN and SM, with 7 and 13 years, respectively, of experience evaluating MRIs) independently assessed the MIP images for each DWIBS image on a commercial viewer (Shared Quest/ViewR; Yokogawa Medical Solutions Corporation, Tokyo, Japan). They were blinded to information regarding the acquisition sequence. The three DWIBS images were analyzed using a 5-point scale for image distortion, uniformity of fat suppression, overall artifacts, and overall image quality.^{20,22} Regarding image distortion, we referred to the spinal cord signal and scored as follows: 1, anatomical evaluation is difficult, and spinal cord

Table 1. Specific MRI protocol parameters

	TSE-DWIBS	corEPI-DWIBS	traEPI-DWIBS
Sequence	single-shot turbo spin-echo	single-shot echo planner imaging	single-shot echo planner imaging
FOV (FH × RL × AP)	320 × 450×180 mm	320 × 450×180 mm	320 × 450×320 mm
Orientation	Coronal	Coronal	
Slice thickness / gap	5.0/0 mm	5.0/0 mm	
Slices	36 per station	36 per station	
Matrix / pixel size	128 × 176/2.5 × 2.5	128 × 176/2.5 × 2.5	180 × 127/2.5 × 2.5
TE	54 ms	52 ms	
TR	shortest (about 8000 ms)	shortest (about 8000 ms)	shortest (about 14000 ms)
Phase-encoding direction	Right-left	Right-left	Anteroposterior
Pallarel imaging reduction factor	4.0	4.0	
Receiver bandwidth (Hz)	540	3155	
Refocusing flip angle 80°	80	–	
b-values	0, 800	0, 800	
k-space ordering	low-high	–	
Fat suppression methods		STIR + SSGR (Inversion time = 250 ms)	
Number of signal average	3	3	
Aquisition time	about 3:15 min per station	about 1:20 min per station	about 2:20 min per station

AP, Anterior-Posterior; DWIBS, diffusion-weighted whole-body imaging with background suppression; FH, Feet-Head; FOV, field of view; RL, Right-Left; SSGR, section select gradient reversal; STIR, short tau inversion recovery; TE, echo time; TR, repetition time; TSE, turbo spin-echo; corEPI, direct-coronal EPI-DWIBS; traEPI, transverse EPI-DWIBS.

signal is interrupted; 2, anatomical evaluation is difficult, but spinal cord signal is continuous; 3, anatomical evaluation is possible, but image distortion is moderate; 4, image distortion is slight; 5, no image distortion. The uniformity of fat suppression, as indicated by chemical shift artifacts and residual fat signals due to susceptibility artifacts, was scored as follows: 1, fat signals affect diagnosis and are found in more than three areas; 2, fat

signals affect diagnosis and are found in one or two areas; 3, fat signals do not affect diagnosis and are found in more than three areas; 4, fat signals do not affect diagnosis and are found in one or two areas; 5, fat suppression is uniform.

The overall assessment of artifacts was evaluated comprehensively, taking into account all artifacts, including image distortion

Figure 3. Example of frontal and lateral views of the maximum intensity projection (MIP) images generated from each DWIBS. (A) TSE-DWIBS (coronal-plane); (B) EPI-DWIBS (coronal-plane); and (C) EPI-DWIBS (transverse-plane). The MIP images of EPI-DWIBS generated from coronal-plane and transverse-plane show image distortion (arrows) and residual fat signals (dotted circles). DWIBS, diffusion-weighted whole-body imaging with background suppression; EPI, echo-planar imaging; IP, maximum intensity projection; TSE, turbo spin-echo.

and uniformity of fat suppression, as follows: 1, no diagnosis due to artifacts; 2, poor image due to artifacts; 3, moderate artifacts; 4, slight artifacts; 5, almost no artifacts.

The overall image quality was evaluated taking SNR and contrast into account, in addition to artifacts, as follows: 1, not diagnostic; 2, poor; 3, moderate; 4, good; 5, excellent.

Although the term “diagnosis” is used above, lesions do not exist in this study as it is considered in healthy volunteers. Therefore, “diagnosis” means that the residual fat signals are misidentified as false lesions, or it is not known whether there are lesions due to the residual fat signals or image distortion.

Demonstrative case

One patient (27 years old, female, subcutaneous and muscle hemangiomas), who provided written informed consent, was assessed using MRI scans and DWIBS as in the volunteer study. In addition, coronal T_2 weighted images were assessed. The parameters of the T_2 weighted images are as follows: TR, 40000 ms; TE, 107 ms; slice thickness, 5 mm; FOV, 320 × 340 mm; matrix, 212 × 282; PI factor, 3.0; acquisition time, 39 s per station.

Statistical analyses

A coefficient of variance (CV) was used to evaluate the repeatability of the SNR and CR obtained for each DWIBS in the phantom study. The CVs of 5 or 10 repeated measurements were calculated by dividing the standard deviation by the mean value. The results of visual assessment were compared using the Steel–Dwass test for multiple comparisons. Probability values less than 0.05 were considered statistically significant. Inter observer agreement was analyzed with weighted Cohen's κ analysis: fair (0.21–0.40), moderate (0.41–0.60), substantial (0.61–0.80), almost perfect (0.81–1.00) agreement. Statistical analyses were performed using JMP v. 11 (SAS Institute Inc., Cary, NC).

Table 2. Comparison of contrast ratio, signal-to-noise ratio, and coefficient of variance between TSE-DWIBS and EPI-DWIBS in phantom study

		Mean $\pm$ SD	CV
SNR	TSE-DWIBS	4.76 $\pm$ 0.10	0.021
	EPI-DWIBS	8.41 $\pm$ 0.09	0.011
CR of tumor to muscle	TSE-DWIBS	10.57 $\pm$ 0.45	0.043
	EPI-DWIBS	15.38 $\pm$ 0.27	0.017
CR of tumor to water	TSE-DWIBS	19.97 $\pm$ 0.67	0.034
	EPI-DWIBS	17.80 $\pm$ 0.54	0.03
CR of water to muscle	TSE-DWIBS	9.61 $\pm$ 0.66	0.069
	EPI-DWIBS	2.52 $\pm$ 0.60	0.238

CR, contrast ratio; CV, coefficient of variance; DWIBS, diffusion-weighted whole-body imaging with background suppression; EPI, echo-planar imaging; SD, standard deviation; SNR, signal-to-noise ratio; TSE, turbo spin-echo.

Scores are presented as the mean $\pm$ standard deviation.

RESULTS

Phantom study

Table 2 shows the CR, SNR, and CV determined in the phantom study. The SNR for EPI-DWIBS was about double that for TSE-DWIBS. The CR between tumor and muscle was about 1.5 times lower for TSE-DWIBS than for EPI-DWIBS. The CR between water and muscle was about four times higher for TSE-DWIBS than for EPI-DWIBS. The CV of the CR between water and muscle for EPI-DWIBS was slightly elevated (0.23) as compared to other CVs (less than 0.1), with good reproducibility.

Volunteer study

The results of qualitative analysis of the scores for image distortion, uniformity of fat suppression, overall artifacts, and overall image quality are summarized in Table 3. Good inter observer agreement was observed for image distortion, uniformity of fat suppression, overall artifacts, and overall image quality, with weighted Cohen's κ coefficients of 0.91, 0.74, 0.87, and 0.72, respectively. TSE-DWIBS had significantly less image distortion, uniformity of fat suppression, and overall artifacts than did corEPI-DWIBS and traEPI-DWIBS ($p < 0.001$). corEPI-DWIBS and traEPI-DWIBS did not differ significantly with respect to image distortion, uniformity of fat suppression, and overall artifacts; however, corEPI-DWIBS scores were lower than those of traEPI-DWIBS.

The overall image quality score for TSE-DWIBS was significantly higher than that of traEPI-DWIBS and corEPI-DWIBS. Comparison of corEPI-DWIBS and traEPI-DWIBS revealed that corEPI-DWIBS was strongly influenced by artifacts and had lower scores than traEPI-DWIBS for all visual assessments.

Demonstrative case

Figure 4 shows subcutaneous and muscle hemangiomas in the left side of pelvis. TSE-DWIBS and both EPI-DWIBS visualized these lesions. Fusion images of T_2 weighted images and TSE-DWIBS were helpful for clarifying the anatomical details of the lesions.

DISCUSSION

We observed that CR of tumor-mimicking phantom to muscle-mimicking phantom is lower for TSE-DWIBS than that for EPI-DWIBS, and that CR of water to muscle-mimicking phantom is higher for TSE-DWIBS than that for EPI-DWIBS. On the contrary, qualitative analysis scores for image distortion, uniformity of fat suppression, overall artifacts, and overall image quality were significantly higher for TSE-DWIBS than for corEPI-DWIBS or traEPI-DWIBS ($p < 0.05$).

The volunteer study revealed good inter observer agreement between TSE-DWIBS and EPI-DWIBS with respect to image distortion, uniformity of fat suppression, overall artifacts, and overall image quality. In the phantom study, we observed that the SNR was lower for TSE-DWIBS than for EPI-DWIBS images of the same voxel size and NSA. One explanation for this observation is that TSE-DWI uses phase-insensitive preparation to maintain signal uniformity, thereby eliminating deviated phase components by the MPG and decreasing the SNR.²³ As described

Table 3. Image quality scoring for TSE-DWIBS, corEPI-DWIBS, and traEPI-DWIBS as assessed by two readers in volunteer study

	Image distortion	Fat-suppression quality	Overall artifacts	Overall image quality
TSE-DWIBS				
Reader 1	4.50 ± 0.50	4.60 ± 0.49	4.30 ± 0.46	3.90 ± 0.70
Reader 2	4.10 ± 0.30	4.80 ± 0.40	4.10 ± 0.30	3.70 ± 0.46
Average	4.30 ± 0.46	4.70 ± 0.46	4.20 ± 0.40	3.80 ± 0.60
corEPI-DWIBS				
Reader 1	1.70 ± 0.64	2.20 ± 1.08	1.80 ± 0.87	2.10 ± 0.83
Reader 2	1.60 ± 0.49	1.70 ± 0.78	1.90 ± 0.70	2.60 ± 0.66
Average	1.65 ± 0.57	1.95 ± 0.97	1.85 ± 0.79	2.35 ± 0.79
traEPI-DWIBS				
Reader 1	2.20 ± 0.98	3.40 ± 0.49	2.80 ± 0.75	3.50 ± 0.50
Reader 2	1.90 ± 0.83	2.10 ± 1.04	2.10 ± 0.83	3.10 ± 0.30
Average	2.05 ± 0.92	2.75 ± 1.04	2.45 ± 0.86	3.30 ± 0.46
p-value				
TSE-DWIBS vs corEPI-DWIBS	<.001	<.001	<.001	<.001
TSE-DWIBS vs traEPI-DWIBS	<.001	<.001	<.001	0.025
corEPI-DWIBS vs traEPI-DWIBS	0.382	0.087	0.070	<.001

DWIBS, diffusion-weighted whole-body imaging with background suppression; EPI, echo-planar imaging; TSE, turbo spin-echo; corEPI, direct-coronal EPI-DWIBS; traEPI, transverse EPI-DWIBS.

Scores are reported as the mean ± standard deviation.

p-values < 0.05 were considered statistically significant.

above, the SNR of TSE-DWI is lower than that of other DWI methods; accordingly the SNR of TSE-DWIBS is lower than that of EPI-DWIBS.

The CR of tumor to muscle indicates the visibility of the lesion, whereas the CR of water to muscle indicates the degree of suppression of the water signal intensity. High water signal intensity is an obstacle for clear visualization of the lesion. In the phantom study, we observed that the higher SNR of EPI-DWIBS allows for better visibility of the lesion and greater water signal suppression than that observed for TSE-DWIBS. The difference in the RFA between these methods may contribute to this observation. Although the TE for TSE-DWIBS and EPI-DWIBS were nearly equal, TSE-DWIBS uses a variable RFA, such that the equivalent TE differs slightly.^{24,25} However, this study used low-high k-space ordering for TSE-DWIBS, thereby eliminating any large difference in equivalent TE.

A further contributing factor to the observation that EPI-DWIBS yields greater water signal suppression than TSE-DWIBS is the low refocus angle (as low as 80°) used during TSE-DWIBS in this study. TSE-DWIBS has a high specific absorption ratio because many RFAs are used, and a lower RFA can reduce the acquisition time. The lower RFA accounts for the difference in signal intensity between TSE-DWIBS and EPI-DWIBS because of the contribution of longitudinal magnetization.^{24,25} In addition, we assume that the low SNR for TSE-DWIBS also contributes to the difference in the CR. Thus, the phantom study results indicate

that EPI-DWIBS yields better images than does TSE-DWIBS. However, the opposite result was obtained in the volunteer experiment.

In the visual assessment of artifacts, image distortion, and fat suppression inhomogeneity were more prevalent in images from EPI-DWIBS than TSE-DWIBS. Furthermore, the standard deviations for these parameters were greater for EPI-DWIBS than for TSE-DWIBS. These results suggest that image distortion and fat suppression inhomogeneity of EPI-DWIBS were caused by magnetic field inhomogeneity. Thus, the uniformity of the magnetic field varies with each image in response to variables such as the physique of the patient and the magnetic shimming settings. As a result, the EPI-DWIBS image quality, which is vulnerable to magnetic field inhomogeneity, is not stable.

In overall image quality assessment, EPI-DWIBS had a high SNR and spinal cord-to-background contrast, making the assessment scores closer to those of TSE-DWIBS, but still significantly lower. corEPI-DWIBS had the shortest acquisition time but the poorest image quality. Since TSE-DWIBS has little distortion, coronal imaging is easy and can compensate for the long acquisition time, one of the disadvantages of TSE-DWIBS. Conventional EPI-DWIBS has strong image distortion, making determination of the anatomical position of small lesions difficult to determine. In the demonstrative case, the lesion is equally visible in TSE-DWIBS and T_2 weighted images, with the anatomical position of the lesion clear in each. Further studies are needed

Figure 4. Clinical case of subcutaneous and muscle hemangiomas. Lesions are noted in the left side of pelvis (arrows). (A) TSE-DWIBS (coronal-plane); (B) EPI-DWIBS (coronal-plane); (C) EPI-DWIBS (transverse-plane); and (D) fusion image of T_2 weighted and TSE-DWIBS images. DWIBS, diffusion-weighted whole-body imaging with background suppression; EPI, echo-planar imaging; TSE, turbo spin-echo.

to confirm the detection rates for lesions, because visual assessment is only used in healthy and narrow age range (24–47 years) volunteers. The results may not be well-extrapolated to patients for whom whole-body-DWI is becoming increasingly used, such as metastatic prostate and breast cancers. In conclusion, direct-coronal TSE-DWIBS has a slightly longer acquisition time

than traEPI-DWIBS, but it is robust against magnetic field inhomogeneity. Good-quality images were obtained that were free from distortion and fat suppression inhomogeneity. Our results indicate that TSE-DWIBS is useful because the image distortion and fat suppression inhomogeneity of EPI-DWIBS are conspicuous in images made with an MRI scanner at 3.0 T.

REFERENCES

1. Takahara T, Imai Y, Yamashita T, Yasuda S, Nasu S, Van Cauteren M. Diffusion weighted whole body imaging with background body signal suppression (DWIBS): technical improvement using free breathing, STIR and high resolution 3D display. *Radiat Med* 2004; **22**: 275–82.
2. Fruehwald-Pallamar J, Czerny C, Mayerhoefer ME, Halpern BS, Eder-Czembirek C, Brunner M, et al. Functional imaging in head and neck squamous cell carcinoma: correlation of PET/CT and diffusion-weighted imaging at 3 Tesla. *Eur J Nucl Med Mol Imaging* 2011; **38**: 1009–19. doi: <https://doi.org/10.1007/s00259-010-1718-4>
3. Michielsen KL, Vergote I, Dresen R, Op de Beeck K, Vanslembrouck R, Amant F, et al. Whole-body diffusion-weighted magnetic resonance imaging in the diagnosis of recurrent ovarian cancer: a clinical feasibility study. *Br J Radiol* 2016; **89**: 20160468. doi: <https://doi.org/10.1259/bjr.20160468>
4. Pearce T, Philip S, Brown J, Koh DM, Burn PR. Bone metastases from prostate, breast and multiple myeloma: differences in lesion conspicuity at short-tau inversion recovery and diffusion-weighted MRI. *Br J Radiol* 2012; **85**: 1102–6. doi: <https://doi.org/10.1259/bjr/30649204>
5. Perez-Lopez R, Mateo J, Mossop H, Blackledge MD, Collins DJ, Rata M, et al. Diffusion-weighted imaging as a treatment response biomarker for evaluating bone metastases in prostate cancer: A pilot study. *Radiology* 2017; **283**: 168–77. doi: <https://doi.org/10.1148/radiol.2016160646>
6. Sun M, Cheng J, Zhang Y, Wang F, Meng Y, Fu X. Application value of diffusion weighted whole body imaging with background body signal suppression in monitoring the response to treatment of bone marrow involvement in lymphoma. *J Magn Reson Imaging* 2016; **44**: 1522–9. doi: <https://doi.org/10.1002/jmri.25300>
7. Barnes A, Alonzi R, Blackledge M, Charles-Edwards G, Collins DJ, Cook G, et al. UK quantitative WB-DWI

- technical workgroup: consensus meeting recommendations on optimisation, quality control, processing and analysis of quantitative whole-body diffusion-weighted imaging for cancer. *Br J Radiol* 2018; **91**: 20170577. doi: <https://doi.org/10.1259/bjr.20170577>
8. Le Bihan D, Poupon C, Amadon A, Lethimonnier F. Artifacts and pitfalls in diffusion MRI. *J Magn Reson Imaging* 2006; **24**: 478–88. doi: <https://doi.org/10.1002/jmri.20683>
 9. Poustchi-Amin M, Mirowitz SA, Brown JJ, McKinsty RC, Li T. Principles and applications of echo-planar imaging: a review for the general radiologist. *Radiographics* 2001; **21**: 767–79. doi: <https://doi.org/10.1148/radiographics.21.3.g01ma23767>
 10. Mürtz P, Krautmacher C, Träber F, Gieseke J, Schild HH, Willinek WA. Diffusion-weighted whole-body MR imaging with background body signal suppression: a feasibility study at 3.0 Tesla. *Eur Radiol* 2007; **17**: 3031–7. doi: <https://doi.org/10.1007/s00330-007-0717-8>
 11. Koh DM, Blackledge M, Padhani AR, Takahara T, Kwee TC, Leach MO, et al. Whole-body diffusion-weighted MRI: tips, tricks, and pitfalls. *AJR Am J Roentgenol* 2012; **199**: 252–62. doi: <https://doi.org/10.2214/AJR.11.7866>
 12. Yoshida T, Urikura A, Shirata K, Nakaya Y, Terashima S, Hosokawa Y. Image quality assessment of single-shot turbo spin echo diffusion-weighted imaging with parallel imaging technique: a phantom study. *Br J Radiol* 2016; **89**: 20160512. doi: <https://doi.org/10.1259/bjr.20160512>
 13. Sakamoto J, Sasaki Y, Otonari-Yamamoto M, Nishikawa K, Sano T. Diffusion-weighted imaging of the head and neck with HASTE: influence of imaging parameters on image quality. *Oral Radiol* 2012; **28**: 87–94. doi: <https://doi.org/10.1007/s11282-012-0091-3>
 14. Bammer R, Augustin M, Prokesch RW, Stollberger R, Fazekas F. Diffusion-weighted imaging of the spinal cord: interleaved echo-planar imaging is superior to fast spin-echo. *J Magn Reson Imaging* 2002; **15**: 364–73. doi: <https://doi.org/10.1002/jmri.10075>
 15. De Foer B, Vercruyse JP, Pilet B, Michiels J, Vertriest R, Pouillon M, et al. Single-shot, turbo spin-echo, diffusion-weighted imaging versus spin-echo-planar, diffusion-weighted imaging in the detection of acquired middle ear cholesteatoma. *AJNR Am J Neuroradiol* 2006; **27**: 1480–2.
 16. Baltzer PA, Renz DM, Herrmann KH, Dietzel M, Krumbein I, Gajda M, et al. Diffusion-weighted imaging (DWI) in MR mammography (MRM): clinical comparison of echo planar imaging (EPI) and half-Fourier single-shot turbo spin echo (HASTE) diffusion techniques. *Eur Radiol* 2009; **19**: 1612–20. doi: <https://doi.org/10.1007/s00330-009-1326-5>
 17. Zhang Z, Zhang B, Li M, Liang X, Chen X, Liu R, et al. Multishot cartesian turbo spin-echo diffusion imaging using iterative POCMUSE Reconstruction. *J Magn Reson Imaging* 2017; **46**: 167–74. doi: <https://doi.org/10.1002/jmri.25522>
 18. Sakai T, Doi K, Yoneyama M, Watanabe A, Miyati T, Yanagawa N. Distortion-free diffusion tensor imaging for evaluation of lumbar nerve roots: Utility of direct coronal single-shot turbo spin-echo diffusion sequence. *Magn Reson Imaging* 2018; **49**: 78–85. doi: <https://doi.org/10.1016/j.mri.2018.01.003>
 19. Babourina-Brooks B, Cowin GJ, Wang D. Diffusion-weighted imaging in the prostate: an apparent diffusion coefficient comparison of half-Fourier acquisition single-shot turbo spin-echo and echo planar imaging. *Magn Reson Imaging* 2012; **30**: 189–94. doi: <https://doi.org/10.1016/j.mri.2011.09.024>
 20. Jacobsen MC, Dogan BE, Adrada BE, Plaxco JS, Wei W, Son JB, et al. 3-T breast diffusion-weighted MRI by echo-planar imaging with spectral spatial excitation or with additional spectral inversion recovery: an in vivo comparison of image quality. *J Comput Assist Tomogr* 2015; **39**: 343. doi: <https://doi.org/10.1097/RCT.0000000000000223>
 21. Watanabe H, Kanematsu M, Goshima S, Kondo H, Onozuka M, Moriyama N, et al. Staging hepatic fibrosis: comparison of gadoxetate disodium-enhanced and diffusion-weighted MR imaging--preliminary observations. *Radiology* 2011; **259**: 142–50. doi: <https://doi.org/10.1148/radiol.10100621>
 22. Lai LM, Cheng JY, Alley MT, Zhang T, Lustig M, Vasanawala SS, SSJJoMRI V. Feasibility of ferumoxyl-enhanced neonatal and young infant cardiac MRI without general anesthesia. *J Magn Reson Imaging* 2017; **45**: 1407–18. doi: <https://doi.org/10.1002/jmri.25482>
 23. Alsop DC. Phase insensitive preparation of single-shot RARE: application to diffusion imaging in humans. *Magn Reson Med* 1997; **38**: 527–33. doi: <https://doi.org/10.1002/mrm.1910380404>
 24. Alsop DC. The sensitivity of low flip angle RARE imaging. *Magn Reson Med* 1997; **37**: 176–84. doi: <https://doi.org/10.1002/mrm.1910370206>
 25. Busse RF, Hariharan H, Vu A, Brittain JH. Fast spin echo sequences with very long echo trains: design of variable refocusing flip angle schedules and generation of clinical T2 contrast. *Magn Reson Med* 2006; **55**: 1030–7. doi: <https://doi.org/10.1002/mrm.20863>