MAJOR PAPER

Diffusion-weighted Imaging of the Abdomen during a Single Breath-hold Using Simultaneous-multislice Echo-planar Imaging

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Purpose: This multi-scanner study aimed to investigate the validity of single breath-hold (BH) diffusion-weighted imaging (DWI) using simultaneous-multislice (SMS) echo-planar imaging in multiple abdominal organs to enable faster acquisition and reliable quantification of apparent diffusion coefficient (ADC).

Methods: SNR, geometric distortion (GD), and ADC in a phantom; the ADC in the liver, renal cortex, paraspinal muscle, spleen, and pancreas; and the signal intensity ratio of the portal vein-to-muscle (SIR_{PV-M}) in healthy volunteers were compared between BH- and respiratory-triggered (RT) DWI with b-values of 0 and 800 s/mm² in two different MRI scanners.

Results: The phantom study showed that the SNR of BH-DWI was significantly lower than that of the RT-DWI ($P < 0.05$ for both scanners), whereas the GD and ADC of BH-DWI did not differ significantly from those of the RT-DWI ($P = 0.09-0.60$). In the volunteer study, the scan times were 23 seconds for BH-DWI and 184 ± 33 seconds for RT-DWI, respectively. The ADC of the liver in BH-DWI was significantly lower than that in RT-DWI ($P < 0.05$ for both scanners), whereas there were no significant differences in the ADCs of the renal cortex, paraspinal muscle, spleen, or pancreas between BH-DWI and RT-DWI ($P = 0.07-0.86$). The SIR_{PV-M} in BH-DWI was significantly smaller than in RT-DWI ($P < 0.05$ for both scanners).

Conclusion: The proposed method enables the acquisition of abdominal diffusion-weighted images in a single BH.

Keywords: *apparent diffusion coefficient, breath-holding, diffusion-weighted imaging, echo-planar imaging, simultaneous-multislice*

Introduction

Diffusion-weighted imaging (DWI) of the abdomen has been widely used to detect and characterize abdominal lesions, as well as to evaluate the treatment response of the lesions.¹ The apparent diffusion coefficient (ADC) calculated from DWI with two or more b-values provides quantitative information

regarding the diffusivity of water molecules in tissues, whereas most acquired images from other sequences in abdominal MRI are qualitative. Abdominal DWI has been commonly performed with either free breathing or respiratory-triggered (RT) approaches.² However, with either approaches, low image quality and misregistration between b-values resulting from unstable respiration or false triggering can strongly affect ADC measurement.^{3,4}

In clinical DWI, simultaneous-multislice (SMS) acquisition has been introduced as an acceleration technique in which multiple slices can be simultaneously excited by a multiband radiofrequency pulse and acquired with a shorter TR and sufficient image quality as compared with the standard technique.⁵ SMS offers two distinct advantages over non-SMS DWI, as follows: a shorter scan time and a higher number of signals averaged (NSA) without increased scan time. The former can reduce the DWI scan time and improve the efficiency of daily MRI workflow in tight examination schedules. The latter can increase the SNR within the same scan time and improve the image quality as compared with

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Fig. 1 Examples of ROIs placed on an apparent diffusion coefficient map of the phantom.

non-SMS DWI. Recently, the application of SMS acquisition has been expanded to include abdominal scanning, including the liver, kidney, and pancreas.^{6–9} Pei et al. reported the feasibility of single breath-hold (BH) DWI with SMS in the liver and the reliability of ADC quantification in healthy subjects capable of adequate breath-holding.¹⁰ However, this technique has not been well evaluated in different organs, scanners, or practical protocols.

The purpose of this study is to investigate the validity of single BH-DWI with SMS echo-planar imaging (EPI) for faster acquisition and reliable quantification of ADC in multiple abdominal organs by comparing with RT-DWI. Moreover, the study was performed with two different MRI scanners to demonstrate the versatility of a single BH-DWI with SMS.

Materials and Methods

Preparation of phantom

We conducted a phantom study using a quantitative imaging biomarker alliance diffusion phantom (High Precision Devices, Boulder, CO, USA) to assess SNR, geometric distortion (GD), and ADC of the phantom. The phantom consisted of 13 vials with various concentrations of polymer polyvinylpyrrolidone (PVP) in aqueous solution (Fig. 1). The PVP concentration of each vial was as follows: 0% (No. 1–3), 10% (No. 4–5), 20% (No. 6–7), 30% (No. 8–9), 40% (No. 10–11), and 50% (No. 12–13). The inside of the phantom was filled with purified water. To eliminate the variability in the temperature of the phantom during the scan, the phantom was placed in the MRI scanner room

24 hours before the scan was initiated. The phantom temperature was equilibrated at a temperature in the scanner room (22°C) to resemble the range of ADCs in different abdominal organs and tumors (e.g., $0.7 \times 10^{-3} \text{ mm}^2/\text{s}$ for poorly differentiated hepatocellular carcinoma, $1.2 \times 10^{-3} \text{ mm}^2/\text{s}$ for normal liver, and $2.0 \times 10^{-3} \text{ mm}^2/\text{s}$ for normal renal cortex).^{9,11} We measured the inner phantom temperature before and after the scan using a digital thermometer (1227U09; Thomas Scientific, Swedesboro, NJ, USA) and confirmed that the variability was less than 1.0°C.

Scan protocol of the phantom study

The phantom was scanned with BH- and RT-DWI protocols on two different clinical 3.0-T MRI scanners (scanner A, Ingenia; Philips Healthcare, Best, The Netherlands and scanner B, Signa Premier; GE Healthcare, Milwaukee, WI, USA) using a 16-channel phased-array anterior coil in combination with a 16-channel posterior table coil for scanner A and a 30-channel anterior array coil in combination with a 60-channel posterior table coil for scanner B. During BH-DWI, we incorporated the SMS technique with an SMS acceleration factor of 2 into DWI, with an NSA of 2 to enable the acquisition of the upper abdominal DWI data within 23 seconds, which is practical for breath-holding. In addition, variable-rate selective excitation pulses with time-varying slice-selective gradients were used in the prototype sequence of scanner A for BH-DWI to reduce the duration of excitation pulses (i.e., increase the bandwidth of excitation pulses), which can shorten the TE and minimize water suppression in slice-selective gradient reversal-based fat suppression, resulting in improved image quality.¹² We used an NSA of 4 in RT-DWI, which is commonly used for liver DWI.⁶ The TR was set at 5000 ms, which simulated a normal respiration cycle and differed from the TR in BH-DWI (2500 ms). We kept the other imaging parameters constant between the protocols as follows: pulse sequence, single-shot diffusion EPI; TE, 55 ms; slice thickness, 6 mm; slice gap, 0 mm; the number of slices, 30 (which was enough to completely cover at least the whole liver); FOV, 360 mm; imaging matrix, 128×128 ; parallel imaging factor, 2; half-scan factor, 0.7; diffusion gradients, three orthogonal directions; and b-values, 0 and 800 s/mm^2 . The scan times were 23 seconds and 85 seconds for BH- and RT-DWI, respectively. Two scans with identical imaging parameters were performed in both protocols to measure the SNR using the subtraction method.¹³ We obtained transverse images of the phantom, and subsequently created the ADC images. Moreover, we performed T_1 mapping on scanner A to evaluate the effect of T_1 saturation for the short TR in BH-DWI on the ADC measurements. A T_1 map was obtained using an inversion recovery spin-echo EPI sequence with the following imaging parameters: TR, 10,000 ms; TE, 60 ms; slice thickness, 5 mm; FOV, 256 mm; and imaging matrix, 128×128 ; parallel imaging factor, 2; and inversion times, 50, 100, 200, 400, 800, 1600, 3200, and 4000 ms.

Fig. 2 Examples of ROIs placed on b0 images in a healthy volunteer: blue, liver; orange, spleen; pink, portal vein; dark green, right paraspinal muscle; purple, left paraspinal muscle; red, right renal cortex; light green, left renal cortex; and dark red, pancreas.

Evaluation of the phantom study

The SNR, GD, ADC, and T_1 relaxation time were measured manually in ROIs placed on each vial of the phantom. We used the subtraction method to determine the SNR on the b800 image.¹³ We obtained a different image by subtracting two identical images. The noise was determined by dividing the standard deviation in the ROI on the difference image by an appropriate correction factor. The SNR was then calculated by dividing the mean signal intensity in the ROI on the b800 image by the estimated noise. The GD in the phantom was measured based on the standard of the National Electrical Manufacturers Association.¹⁴ The diameters of each vial were measured at four different angles of 0° , 45° , 90° , and 135° . The GD (in percentage) was calculated using the following equation:

$$GD = \text{Maximum}(|D_M - D_A|/D_A) \cdot 100 \quad (1)$$

where D_M and D_A are the measured and actual diameters of the vial, respectively.

Healthy volunteer study

Nine healthy subjects (all men, 23.2 ± 0.8 years of age) participated in this study. The inclusion criteria were subjects aged 20–30 years (to exclude the possible age-related bias in ADC of the liver, kidney, and spleen^{15–17}) with no known history of abdominal disease. This study was approved by the appropriate institutional review board. The purpose and procedures of the study were fully explained to all subjects, who provided their written informed consent before undergoing scans.

The healthy subjects were scanned using the same two 3.0-T MRI scanners as in the phantom study, with BH- and RT-DWI. Transverse abdominal images were obtained with the same protocols as the phantom study, except for TR in

RT-DWI: the TR changed depending on the subject's respiration cycle. ROIs were manually drawn in the liver (the liver parenchyma of the right lobe), renal cortex, paraspinal muscle, spleen, and pancreas (pancreatic tail) on the b0 images by a single observer (N.O.). Subsequently, all ROIs were copied and pasted on the corresponding ADC maps, and the ADC values were measured. To avoid large vessels, subcutaneous fat, and any artifacts from cardiac pulsation and intestinal gas, we carefully focused on the ROIs. We generated Bland–Altman plots to evaluate the measurement agreement of ADC between the BH- and RT-DWI protocols. In addition, the mean signal intensities in the ROIs inside the portal vein (right branch) and right paraspinal muscle were measured on the b0 image. We then compared the signal intensity ratio of the portal vein-to-muscle (SIR_{PV-M}) between the BH- and RT-DWI to evaluate the T_1 saturation effect of the blood due to the short TR in BH-DWI. A representative example of ROI placement is shown in Fig. 2.

Statistical analyses

We performed the statistical analyses using SPSS for Windows, version 25 (IBM, Armonk, NY, USA). We compared the SNR, GD, and ADC of the phantom, the ADC in the different organs of the healthy subjects, and the SIR_{PV-M} between both protocols using the Wilcoxon signed-rank test. A P value of < 0.05 was considered to be statistically significant.

Results

Phantom study

Table 1 displays the T_1 , SNR, GD, and ADC values of BH-DWI and RT-DWI of the phantom in scanners A and B. The SNR of BH-DWI was significantly lower than that of RT-DWI ($P < 0.05$ for both scanners), with a mean SNR reduction of $76\% \pm 5\%$ for scanner A and $45\% \pm 21\%$ for scanner B. However, the GD and ADC of BH-DWI did not differ

Table 1 T_1 relaxation time of the phantom and comparisons between BH- and RT-DWI for SNR, GD, and ADC of the phantom in scanners A and B.

Vial No.	PVP (%)	T_1 (ms)	Scanner A						Scanner B					
			SNR		GD (%)		ADC ($\times 10^{-3}$ mm ² /s)		SNR		GD (%)		ADC ($\times 10^{-3}$ mm ² /s)	
			BH	RT	BH	RT	BH	RT	BH	RT	BH	RT	BH	RT
1	0	2980	49	210	5	4	2.20	2.25	73	217	5	2	2.29	2.29
2	0	2946	43	180	4	4	2.22	2.26	78	190	7	5	2.30	2.28
3	0	2857	27	169	7	8	2.33	2.29	43	167	8	8	2.44	2.36
4	10	2238	94	413	10	9	1.74	1.74	160	411	8	10	1.82	1.81
5	10	2230	74	325	7	7	1.79	1.77	144	250	7	7	1.93	1.89
6	20	1773	122	439	12	11	1.31	1.33	259	517	13	11	1.37	1.38
7	20	1752	112	400	9	10	1.37	1.36	203	547	8	8	1.47	1.44
8	30	1228	84	387	12	12	0.92	0.93	326	527	12	11	0.96	0.96
9	30	1180	76	369	10	12	0.93	0.93	195	451	13	10	1.03	0.99
10	40	779	90	486	10	11	0.61	0.61	332	829	9	7	0.62	0.63
11	40	775	64	542	9	6	0.59	0.61	345	553	6	6	0.65	0.64
12	50	449	65	586	1	8	0.34	0.34	536	694	8	7	0.36	0.36
13	50	447	56	426	4	4	0.34	0.35	530	664	7	7	0.37	0.37
<i>P</i> value (BH vs. RT)			< 0.05		0.09		0.2		< 0.05		0.6		0.09	

ADC, apparent diffusion coefficient; BH, breath-hold; DWI, diffusion-weighted imaging; GD, geometric distortion; PVP, polymer polyvinylpyrrolidone; RT, respiratory-triggered.

significantly from those of RT-DWI ($P = 0.09$ – 0.60). The linearity of ADC between BH- and RT-DWI was also demonstrated by linear regression analysis (Fig. 3). Figure 4 shows the representative diffusion-weighted and ADC images of the phantom with both protocols.

Healthy volunteer study

In the volunteer study, the scan times were 23 seconds for BH-DWI and 184 ± 33 seconds for RT-DWI, respectively. Table 2 presents the comparisons of the ADC between BH-DWI and RT-DWI in healthy volunteers in scanners A and B. The ADC of the liver in BH-DWI was significantly lower than that in RT-DWI ($P < 0.05$ for both scanners). The mean differences in the ADCs of the liver between both protocols were -0.14×10^{-3} mm²/s (9.6%) for scanner A and -0.13×10^{-3} mm²/s (9.3%) for scanner B. However, we found no significant differences in the ADCs of the renal cortex, paraspinal muscle, spleen, or pancreas between BH- and RT-DWI ($P = 0.07$ – 0.86). Figure 5 shows the Bland–Altman plots of the ADC in the liver and other abdominal regions, which demonstrate no apparent systematic bias for the ADC. The SIR_{PV-M} in BH-DWI (2.4 ± 1.0 and 2.3 ± 0.5 for scanners A and B, respectively) was significantly smaller than that in RT-DWI (4.0 ± 1.2 and 4.3 ± 0.9 for scanners A and B, respectively, with $P < 0.05$ for both scanners; Fig. 6). Figure 7

presents the representative diffusion-weighted and ADC images in scanners A and B, which show a slight reduction in SNR and the disappearance of blood in the liver on the b0 and ADC images.

Discussion

The present study validated the feasibility of a single BH-DWI with SMS covering the upper abdominal organs in the phantom and healthy subjects. In the phantom study, the SNR of BH-DWI was significantly lower than that of RT-DWI, whereas we found no significant differences in the GD and ADC between BH- and RT-DWI. There was a significantly lower ADC of the liver in BH-DWI as compared with RT-DWI in the healthy volunteer study, whereas there were no significant differences in the ADCs of the other regions between BH- and RT-DWI.

In clinical practice, both free breathing and RT-DWI of the abdomen are widely used; however, unstable respiration or false triggering can reduce the image quality and result in the misregistration between b-values. These disadvantages often cause a low accuracy of ADC measurement. To overcome these issues, Nasu et al. demonstrated the feasibility of BH-DWI of the liver for reducing the motion-related artifacts.¹⁸ However, the scan time used in their protocol is

Fig. 3 Plots showing the ADC of the phantom with BH- versus RT-DWI in scanners A (a) and B (b). Results of the linear regression analysis are summarized in the plot. ADC, apparent diffusion coefficient; BH-DWI, breath-hold diffusion-weighted imaging; RT-DWI, respiratory-triggered diffusion-weighted imaging.

Fig. 4 Representative diffusion-weighted images ($b = 0$ and 800 s/mm^2) and ADC maps of the phantom with BH- and RT-DWI in scanners A and B. ADC, apparent diffusion coefficient; BH-DWI, breath-hold diffusion-weighted imaging; RT-DWI, respiratory-triggered diffusion-weighted imaging.

Table 2 Comparisons between BH- and RT-DWI in the ADC of different upper abdominal regions in scanners A and B. Mean $\pm$ standard deviation and *P* values are shown.

	Scanner A			Scanner B		
	ADC ($\times 10^{-3}$ mm ² /s)			ADC ($\times 10^{-3}$ mm ² /s)		
	BH	RT	<i>P</i>	BH	RT	<i>P</i>
Liver	1.31 $\pm$ 0.08	1.45 $\pm$ 0.14	< 0.05	1.33 $\pm$ 0.11	1.47 $\pm$ 0.09	< 0.05
Right renal cortex	1.98 $\pm$ 0.07	2.02 $\pm$ 0.13	0.26	2.14 $\pm$ 0.11	2.29 $\pm$ 0.41	0.14
Left renal cortex	2.15 $\pm$ 0.12	2.13 $\pm$ 0.15	0.86	2.15 $\pm$ 0.10	2.27 $\pm$ 0.11	0.09
Right paraspinal muscle	1.74 $\pm$ 0.06	1.66 $\pm$ 0.13	0.11	1.88 $\pm$ 0.06	1.81 $\pm$ 0.06	0.07
Left paraspinal muscle	1.79 $\pm$ 0.09	1.74 $\pm$ 0.15	0.14	1.84 $\pm$ 0.08	1.81 $\pm$ 0.09	0.68
Spleen	0.88 $\pm$ 0.13	0.85 $\pm$ 0.18	0.21	0.88 $\pm$ 0.07	0.90 $\pm$ 0.15	0.37
Pancreas	1.48 $\pm$ 0.14	1.62 $\pm$ 0.20	0.11	1.53 $\pm$ 0.19	1.70 $\pm$ 0.24	0.09

ADC, apparent diffusion coefficient; BH, breath-hold; DWI, diffusion-weighted imaging; RT, respiratory-triggered.

Fig. 5 Bland–Altman plots of ADC in the liver (**a** and **b**) and other abdominal regions (**c** and **d**) with BH- and RT-DWI in scanners A (**a** and **c**) and B (**b** and **d**). Black solid line, mean difference (bias); black dashed lines, upper and lower LOA (bias $\pm$ two standard deviation). ADC, apparent diffusion coefficient; BH-DWI, breath-hold diffusion-weighted imaging; LOA, limits of agreement; RT-DWI, respiratory-triggered diffusion-weighted imaging.

not appropriate for obtaining whole-liver diffusion-weighted images in a single BH with sufficient image quality, and split acquisition into multiple breath holds is required. The inherent inconsistency between multiple acquisitions leads to the

misalignment of imaging slices. Thus, multiple BH acquisition is a technical challenge. In this study, we aimed to complete the scan of the entire upper abdomen diffusion-weighted images in a single BH.

Fig. 6 Comparisons of SIR_{PV-M} in b_0 images between the BH- and RT-DWI in scanners A (a) and B (b). BH-DWI, breath-hold diffusion-weighted imaging; RT-DWI, respiratory-triggered diffusion-weighted imaging; SIR_{PV-M} , signal intensity ratio of the portal vein-to-muscle.

Fig. 7 Representative diffusion-weighted images ($b = 0$ and 800 s/mm^2) and ADC maps of the abdomen with BH-DWI and RT-DWI in scanners A and B. ADC, apparent diffusion coefficient; BH-DWI, breath-hold diffusion-weighted imaging; RT-DWI, respiratory-triggered diffusion-weighted imaging.

In this study, the results of the phantom study indicate that the significant SNR penalty associated with the substantial reduction in scan time in BH-DWI does not affect the GD of images or the accuracy of ADC quantification. Although

ADC can be biased in insufficient SNR, leading to the underestimation of the ADC,¹⁹ the observed consistency of the ADC between both protocols supports the notion that the SNR in BH-DWI is adequate for accurate ADC estimation.

A potential concern of the proposed method is the short TR. The SMS technique incorporated into BH-DWI can reduce the scan time by shortening the TR through the simultaneous excitation of multiple slices. However, the shortened TR can introduce incomplete recovery of longitudinal magnetization (i.e., the T_1 saturation effect), leading to signal saturation. The effect can be greater in tissues with a longer T_1 value (e.g., blood and cystic lesions). In addition, the effect of signal saturation on image contrast can be more pronounced in b_0 images that are acquired at the beginning of the sequence, as longitudinal magnetization can be strongly affected by the excitation pulse at the short TR. In contrast, because the longitudinal magnetization approaches steady state, it can be mitigated on b_{800} images acquired in the latter. Consequently, these effects could have contributed to the difference in the appearance of b_0 images and the visual similarity of b_{800} images between both protocols. Moreover, the discrepancy in the T_1 saturation effect between images with different b -values potentially causes underestimation of the measured ADC. Some vials in the phantom exhibited longer T_1 values over the physiological ranges for abdominal organs (e.g., 809 ms for liver and 1142 ms for renal cortex);²⁰ hence, we expected the corresponding ADC values in BH-DWI to be more susceptible to the T_1 saturation. Nonetheless, the consistency we observed between BH- and RT-DWI in a wide range of ADCs covering the physiological values in abdominal organs and tumors supports the concept that BH-DWI can be used to obtain the ADC with accuracy comparable to that of RT-DWI. In other words, the TR of 2500 ms used in this study was demonstrated to be within the acceptable range for accurate ADC measurement in the abdominal organs. This may be attributed to the dummy scans performed before the actual data acquisition, which bring the longitudinal magnetization close to steady state. This study differs from a previous study¹⁰ in that we expanded the application of BH-DWI to various upper abdominal organs covered in a single BH acquisition and validated the feasibility in the phantom study with more practical imaging conditions and multi-scanners.

The results from the healthy subjects showed that the ADC values of the liver in BH-DWI were significantly lower than those in RT-DWI, whereas we found no significant differences in the ADC of abdominal organs except for the liver. A possible explanation for the difference in ADC could be the shorter TR (2500 ms) used in BH-DWI as compared with RT-DWI (3000–5000 ms depending on the subject's respiration cycle). The lower ADC of the liver in BH-DWI may be attributed to the weakened signal intensity of blood, especially on the b_0 images: the short TR can weaken blood signals because of the T_1 saturation effect for the long T_1 (1800 ms),²¹ which may have contributed to the signal intensity changes in the liver on the b_0 image. This hypothesis can be explained by the result that the SIR_{PV-M} in the b_0 images was lower in BH-DWI than in RT-DWI.

Meanwhile, the influence of T_1 saturation for blood signals is negligible in the b_{800} images because the blood signals can be completely suppressed by the diffusion sensitizing gradients. Consequently, the difference in the T_1 saturation effect of the blood signals between the b_0 and b_{800} images resulted in the lower ADC of the liver. This is supported by the fact that the blood signals in the liver appeared darker on the b_0 and ADC images in BH-DWI as compared with those in RT-DWI. On the other hand, the bias of liver ADC between BH- and RT-DWI was smaller than the variation due to scanner differences (13.8%).²² Furthermore, the contribution of blood signals to the DWI signal decay is likely to be larger in the liver as compared with other abdominal organs (45% for liver vs. 20% for kidney and 9% for spleen), which indicates that the ADC in the liver is more susceptible to blood signals.²³ This might also explain the significant difference in ADC observed only in the liver. Our results were not consistent with those reported in the study by Pei et al.,¹⁰ in which a good agreement of liver ADC between BH- and RT-DWI was demonstrated. In contrast to our present study, the work by Pei et al.¹⁰ used a relatively similar TR between BH-DWI (1800 ms) and RT-DWI (1500 ms), despite the fact that there are questions regarding the TR of RT-DWI, because it was supposed to be one respiration cycle (normally 3000–6000 ms) or longer in RT-DWI. In this case, we expected the effect of T_1 saturation to be comparable between both protocols. This makes it difficult to clarify the impact of the shorter TR in BH-DWI with SMS on the image quality and quantification of ADC. In addition, the use of a higher minimum b -value (50 s/mm²) in their study potentially reduces the contribution of blood signals to the measurement of ADC. These factors could explain the discrepancy between the present and previous studies. By using near-identical imaging parameters, we revealed the comparable results between different MRI scanners (A and B) in both phantom and healthy subject studies, demonstrating the versatility of BH-DWI with SMS.

Given these results, BH-DWI has the potential to serve as an alternative to RT-DWI for abdominal DWI. This could have significant clinical implications for abdominal MRI, especially in patients with an irregular breathing pattern. In such patients, BH-DWI would be a preferred approach to achieving sufficient image quality and accurate ADC measurement.^{18,24} The results of the present study demonstrated that BH-DWI with SMS achieved an acquisition with upper abdominal coverage in a single BH while maintaining sufficient image quality and accuracy of ADC quantification. In addition, abbreviated MRI has received great interest from radiologists and clinicians, as it potentially improves MRI workflow, reduces costs, and meets the growing clinical demands for MRI.²⁵ Thus, RT-DWI, which typically requires 3 to 5 mins, in standard protocols could be replaced with BH-DWI in the abbreviated abdominal protocols with a total scan time of less than 10 mins. Although the relatively long duration of BH (23 seconds) in BH-DWI was not

disruptive in healthy subjects, a reduction in BH duration (< 15 seconds) for patients with respiratory diseases can be achieved by further reducing the NSA at a tradeoff against further SNR penalty.

This study has some limitations. First, this research was based on only a limited number of healthy young volunteers with an adequate ability for breath-holding. However, given that patients of different ages with various abdominal diseases are examined in clinical practice, the clinical usefulness of the proposed method remains unknown. The SNR penalty in BH-DWI might affect the detectability of small lesions. Thus, based on our preliminary results in both the phantom and healthy subjects, to demonstrate the validity of our study, we emphasize that further research should be pursued with a larger number of healthy subjects and patients with abdominal lesions. Second, the choice of b-values potentially affects the quantification of ADC. In highly perfused tissues including the liver, blood perfusion induces rapid signal attenuation especially at a low range of b-value (0–100 s/mm²), resulting in an overestimation of the ADC. Although we used a combination of b-values of 0 and 800 s/mm² in this study, a higher minimum b-value (> 0 s/mm²) instead of b = 0 s/mm² would enable the suppression of the perfusion effect in tissues and result in a more reliable ADC quantification.²⁶ In addition, further studies should be conducted to demonstrate these consistencies at 1.5T, in which either or both a shortening of the T₁ of tissues and further reduction in the SNR potentially affect ADC quantification.

Conclusion

The BH-DWI with the SMS technique reduced the scan time by approximately eightfold in healthy subjects and demonstrated accuracy comparable with that of ADC quantification with conventional RT-DWI. The proposed method enables the acquisition of abdominal diffusion-weighted images in a single breath-hold.

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Conflicts of Interest

Yu Ueda is an employee of Philips Japan Ltd. The other authors declare no conflicts of interest.

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